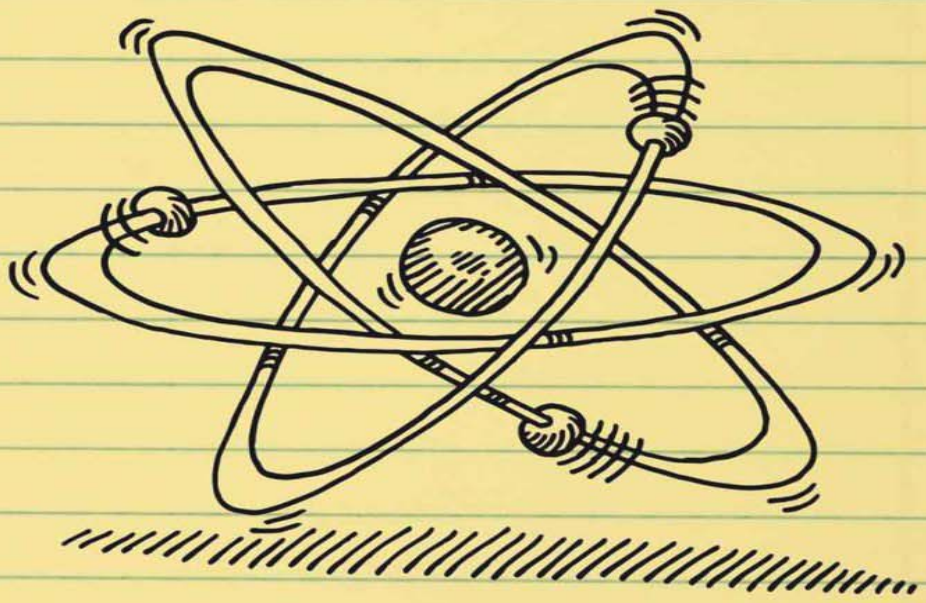
A Radiologist's Notes on Physics



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A Radiologist's Notes on Physics

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Preface

Radiology is a technically challenging subject. The principles that underpin our everyday work are complex and myriad but are fundamental to good clinical radiological practice. It is right and proper that the Royal College of Radiologists continues to teach and assess these concepts in the form of the physics component of the FRCR examination. What is wrong is the ongoing lack of clear and concise learning material to grasp these concepts.

Whilst preparing for the physics module back in 2010, I realised that although there are established texts on medical physics, none of these provide concise explanations in plain English. I spent months distilling the information in popular textbooks, the R-ITI project, local teaching sessions and even A-level physics lecture material from school into legible and easily comprehensible revision notes.

Fast-forward to 2014 and I find that the situation is unchanged. There still exists no clear set of physics revision notes in plain English for radiologists. That is, of course, until now. In their early form, these notes have been used by my peers on my local training scheme over the past 3 years with much praise. In fact, it's due to feedback from other trainees that I decided to revisit, re-craft and redesign them for a wider audience.

This book is primarily aimed at clinical radiology trainees preparing to sit the physics module of the Royal College of Radiologists' First FRCR examination. I hope it will also serve as an aide-memoire for more senior registrars and consultants as well as those from other specialities, such as radiographers or indeed any other curious individual.

The structure of the book mirrors the RCR physics syllabus. In fact, the chapter titles are the same as the curriculum headers. The notes are principally in bullet-form with exam favourites highlighted in bold. Crucial equations and other concepts are clearly indicated. There more than 75 illustrations and graphs to help explain complex concepts such as MR imaging.

I really hope that you find the information in this book helpful. I certainly learned a lot writing it. If you're sitting the exam soon, I wish you the best of luck.

Garry Pettet
January 2014

About the author

I always like to know a little bit about the chap trying to teach me something so it seems only fair that I should tell you a few things about myself.

I qualified from Imperial College School of Medicine with distinction in 2005. I completed Foundation training in the south west of England. Following this, I spent some time acquiring a comprehensive set of clinical skills before my career in radiology; spending eighteen months in emergency medicine in Australia followed by a further eighteen months of core surgical training in the UK. I'm currently a fourth year radiology registrar in Bristol with an interest in paediatric imaging.

I passed the FRCR examination in October 2013 at my first attempt. I have never failed an exam.

I'm very passionate about teaching and training. At the time of writing (January 2014) I'm the current Chairman of the Junior Radiologist's Forum for the Royal College of Radiologists and I help to represent the views of UK trainees. I spend a lot of time teaching junior registrars and medical students and I enjoy reading and writing about medicine in general.

Outside of medicine/radiology, I write software and design pretty websites. I also enjoy photography, particularly wildlife and landscapes.

I have a daughter, Aoife and a very tolerant wife, Fiona. It is to both of them that I dedicate this book.

Any and all feedback on the book is greatly valued. I can be easily tracked down via my website http://garrypettet.com.

Chapter 1

Matter & radiation

Structure of matter, the atom and the nucleus

Atoms

- Atoms are the smallest unit of an element that still retain the chemical and physical properties of the element
- Mostly **empty space** with its mass concentrated in the **central nucleus**
- The nucleus contains nucleons (**protons** & **neutrons**)
- Protons are positively charged, neutrons have zero charge
- Atoms are electrically neutral (number of protons = number of electrons)

Atomic mass (**A**) = number of protons + neutrons
Atomic number (**Z**) = number of protons

Nucleons are held together by short range forces. The neutrons in the nucleus help reduce the repulsive forces of the positively charged protons as they "space" them out. When there are less than about 25 protons in a nucleus, there are the same number of neutrons. As nuclei get heavier ($Z > 25$), the relative number of neutrons needs to increase to counteract the increased electrostatic repulsive forces.

Nuclides

- A nuclide is an atomic species characterised by its number of protons (Z) and neutrons
- Nuclides with the same number of protons are the same element
- Radioactive nuclides are called **radionuclides**

Isotopes

- Radioactive nuclides are called radionuclides
- Isotopes have the same chemical properties but different physical properties

- Radioactive isotopes are called **radioisotopes**

Electrons & their shells

- Electrons are negatively charged particles, much smaller than protons & neutrons
- Electrons orbit the nucleus, like planets around the sun, in specific **shells**
- The **innermost shell is K**. They are then labelled sequentially, e.g. K, L, M, etc
- Each shell can only hold a fixed number of electrons (K = 2, L = 8, M = 18)
- Each shell must fill completely before the next outer one can be filled. The innermost shell is filled first because it has the lowest energy
- The outermost (**valence**) shell determines an element's chemical, electrical and thermal properties
- An atom is in its **ground state** when all of its electrons are in the lowest energy shells

Electron movement between shells

Electrons can only move to another shell if:

- There is a **vacancy** and they **gain or lose** the **exact amount of energy** required to give them the correct energy for that shell
- An electron can gain energy by:
  - Thermal vibration
  - Interaction with another charged particle
  - Absorption of a photon that has an **energy equal to the energy difference between the two shells**

Ionisation

- Atoms become ions when an electron escapes the electrostatic attraction of the nucleus

- An ion has an **unequal number** of protons and electrons

Excitation

- An atom is excited when an electron gains energy and moves to a higher energy shell and leaves a gap in a lower shell
- Excited atoms always try to return to the ground state. This is done by an electron "dropping" down into the gap in the lower energy shell and emitting the extra energy as **electromagnetic radiation**
- Vacancies in a shell are most likely to be filled by an electron from the next shell out

Binding energy (E)

- This is the energy expended to completely remove an electron from an atom
- Depends on the element and the shell the electron is in
- Increases:
  - As the atomic number (Z) increases
  - The closer the electron is to the nucleus (i.e. **highest for the K-shell**)
- Expressed in **electron volts** (eV)

Element	Z	K-shell binding energy (keV)
Iodine	53	33
Barium	56	37
Tungsten	74	70
Lead	82	88

Electromagnetic radiation

Electromagnetic radiation (EMR)

- EMR is energy in the form of a self-propagating wave that travels across empty space or through matter
- In a vacuum, this energy travels at the **speed of light (c)** which is ~ **$3 \times 10^8 ms^{-1}$**
- EMR has both **electrical** and **magnetic** components
- Classified according to the frequency of its wave
- X-rays and gamma rays are both types of EMR:
  - X-rays are emitted by electrons outside the nucleus
  - Gamma (γ) rays are emitted by the nucleus

Wave-particle duality

- EMR behaves like a wave and a particle at the same time
- Rather than being composed of particles, EMR is represented as a stream of packets of energy known as **photons** that **travel in straight lines**
- Photons have **no mass**

The wave model

- EMR is a transverse sinusoidal wave
- Frequency is measured in **Hertz** (Hz) where 1 Hz = 1 oscillation per second
- Waves have successive peaks and troughs. The distance between 2 peaks is known the **wavelength** (λ)
- The height of the peak is the **amplitude** (A)
- The time between 2 peaks is the **period** (T)
- As waves cross boundaries between different media, their speed changes but the

frequency stays the same

The particle model

- The **frequency** of a wave of EMR is **proportional to the energy** of its photons
- Photons act as transporters of energy

Two important equations

The **wave equation**:
$v = f \lambda$
(v = velocity, f = frequency, λ = wavelength)

The **Planck-Einstein equation**:
$E = h f$
(E = photon energy in electron volts (eV), h = Planck's constant, f = frequency)

These can be combined to:
$E = hc / \lambda$ or **$E = 1.24 / \lambda$**
(E in keV, λ in nm)

The electromagnetic spectrum

Radiation	Wavelength	Frequency	Energy
Radiowaves	1000 - 0.1 m	0.3 - 3000 MHz	< 10 μeV
Microwaves	100 - 1 mm	3 - 300 GHz	10 - 1000 μeV
Infrared	100 - 1 μm	3 - 300 THz	10 - 1000 meV
Visible light	700 - 400 nm	430 - 750 THz	1.8 - 3 eV
X & gamma rays	1 nm - 0.1 pm	300 PHz - 300 EHz	1 keV - 10 MeV

EMR intensity

- Radiation travels in **straight line rays** that radiate in all directions from a point source

- A collimated set of rays is known as a **beam**
- Take a cross-sectional slice of a beam and count the number of photons. This is the **photon fluence** of the beam at that point
- A beam may contain photons of different energies
- The amount of energy of all the photons in our photon fluence is the **energy fluence** at that point
- The energy fluence per unit time is the energy fluence rate, also known as the **beam intensity**

Inverse square law

- The intensity of the beam is inversely proportional to the square of the distance from the source
- E.g: If you move double the distance from a source, the intensity falls by a factor of four

The production of X-rays

Overview

- Electrons are accelerated through a vacuum in an X-ray tube and strike a metal target (usually tungsten)
- The energy from this collision is lost in 2 ways:
  - Interaction with the **outer shell electrons** of an atom generating **heat**
  - Interaction with the **inner shell electrons** or the **nuclei** themselves generating **X-rays**

Characteristic radiation

- If an incoming electron hits a K-shell electron with an energy level greater than the binding energy of the K-shell (E_K), then the K-shell electron will be ejected from the atom
- The hole in the K-shell needs to be filled by an electron dropping down from an outer shell:
  - When this happens, a photon is emitted
  - The **photon energy is equal to the difference between the binding energies of the 2 shells**
- The most likely situation is that an L-shell electron will drop down to fill the hole:
  - In this case, the emitted photon is termed **K_α radiation** (energy = E_K - E_L)
  - A less likely situation is that an M-shell electron drops down:
    - This is termed **K_β radiation** (energy = E_K - E_M)
- L-radiation (when an electron is knocked out of the L-shell) also occurs but is of such little energy that it plays no significant part in radiology
- Our X-ray photons therefore have a few discrete energy levels and constitute a **spectrum** that is termed the **characteristic radiation**

- Characteristic radiation is **determined by atomic number** and **unaffected by tube voltage**
- A K-shell electron cannot be ejected from the atom if the kV is less than E_K

The characteristic radiation of tungsten

$Z = 74$, $E_K = 70keV$, $E_L = 12keV$
Kα radiation = $E_K - E_L = 70 - 12 =$ **58keV**

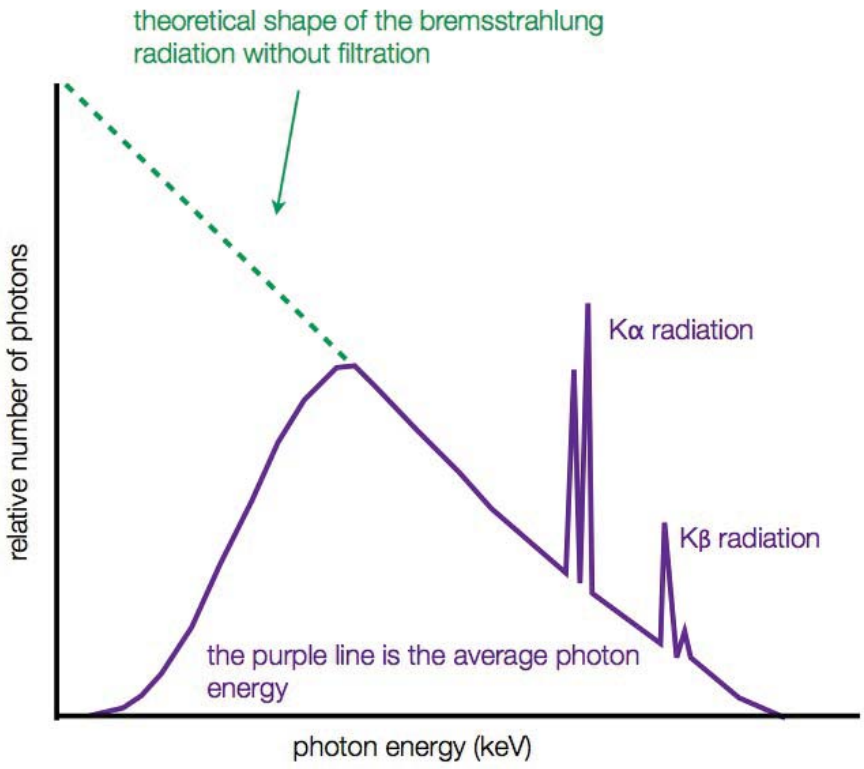
Bremsstrahlung radiation ("braking radiation")

- If an incoming electron penetrates the K-shell and approaches the nucleus, it is deflected
- During the deflection, the **electron slows down and emits an X-ray photon**
- Except in mammography, **80% of X-rays emitted from an X-ray tube are bremsstrahlung**
- The maximum amount of energy that can be emitted equals the kV. This is rare:
  - It occurs when an electron is completely stopped by this braking force
- Most electrons will first lose some energy as heat before interacting with the nucleus
- Bremsstrahlung radiation is a **continuous spectrum**
- The maximum photon energy (in keV) is numerically equal to the kV

The X-ray spectrum

- Minimum and maximum energy levels:
  - The dashed line in the figure represents the total amount of bremsstrahlung produced
  - A substantial amount of the lower energy photons are absorbed by the target, the tube and other materials and produce a **low-energy cut-off** at about **20 keV**

    - The **high level cut-off depends only on the kV**
- The average or **effective photon energy** of the spectrum is **50 - 60% of the maximum**
- As the kV is greater than the K-shell binding energy, characteristic radiation is also produced
- The **area under the curve** represents the beam **intensity** (or total number of photons)
- The **efficiency** of X-ray production **increases with the kV**



Controlling the X-ray spectrum

- **Increasing kV** (tube voltage):
    - Shifts the spectrum up and to the right
    - **Increases** the effective **photon energy** and **increases** the total **number of photons**
- **Increasing mA** (tube current):

    - Does not change the shape of the spectrum
    - **Increases** the **output** of both bremsstrahlung and characteristic radiation
- Decreasing the target atomic number:
    - Decreases the amount of bremsstrahlung radiation
    - Decreases the photon energy of the characteristic radiation
- A **constant kV** potential produces **more X-rays** and at **higher energies**
- Filtration (see Filtration of X-ray beams)

Interaction of high energy photons with matter

Overview

- Three things can happen to a photon as it travels through matter:
  1. **Transmission**
  2. **Absorption**
  3. **Scatter**
- The X-ray image is formed by the transmitted photons. Those that are absorbed or scattered are said to have been **attenuated**

Attenuation

- Depends on **photon energy** and the material's **atomic number**
- All of our calculations make the (incorrect) assumption that X-ray beams are **monoenergetic**. We know, of course, that they are a spectrum

Half-value layer (HVL)

- Defined as the **thickness of a material that reduces the intensity of an X-ray beam to 50% of its original value**
- Is a measure of the penetrating power of the beam

Linear attenuation coefficient (μ)

$$\mu = 0.693 / HVL \text{ (unit is } m^{-1})$$

- Is the fractional reduction in intensity of a parallel beam of radiation **per unit thickness**
- A parameter that quantifies the attenuating properties of a material

- **Inversely proportional to HVL**
- **Depends on the density** of the material
- It's the probability that a photon interacts per unit length of the material it travels through
- **Increases as**:
  - **Density increases**
  - **Atomic number increases**
  - **Photon energy decreases**

Mass attenuation coefficient (mac)

$$\textbf{mac} = \mu \textbf{ / density} \text{ (unit is } cm^2g^{-1})$$

- The fractional reduction in intensity of a parallel beam of radiation **per unit mass**
- **Depends only on photon energy and atomic number**

Attenuation of a heterogenous beam

- X-ray beams are a spectrum and are therefore heterogeneous
- More photons are attenuated as the **effective energy of a heterogeneous beam is less** than a monoenergetic beam
- As the X-ray beam penetrates a material it becomes progressively more homogeneous:
  - This is because the lower energy photons are attenuated proportionally more than the higher energy ones
  - Known as **beam hardening**

Photon interactions

- There are four ways that photons interact with matter to cause attenuation:

1. **Photoelectric absorption**
2. **Compton scatter**
3. Elastic (Rayleigh) scattering
4. Pair production

- Photoelectric absorption and Compton scattering are the most important types of interaction

Photoelectric absorption

1. An incoming photon collides with an electron and has sufficient energy to overcome the binding energy
2. The photon is completely absorbed
3. The electron is ejected from the atom with a kinetic energy equal to the difference between the binding energy and the initial photon energy
4. The ejected electron is known as a **photoelectron**
5. The "hole" in the electron shell is filled by an outer shell electron "dropping down" with the emission of another photon of energy
6. **The emitted photons and the photoelectron are completely absorbed close to the atom**

Atomic number ↑, photoelectric ↑
Photon energy ↑, photoelectric ↓

photoelectric absorption α Z^3 / E^3
(Z = atomic number, E = photon energy)

Compton scatter

- The higher the energy of an incoming photon, the less electrons appear bound
- Electrons effectively become "free" electrons
  1. An incoming high energy photon collides with a free electron

2. The **electron recoils** and takes away some of the photon's energy
3. The **photon is scattered** in a new direction with less energy

- In diagnostic radiology, **only 20% of the photon energy is absorbed, the rest is scattered**

The scatter angle

- The scatter angle (θ) is the angle between the scattered photon and the incoming photon
- **Photons** can be scattered **in any direction** but **electrons** can only move **forwards**
- The change in photon energy is determined only by the scatter angle
- **Direct hit**:
  - The **electron** will travel **forwards** and receives **maximum** energy
  - The scattered photon travels backwards (θ = 180°) and receives minimum energy
- **Glancing hit**:
  - The **electron** travels at 90° and receives **minimum** energy
  - The scattered **photon** goes almost straight **forwards** (θ = 0°) and receives **maximum** energy
- Thus, as **scatter angle** ↑, **scattered photon energy** ↓

Effect of photon energy and Compton scatter

- As the incoming photon energy increases, more photons are scattered forwards
- Incoming photon energy ↑, scattered photon energy ↑
- Incoming photon energy ↑, electron energy ↑, electron range ↑

Compton vs photoelectric

- **Photoelectric absorption**:
  - Increases with Z^3
  - Decreases with photon energy
- **Compton scatter**:
  - Increases with density
  - Independent of Z
  - Decreases (slightly) with photon energy
- **Compton scatter** is important with **low Z** materials at **high photon** energies
- **Photoelectric absorption** is important with **high Z** materials at **low photon** energies

Elastic (Rayleigh) scattering

- Occurs when the incoming photon has an energy of less than the binding energy
- Incoming photon hits a firmly bound electron and is deflected with **no loss of energy**
- Most likely to occur at **high Z** and **low photon** energies
- Not important in the realms of diagnostic radiology

Pair production

- Only happens at very high energies (**at least 1.02 MeV**, which is 2 * 511 keV)
  1. Incoming photon passes close to the nucleus
  2. A **positron and an electron are formed** from the photon's energy ($E = mc^2$)
  3. If the incoming photon had > 1.02 MeV of energy then what's left over is given to the electron and positron as kinetic energy
  4. The electron causes excitations and ionisations and gradually loses energy
  5. The positron combines with a "free" electron and is **annihilated**:
     - Produces **two 511 keV photons** that are emitted at **180°** to each other

- Net effect of pair production is that **all** of the initial incoming photon's energy is transferred to the material

Absorption edges

- Generally, absorption decreases as photon energy increases
- Orbiting electrons can only absorb energy from a photon when the photon has an energy greater than its binding energy
- The probability of an interaction between a photon and an electron increases dramatically when the photon has just a little bit more energy than the binding energy
- This probability falls again as the photon energy increases
- Maximum absorption occurs when the photon has **just enough energy** to eject a bound electron
- This **sudden jump in the attenuation coefficient** is known as the **K-edge** (a less important L-edge also exists)
- As **Z** ↑, K-shell binding energy ↑ and **K-edge** ↑
- K-edges are important for **contrast agents** (Iodine 33 keV, Barium 37 keV, Lead 88 keV)

Filtration of X-ray beams

Overview

- Low energy photons are largely absorbed by the patient and contribute to dose without contributing to the image
- The purpose of filtration is to reduce the dose to the patient without compromising the image
- Two types of filtration:
  - **Inherent filtration**:
    - Anode, tube housing, insulating oil and the glass insert
    - Typically equivalent to ~ **1 mm Al** (aluminium)
  - **Added filtration**:
    - Uniform flat sheet(s) of metal, usually aluminium or copper
- **Total filtration = inherent filtration + added filtration**:
  - Should be **at least 2.5 mm Al equivalent**
  - Therefore, we usually add about 1.5 mm Al

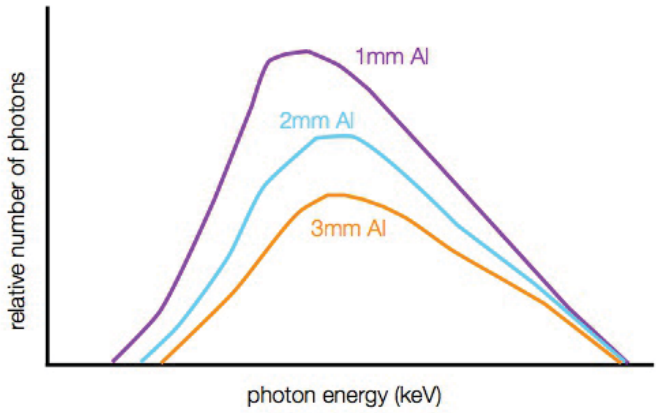
Filter materials

- We want the filter material to cause predominantly **photoelectric absorption**:
  - This is the only type of attenuation that is energy-dependent (remember Z^3 / E^3)
  - Z must not be too high as we don't want to soften the beam
- **Aluminium** (Z = 13):
  - Usually 1.5 mm
- **Copper** (Z = 29):
  - Usually **0.1 - 0.3 mm**

- Produces 9 keV characteristic radiation which must be absorbed by an Al backing filter

Effects of filtration

- Filters attenuate low energy photons proportionally more than high energy ones
- **Hardens the beam** (increases penetrating power and HVL) but **lowers intensity**
- **Reduces skin dose** without affecting image quality
- Increasing filtration:
  - Shifts the X-ray spectrum to the right
  - Increases minimum photon energy
  - Increases effective photon energy
  - Reduces the total number of photons
  - Does not affect the maximum photon energy



The graph illustrates the spectrum being shifted to the right and reduced in intensity with the addition of filtration

K-edge filters

- Rarely used except in **mammography**
- Are materials with K-edges in the higher part of the X-ray spectrum
- Remove both high & low energy photons but are transparent to photons just

below the K-edge

- An example is erbium (Z = 68)

Luminescence

Definitions

- A process where energy from an external source is absorbed and re-emitted as visible light
- There are two types:
  - **Fluorescence**:
    - The (near) **instantaneous** emission of light following energy input
  - **Phosphorescence**:
    - **Delayed** emission of light (known as afterglow)
- A luminescent material is known as a **phosphor**
- Crystalline materials used to detect gamma radiations are known as **scintillants**
- Some phosphors only emit light after further input of energy:
  - **Thermoluminescence** (after heat input)
  - **Photostimulable** luminescence (after light input)

The band structure of a phosphor

- We can think of crystalline solids as having three energy levels:
  - **Valence band**:
    - The lowest energy level
    - Occupied by electrons and completely filled
  - **Conduction band**:
    - A higher energy level than the valence band
    - Is vacant
  - **Forbidden zone**:
    - Between the valance and the conduction bands

        - Describes the energy levels that cannot be occupied by an electron in that material

- Within the phosphor, there may be **impurities** that have energy levels different from those of the phosphor itself:
    - Introduces discrete energy shells within the forbidden shell that are unoccupied
    - Known as **electron traps**

What happens when a photon hits a luminescent material?

1. Photon interacts with an atom or electron:
    - Causes an **electron from the valence band to "jump" to the conduction band**
    - Leaves a vacancy in the valence band
2. Electrons in the conduction band are able to move freely within the material
3. Electron may either **fall back down** to the valence band and emit light (fluoresce) or get "stuck" in an electron trap (phosphoresce)
4. An electron is unable to leave an electron trap unless it gains further energy in the form of heat (thermoluminescence) or light (photostimulable luminescence)
5. The intensity of emitted light is proportional to the incident X-ray beam intensity

Thermoluminescent phosphors

- An example in radiology is **lithium fluoride**
- Used in TLDs (thermoluminescent dosimeters)
- After irradiation, there are electrons in the electron traps
- At room temperature, these electrons have insufficient energy to jump back to the conduction band (and thereafter fall back down to the valence band)
- When heated to a high temperature, the electrons escape and luminescence occurs

Photostimulable phosphors

- An example in radiology is europium-activated barium fluorohalide:
  - Stimulated at 500–700 nm (red laser light)
  - Emits light at 400 nm (blue-green light)

Chapter 2

Ionising radiation dose

Absorbed dose and kinetic energy released to matter

Absorbed dose

- The **energy deposited per unit mass**
- Unit is the **Gray** (J / Kg)
- Defines the quantity of radiation delivered at a specific point in the radiation field

Kerma

- Kerma is an acronym: **K**inetic **E**nergy **R**eleased to **MA**tter
- It is the **energy transferred per unit mass** at a specified position
- Air kerma is the energy (in J) transferred to a unit mass (in Kg) of air
- The unit of kerma is the **Gray** (Gy)
- Beam intensity is hard to measure so it is indirectly measured using the **air kerma rate**
- At diagnostic energies, kerma and absorbed dose are effectively equal

Equivalent dose and effective dose

Linear energy transfer (LET)

- LET is the **energy transferred to a tissue per unit length**
- Depends on radiation type and energy
- **High LET = more damaging**
- Low LET radiation:
  - X-rays
  - Gamma rays
  - Electrons
- High LET radiation:
  - Alpha particles:
    - Less energy than X-rays and travel less distance but cause more ionisations in a smaller space which are more likely to be irreparable
  - Neutrons

Equivalent dose

equivalent dose = absorbed dose x ω_R
(where ω_R = radiation weighting factor)

- ω_R is 1 for X-rays, gamma rays and β-particles
- ω_R is 20 for α-particles (as they are high LET particles)
- Unit is the **sievert (Sv)** which is measured in J / Kg
- In diagnostic radiology, equivalent dose = absorbed dose

Effective dose

- Takes into account the variation in the radiosensitivity of different tissues
- Unit is the **sievert**

$$\textbf{effective dose} = \textbf{equivalent dose} \times \omega_T$$

(where ω_T = tissue weighting factor)

- **Directly related to risk** to the person irradiated
- Based on best scientific evidence of the effects of radiation

Effective dose examples

The table below gives a few examples of the risk of developing a fatal cancer per Sv:

Organ / tissue	ω_T	Risk factor (% per Sv)
Gonads, stomach, colon, lung, bone marrow	0.12	0.5 - 1.1
Breast, oesophagus, liver, thyroid	0.05	0.1 - 0.3
Bone, skin, brain	0.01	0.02 - 0.05

Effects of ionising radiation on living tissue

Overview

- **Somatic** effects influence the irradiated individual
- Genetic or hereditary effects influence the offspring of irradiated individuals
- These effects can be either **deterministic** or **stochastic**

Deterministic effects

- These have a **threshold dose** below which the effect will not occur
- The threshold dose is fairly constant between individuals
- Once the threshold dose is exceeded, the likelihood of the effect occurring increases rapidly, up to a level at which the effect will definitely occur
- Most deterministic effects have **repair mechanisms** such that the rate at which the dose is delivered influences the threshold dose:
  - **Cataracts** in the eye are an exception

Stochastic effects

- Arise by **chance**
- No threshold
- **Risk increases linearly with dose**
- **Severity of the effect does not increase with dose** (the patient either gets cancer or does not)

Biological damage

- Caused by **ionisation**

- The body is 70% water and so most damage relates to the ionisation of H2O:
  - Leads to the formation of **free radicals** and **hydrogen peroxide**
  - These damage other molecules such as DNA
- The energy released during ionisation is also capable of breaking molecular bonds without the production of free radicals
- Biological damage could cause:
  - Premature cell death (perhaps via cell membrane damage)
  - Prevention of cell division
  - **Neoplastic** transformation
  - Genetic **mutation** that can be passed on to daughter cells
- Damage may occur immediately or after a number of hours

Factors influencing radiation-induced damage

- **Dose rate**:
  - Slower rate = less biological effect
  - Allows time for recovery in the irradiated tissue
  - This is the principle underlying radiotherapy **fractionation**
- **Type of radiation**:
  - High vs low LET radiation
- Tissue **radiosensitivity**:
  - Rapidly dividing & poorly differentiated cells are more sensitive to radiation-induced damage
- **Age at exposure**:
  - The younger the irradiated person, the greater the risk

Radiation risk

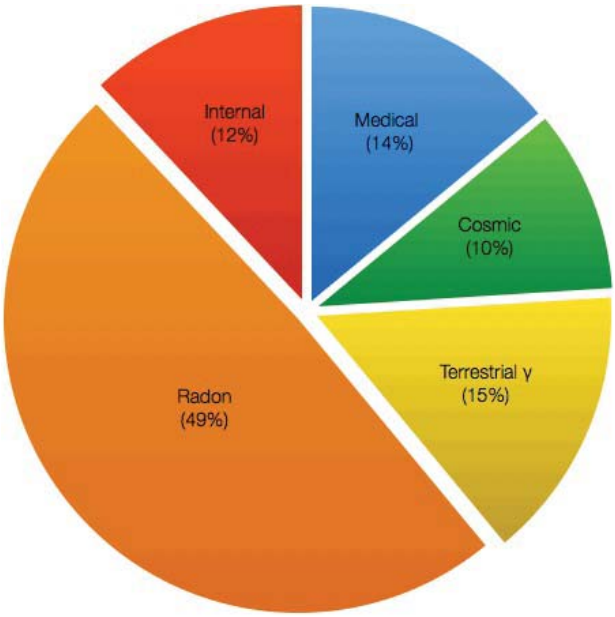
Overall risk per mSv

	Cancer	Hereditary
Whole population	1 in 20,000 per mSv	1 in 500,000 per mSv
Adult population	1 in 24,000 per mSv	1 in 100,000 per mSv

Threshold doses for deterministic effects

Effect	Threshold dose (Gy)
Foetal abnormality	0.1 - 0.5
Sterility	2 - 3
Skin erythema, hair loss	2 - 5
Bone marrow failure	3 - 5
Cataracts	5
Fatal CNS damage	15

Population doses from natural and artificial sources



Example doses in radiology

	Dose (mSv)	Risk of fatal cancer
Background radiation	2.2	
CXR	0.01	1 in 1,000,000
Mammogram	0.1	1 in 200,000
AXR	0.5	1 in 40,000
CT brain	2	1 in 10,000
Bone scan (^{99m}Tc)	4	1 in 5,000
Ba enema	5	1 in 4,000
CT abdomen + pelvis	10	1 in 2,000

Foetal dose

- **Deterministic** effects have a threshold of **100 mGy**:
  - Death, low IQ, malformations
  - Risk is greater in early pregnancy
- **Cancer risk** is about **1 in 13,000 per mGy**
- Termination of pregnancy:
  - Dose < 100 mGy:
    - Cannot justify termination
    - > 99% chance that child will not develop a childhood cancer
  - Dose 100 - 500 mGy:
    - Decision to terminate is based on individual circumstances
  - Dose > 500 mGy:
    - There may be significant foetal damage

Foetal doses from maternal diagnostic radiology procedures

	Foetal dose (mGy)	Childhood cancer risk per exam
CXR, lung ventilation scan (Kr)	< 0.01	1 in 1,000,000
CTPA	0.01 - 0.1	1 in 100,000
AXR, renal DMSA scan	0.1 - 1	1 in 10,000
Barium enema, CT abdomen, IVU	1 - 10	1 in 1,000 to 1 in 10,000
CT pelvis, whole body PET	10 - 50	> 1 in 1,000
Fluoroscopy	10 - 100 (or more)	

Chapter 3

Radiography with X-rays

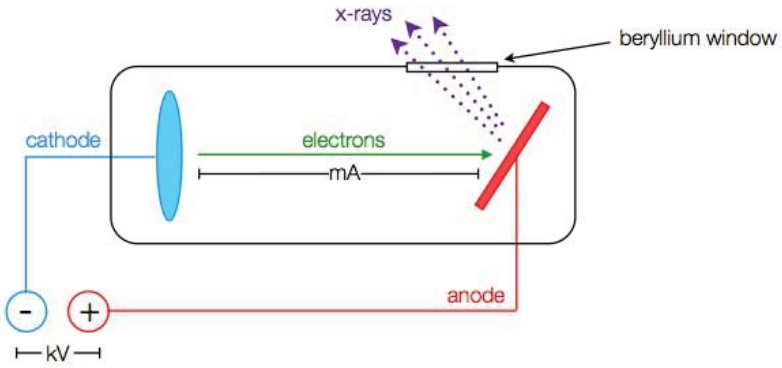
The X-ray tube

Overview

- X-rays are produced when **fast moving electrons are suddenly stopped by hitting a metal target**
- The kinetic energy of the electrons is converted into:
  1. X-rays (1%)
  2. Heat (99%)
- The acceleration and impact needs to occur within a **vacuum** (or else air atoms would interfere)

Inside the tube

- An X-ray tube consists of 2 electrodes sealed within an evacuated glass envelope
- The **cathode** (**negative** electrode) is a fine tungsten **filament** and **focussing cup**
- The **anode** (**positive** electrode) is a smooth metal **target** (usually tungsten)



The focussing cup

- Directs the electrons to the anode and improves resolution
- High melting point

- Poor thermionic emitter

Producing the electrons

- Heat the tungsten filament by passing an electric current through it
- Electrons "boil" off the filament and produce a "cloud" of electrons adjacent to it:
  - A process known as **thermionic emission**

Driving the electrons

- Two sources of electrical energy are required by the tube:
  - Filament heating voltage (~10 V) and current (~10 A)
  - Accelerating voltage (typically 30 - 150 kV) also known as the **tube voltage** or **kV**
- The flow of electrons between the cathode & anode generates a **tube current** or **mA** (typically 0.5 - 1000 mA)

kV and mA

- The kV and the mA can be varied independently
- The **mA is controlled by adjusting the filament temperature** (by changing the filament voltage and current)
- A small increase in filament temperature generates a large increase in mA

Properties of a good anode (target)

- **High atomic number** (increased conversion efficiency into X-rays)
- **Good conductor** (dissipates heat from the focal spot to the rest of the anode)
- **High melting point** (tungsten = 3370 °C)

- **Low vapour point** (prevents the anode "boiling off")

Rotating anode

- Most X-ray tubes utilise a rotating anode for the following reasons:
  1. Increases the area struck by electrons without increasing the focal spot size
  2. Can withstand greater heating
- Spins at 3000 - 10,000 rpm

The focal spot

- The focal spot is the area where electrons strike the target
- A small focal spot gives a better resolution
- X-rays are emitted from all directions from the focal spot:
  - Useful X-rays pass through a glass (or beryllium in mammography due to it's low Z) window
  - The remainder are stopped by lead shielding

Angled focal spot

- The anode is angled for several reasons:
  - Anode **heating is spread over a larger area** for the same effective focal spot size
  - Having a smaller focal spot **reduces penumbra** (edge blurring)
- The **anode heel effect**:
  - A disadvantage of angling the anode
  - Leads to a **reduction in beam intensity towards the side of the anode**
  - Caused by **absorption of the emitted X-rays by the anode itself**:
    - Occurs when the X-rays emerge at a near-grazing angle

- Can be reduced by increasing the focus-image distance (FID)

What happens at the target?

- Each electron arrives at the surface of the target with a kinetic energy (in **keV**) equal to the kV
- Two processes cause an electron to lose its energy:
  1. Small energy interactions with the **outer electrons** of the atoms generating unwanted **heat**
  2. Large energy losses by interactions with the **inner shells** of the atoms or the **nucleus** producing **X-rays**. This is more likely the higher the initial electron energy
- This is discussed further in the production of X-rays

Contrast resolution

Overview

- A structure in a patient is demonstrated by two things:
  1. The sharpness of the image at its boundary
  2. The contrast between it and adjacent tissues caused by differing X-ray transmission
- Subject contrast (C) is given by the equation:

$$\mathbf{C \propto (\mu_A - \mu_B) * t}$$

(where μ_A & μ_B = linear attenuation coefficients of tissue A and B, t = tissue thickness)

- **Contrast increases** as structure **thickness increases**
- **Contrast deceases** as μ decreases (e.g. if **kV increases**)

Contrast media

- There are two main ways to improve soft tissue contrast:
  1. Lower kV (but this increases dose)
  2. Use a contrast medium
- Radiopaque contrast media have a sufficiently high atomic number to maximise photoelectric absorption
- Their K-edge lies just to the left of the spectrum of X-rays leaving the patient:
  - Iodine (Z = 53, E_K = 33 keV)
  - Barium (Z = 56, E_K = 37 keV)

Spatial resolution & noise

Spatial resolution

- Reflects the size of the smallest visible detail
- Quantified as the highest occurring frequency of lines that can be resolved in a bar pattern (**line pairs per mm**)
- The smallest visible detail size is inversely related to the number of line pairs:

$$\text{smallest detail (mm)} = 1 / \text{number of line pairs per mm} * 0.5$$

Examples of spatial resolution values

Modality	Spatial resolution
Film	8-12 lp/mm
General DR	3-5 lp/mm
CT	1-2 lp/mm
Nuclear medicine	0.5 lp/mm

Noise

- **Reduces contrast resolution**
- Refers to the **variations in the levels of grey** in the image that are distributed over its area but are unrelated to the structures being imaged
- Can be random (due to **photon number**) or structured

Random noise (quantum mottle)

- This is the most significant source of noise in radiological imaging
- Increases as the number of photons detected decreases

- Is seen as a graininess in the image
- If N photons are detected then the signal-to-noise ratio (SNR) is defined by the equation:

$$\mathbf{SNR = N / \sqrt{N}}$$

- i.e. **signal-to-noise ratio increases as the number of photons contributing to the image increases**

No. of photons	Noise (%)	SNR
10	3 (30)	3:1
100	10 (10)	10:1
1000	32 (3.2)	32:1
10,000	100 (0.3)	320:1

Structured noise

- Caused by overlying or underlying anatomy
- Electronic noise in the system is caused by instability in the electronic circuitry of the set

Contrast to noise ratio (CNR)

- Widely used in digital systems
- Takes the image receptor into account
- Consider the CNR between two tissues, A and B:

$$\mathbf{CNR = (PV_A - PV_B) / noise}$$

(where PV = pixel value and is proportional to X-ray intensity)

Scatter rejection

Overview

- Primary radiation carries, whilst scattered radiation obscures, useful information
- The amount of scattered radiation (S) is several times the amount of primary radiation (P) reaching the detector
- The ratio of scattered to primary radiation depends on:
  - The area of the beam (**field size**)
  - **Patient thickness**
- Typical S:P ratios are 4:1 for a CXR and 9:1 for a lateral pelvis

Effect of scatter on contrast

- Scatter radiation is, more or less, uniform over the image
- Reduces image contrast by an order of magnitude

$$\mathbf{Cs\ (\%) = 100 / SF}$$

(where Cs = contrast with scatter as a %, SF = scatter factor = scatter : primary ratio)

- We can see from the above equation that if SF = 4 (i.e. 4 x more scattered radiation reaches the detector than primary radiation) then our contrast will be 25% of what it would be if there was no scatter (SF = 1)

How to reduce scatter and improve contrast

1. **Decrease field size** (by using collimation)
2. Decrease patient thickness
3. **Decrease kV**

4. Use an antiscatter **grid**
5. **Increase object-image distance** (OID)

Decreasing kV and scatter

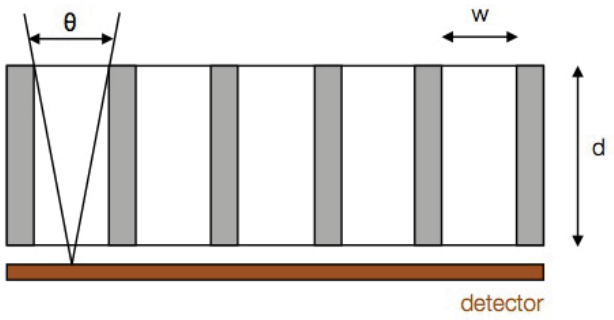
- As kV is decreased, the scatter produced is **less in the forwards direction** so less reaches the detector
- In addition, the **scatter produced is less penetrating**
- Decreasing kV increases the contrast but mainly because of increased photoelectric absorption

Antiscatter grid

- Consists of thin strips of lead sandwiched between thicker strips of low attenuation interspace material
- Scattered photons that hit the grid obliquely are absorbed by the lead
- A high proportion of the primary photons pass through the gaps and reach the detector
- **Improves contrast**
- **Increases patient dose** (as some primary photons are absorbed by the interspace material)
- Not generally required if a long object-image distance is used (the "air gap technique")

Grid construction

- Lead strips are usually **~0.05mm wide**
- The number of strips per cm (the **line density**) typically **20 - 60 strips / cm**
- Interspace material is typically carbon fibre (low attenuation)
- Interspace strips are typically 0.2mm wide



Grid ratio

- This is the ratio between the depth of the interspace channel divided by it's width:

grid ratio = d / w

- A typical ratio is **8:1**
- The **larger the grid ratio**, the smaller the angle of acceptance (θ) and **the better the contrast**
- Antiscatter grids typically improve the contrast by a factor of 2 - 4

Unfocussed grid

- The lead strips are **parallel** with each other and the centre of the X-ray beam
- Away from the centre of the beam, the X-rays strike the grid obliquely due to ray divergence
- The rays are, therefore, **increasingly attenuated until θ/2** when they are completely cut-off
- This effect is reduced by **increasing FID** or **decreasing the grid ratio**
- Are restrictive in terms of the maximum beam size that can be used

Focussed grids

- More commonly used
- The lead strips are **tilted progressively** from the centre to the edges of the grid so that they all point towards the tube focus
- Must be used at a **specified distance from the anode**
- Tube must be **accurately centred** over the grid
- Grid must be tilted at an angle parallel to the lead strips
- If any of the three conditions above are not met then the primary radiation may be cut off leaving very little of the detector exposed

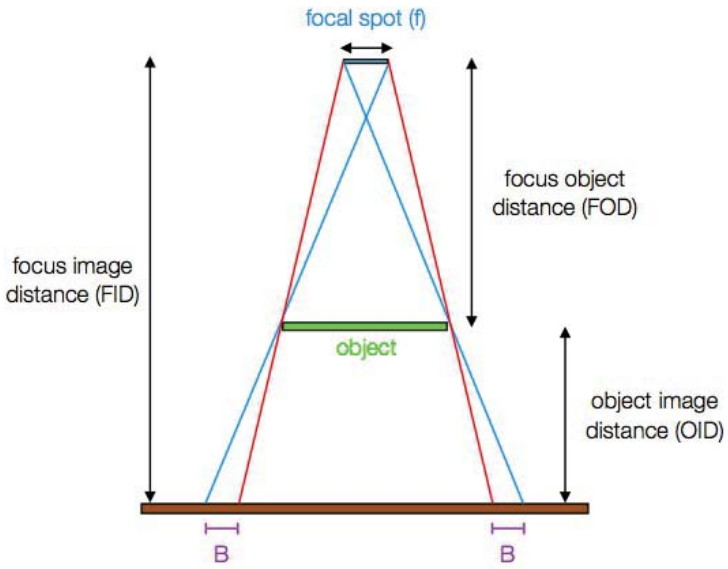
Stationary and moving grids

- **Grid lines** are shadows of the lead strips of a stationary grid superimposed on the image
- If the line density if high enough they may not be visible but, regardless, they do reduce fine detail
- We **blur** the grid lines by moving the grid at a sufficiently fast enough speed
- Moving grids are not practical with **portable devices** (e.g. ward films) and so a stationary grid must be used:
    - Such grids should have a higher **line density** (they will, as a consequence, weigh more)

Planar radiography geometry

Magnification

- Magnification increases as:
  - **OID increases** (air gap)
  - **FID decreases**
  - **FOD decreases**



Distortion

- This refers to a difference between the shape of an object in the image and the patient
- Causes:
  - Tilted objects (e.g. a tilted circle projects as an ellipse)
  - Differential magnification of parts of an object nearer to and further from the image detector
- Can be **reduced by increasing FID**

Geometric unsharpness

- In an ideal world, the image of a stationary object from an infinitely small point source will be perfectly sharp
- Trouble is, we don't have an infinitely small point source of X-rays. We have a focal spot of size f
- Because of this, the **intensity of the shadow changes gradually over distance B** (the penumbra). This is the degree of blurring or unsharpness
- **Geometric unsharpness deceases as**:
  - **Focal spot size decreases**
  - **OID decreases**
  - **FID increases**

Pixels

Overview

- In digital imaging, the image is divided into a matrix of individual **pixels**
- Each pixel is assigned a value that corresponds to the intensity of signal at that point
- High pixel value = high dose = black
- Low pixel value = low dose = white
- The matrix size chosen varies by modality. A common size in CT is 512 x 512 pixels:
  - This matrix will cover a typical 350 mm region (patient width)
  - This gives a pixel size of 350/512 = 0.7 mm
  - We can see that pixel size contributes to the spatial resolution of a system

Bits & bytes

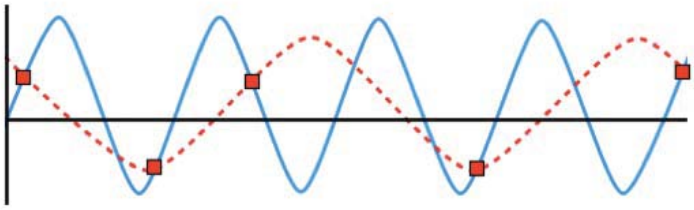
- Pixel values are binary and the maximum value that can be stored is determined by it's bit depth
- The greater the bit depth, the greater the potential to display contrast
- A single bit has 2 possibilities (white or black)
- 12 bits have 212 possibilities (**4096** levels of grey)
- The required bit depth is influenced by the system's noise and dynamic range:
  - Noisy systems such as nuclear medicine may use a bit depth of 8 bits
  - **12 or 14 bits is standard in CR**

Nyquist frequency

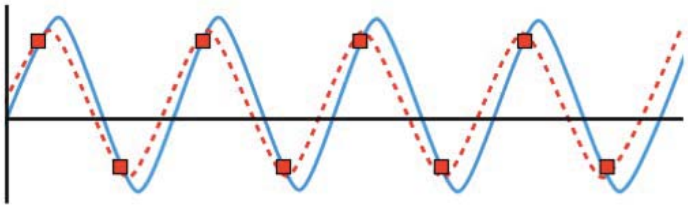
Overview

- **Fourier analysis** is a mathematical method of deconstructing any wave into a series of sine waves that vary in frequency and wavelength
- A digital image is composed of a matrix of samples
- If our sampling frequency is too low then the high frequency components of the wave (which carry information about small structures with sharp edges) will be lost:
  - Results in a low resolution image
- To get an accurate representation of our structure, we must adhere to the **Nyquist criterion**:
  - The signal must be sampled at at least twice the highest frequency present in the signal
  - If we do not adhere to this criterion, high frequency signals will be erroneously recorded as low frequency (a phenomenon known as **aliasing**)
- The maximum signal frequency that can be accurately sampled is called the Nyquist frequency:

nyquist frequency = (sampling frequency) / 2



Under-sampling loses high frequency information



An adequate sampling rate

Computed radiography (CR)

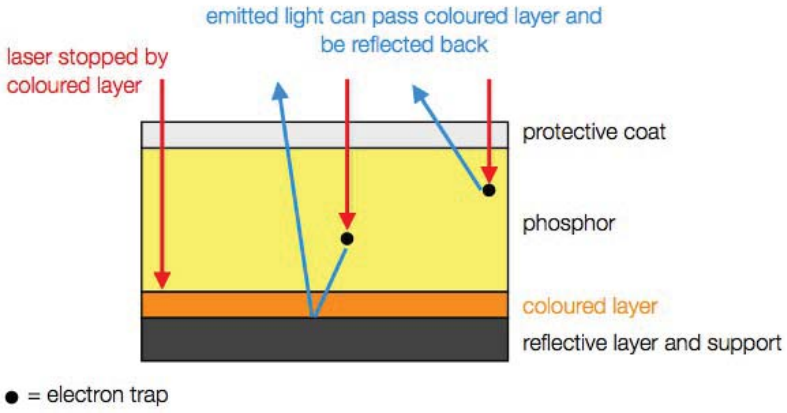
Overview

- The most common way of producing digital radiographic images
- Detector uses a **photostimulable phosphor**:
  - Incident photons are absorbed and electrons excited to conduction band:
    - Approximately 100 electrons become trapped for each photon absorbed
  - Electrons fall back down into electron traps
  - Electrons stay in the trap until **stimulated by red laser light** (500–700 nm):
    - Will fall back eventually over time (leading to decay of the image)
  - Once stimulated, blue-green light (300–400 nm) is emitted and analysed by reader
  - Quantity of emitted light is proportional to absorbed dose
  - Emitted light is detected by a photomultiplier tube

Pathway - from incidence to digitization

1. CR plate exposed to X-rays
2. X-rays absorbed in phosphor
3. Electrons remain in traps until stimulated
4. Insert CR plate into reader
5. Laser light causes luminescence
6. Emitted light collected by photomultiplier tube, amplified and digitized
7. Digital image sent to workstation
8. CR plate cleared

Diagram of a CR image plate



Structure of a CR image plate

- Outer protective coat protects from mechanical wear
- Phosphor layer is usually made from **europium-activated barium fluorohalide**
- **Coloured layer**:
  - **Blocks laser light but permits blue light through**
  - Lowers noise
  - Note that reflected emitted light has reduced spatial resolution
- The reflective layer reflects all emitted light that reaches it back to the reader

Scanning / reading CR image plates

- The film is removed from the cassette within the reader
- The film is moved in one continuous direction and the laser is deflected from side to side to cover the entire plate (the **raster scan** technique)
- An array of optical fibres collect the emitted light and direct it to the photomultiplier tubes
- After reading, the plate must be **erased** by exposing it to a **bright light source** for a short period

Computed radiography image quality

- **Lower spatial resolution than film screen radiography** (3–5 vs 8–12 lp/mm)
- Average pixel size ~100 μm
- Phosphor thickness affects resolution:
  - **Thicker phosphor layer reduces resolution**
  - Due to the laser light used to read the plate being scattered by it
  - Thinner or crystalline phosphors can be used but they are more fragile

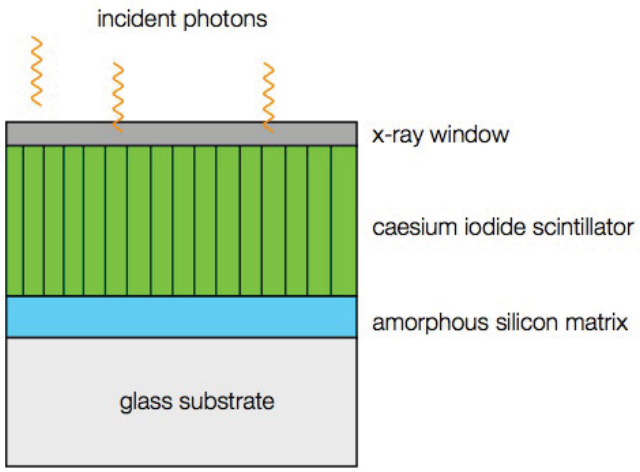
Digital radiography (DR)

Overview

- Based on **thin-film transistor** (TFT) array technology
- A transistor stores electrical charge
- Number of pixels = number of transistors
- Transistor size is **~100–200 μm**
- The active matrix consists of photodiodes incorporated into an **amorphous silicon** TFT array
- Essentially, the image plate consists of a detector overlying the active matrix
- Two types of detector have lead to two types of DR:
  - Indirect
  - Direct

Indirect DR

- Currently more common than direct DR
- X-rays are converted to light by an **overlying phosphor layer**:
  - Usually **crystalline caesium iodide**
- Emitted light is internally reflected downwards within the crystalline structure towards the photodiode where it is converted into an electrical charge
- Electrical charge is stored in the flat panel array
- The stored charge represents a latent image that is later read out



Caesium iodide scintillator layer

- **X-ray absorption efficiency ~ 85%**
- Optically clear to visible light
- Typically **500µm thickness** is used
- **Needle-like crystalline** structure reduces unsharpness due to light scatter

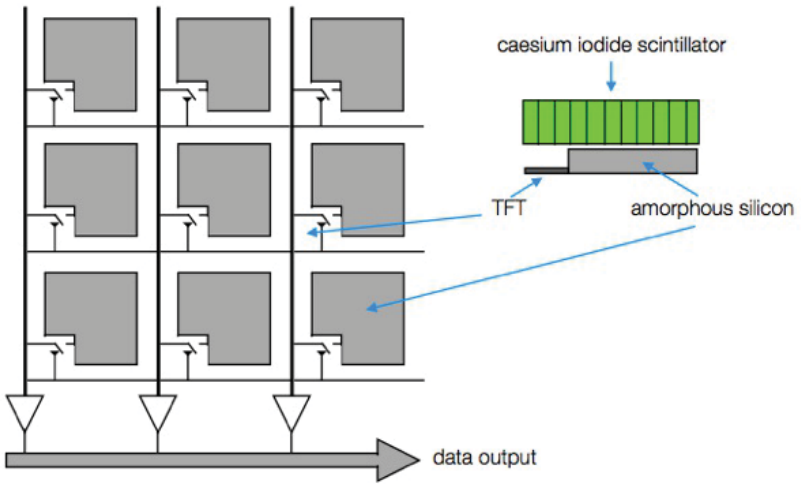
Amorphous silicon

- Light from the scintillator is converted to electric charge
- **Close to 100% efficiency**
- The electric charge is stored in a 2D array of imaging pixels
- Approximately 1500 charge carriers produced for every photon absorbed
- Amorphous silicon is **structurally disordered** and is, therefore, **insensitive to radiation damage**
- Cheap to make in large quantities

The active matrix

- The active matrix consists of an array of pixel regions

- Each pixel region consists of:
  - Photodiode
  - Capacitor (to store charge)
  - A switch (a thin-film transistor or TFT)



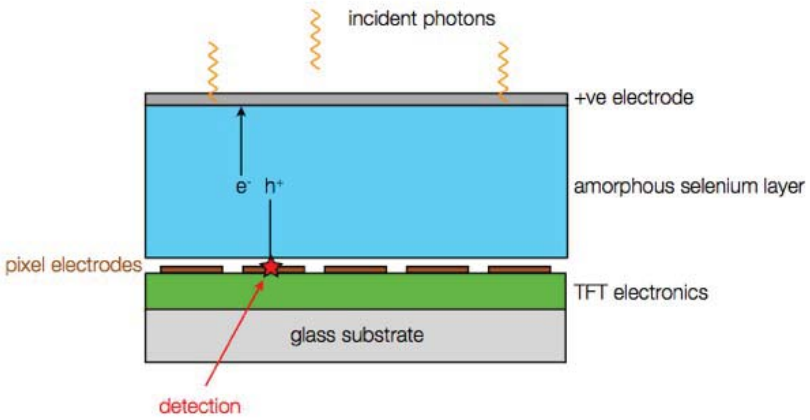
Direct DR

- In this system, **X-rays are converted directly into electrical charge** via a photoconductor layer made of **amorphous selenium**
- **Improved resolution** as there is no visible light stage (we "cut out the middle man")
- Electric charge is again detected and stored in a 2D pixel array forming the latent image
- Removes the need for a phosphor layer
- Expensive and not widely used

Amorphous selenium layer

- Acts as a **photoconductor**
- Typically **500µm thick**
- Coated above an active matrix array manufactured within amorphous silicon

- Absorption of an X-ray photon releases **charge carrier pairs** (i.e. -ve and +ve electron holes)
- Apply a +ve voltage to the outside surface of the amorphous selenium:
  - Drives electrons towards the outside surface and the +ve electron holes towards the electronics
  - Approximately 1000 holes per absorbed X-ray photon



Indirect vs direct digital radiography

- **Indirect**:
  - Superior fractional X-ray absorption (caesium iodide is better than amorphous selenium):
    - **Lower dose for same image quality**, or
    - **Less noise for comparable dose**
  - **Direct**:
    - Slightly **better spatial resolution** as no chance for emitted light to scatter before detection (as none is produced)
    - No cassettes to handle and faster readout times
    - **Better pixel fill factor** (the area of the pixel that is sensitive to light):
      - 86% vs 68%
  - Spatial resolution for DR is about 3–5 lp/mm

Mammography

Overview

Mammography needs to demonstrate both high contrast microcalcifications (can be <100 μm) and larger areas with lower contrast than can be imaged in general radiography.

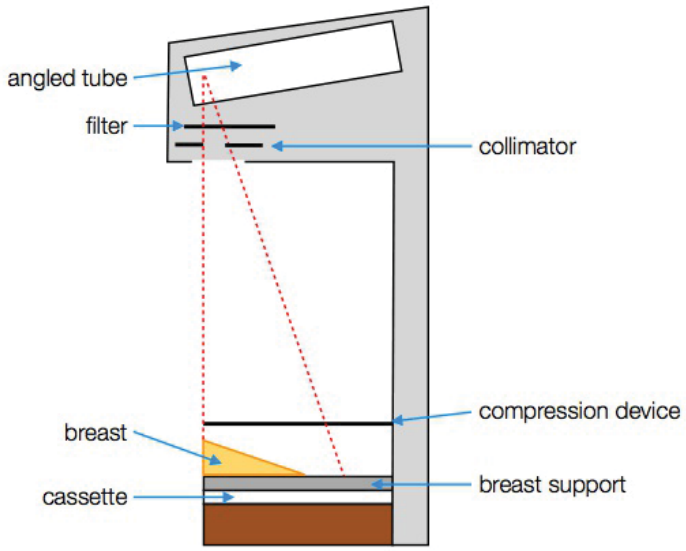
Target and filter materials

- Since the breast is soft tissue, we need to use a **lower kV** than for skeletal radiology
- Mammography sets, therefore, use a lower atomic number (Z) **target** to generate X-rays:
  - Usually **molybdenum–42** (sometimes rhodium–45)
  - These have a narrower, lower energy, characteristic X-ray spectrum than tungsten
- **Molybdenum** (^{42}Mo):
  - K-edge is 20 keV
  - Characteristic radiations at 17 and 19 keV
- **Rhodium** (^{45}Rh):
  - K-edge is 23 keV
  - Characteristic radiations at 20 and 23 keV (thus more penetrating than ^{42}Mo)
- **Filters** are incorporated after the X-ray tube:
  - Usually **molybdenum** again
  - Occasionally use **rhodium filter if the breast is large** (more penetrating)
  - Rule of thumb for progressively increasing breast thickness (target-filter):

Mo-Mo –> Mo-Rh –> Rh-Rh

The X-ray set

- The mammography set has a number of specific features:
  - Angled tube
  - C-arm design
  - Fixed FID (focus-image distance)
  - Compression device
  - Fixed field size



Angled tube

- The radiation across the beam is not uniform due to the **anode heel effect**
- The breast is not uniform in thickness, it is more conical in shape:
  - Thinner at the nipple end and thicker at the base
- To achieve uniform image density, we make use of the anode heel effect:
  - The tube is angled and collimated so that the **highest intensity part of the beam is at the chest wall edge**

Compression

- Reduces the breast thickness by spreading the tissues over a wider area
- Helps to equalise tissue thickness (increases image density homogeneity)
- **Reduces dose**
- **Reduces geometrical unsharpness**
- Immobilises patient:
  - Exposures are long (up to 2 seconds) so helps to **reduce motion artefact**

Patient dose

- A significant issue as most mammograms are done as part of a screening program
- The only part of the patient exposed to radiation is the breast and so we tend to only consider the **average absorbed dose in the tissue** (instead of whole body effective dose):
  - **Typical range is 1.3 - 3 mGy per exposure**
- A dose of 1 mGy would equate to a risk of fatal breast cancer of 1 in 20,000

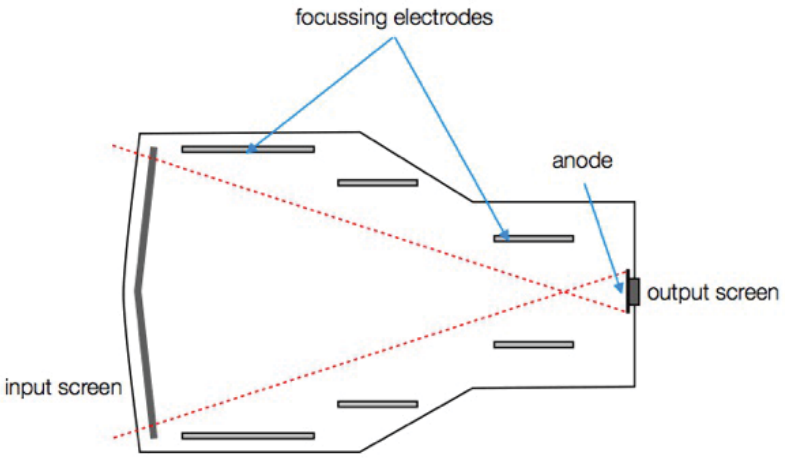
Chapter 4

Fluoroscopy with X-rays

The image intensifier

Construction

- The image intensifier allows **real time imaging** with X-rays
- The components are contained within an evacuated glass envelope that is, itself, enclosed in a metal housing (which prevents light entering the tube)
- Within the tube, there are three main components:
  - Input screen
  - Electron-focussing electrodes (the electron lens)
  - Output screen



The input screen

- Curved
- Varies in size depending on clinical application. Typically 350mm for general work
- **Outer surface** is the **phosphor** layer (caesium iodide)
- **Inner surface** is the **photocathode** layer:
  - Emits electrons when exposed to light

    - Typically made of antimony caesium
    - Approximately 10–20% efficient
- Maintained at a **negative voltage** (potential difference of 25 kV):
    - Any electrons generated are accelerated towards the anode on the output screen

Electron focussing

- Metal rings within the tube held at a positive voltage
- There is a **voltage gradient** to constrain the electrons to travel in a straight line towards the anode
- The pattern of electron intensities falling on the screen is an **exact** (but **minified**) replica of the pattern on the input screen

The output screen

- 25–35 mm in diameter
- **Outer surface is a phosphor layer** (**zinc cadmium sulphide**):
    - Converts **electron** intensities **into light**
- **Inner** (tube) **surface** is a 0.5 μm thick **aluminium** coating:
    - Prevents light generated by phosphor layer travelling back towards the input screen:
        - Would lead to a complete white-out of the image
- **X-ray intensity is directly proportional to output screen brightness**

Gain

- This is the extent to which the tube has intensified the light emitted by the input screen:
    - The **ratio of the brightness of the output phosphor to that of the input**

phosphor

- Two factors are responsible for the gain:
  - **Flux gain**:
    - A single X-ray photon causes a single electron to be emitted from the photocathode
    - Each accelerated electron causes many light photons to be emitted from the output phosphor
    - Typically ~ 50
  - **Minification gain**:
    - This is the intensification caused by reducing the image size from the input to the output screen
    - Is equal to the ratio of the areas of the two screens:
      - E.g. 300mm input screen and 30mm output screen = gain of 300/30 = 10

- **Overall brightness gain = flux gain x minification gain**:
  - 50 x 10 in this example, which equals 500

Conversion factor

- Gain is not a measurable quantity
- We therefore measure the performance of an image intensifier in terms of its conversion factor
- Conversion factor is the **ratio of output brightness** (candela/m^2) **to the dose rate at the input screen** (μGy/s)
- Typical values are **25–30** (Cd/m^2)/(μGy/s)
- Gain (and therefore the conversion factor) **deteriorates with time and usage**:
  - Due to a loss in the detection efficiency of the phosphor

Magnification

- It is possible to alter the voltage gradient of the focussing electrodes so that the electron beams cross over nearer the input screen
- Net effect is to magnify the centre of the input screen
- Magnifies the image and **improves resolution** but **reduces brightness** on the output screen:
  - Due to a reduction in the minification gain
  - To restore brightness we need to **increase the exposure rate** (i.e. the patient skin dose)

The output screen is linked to a charge coupled device (CCD) camera via high quality lenses to allow viewing / digital processing of the image.

The flat panel detector

Image intensifiers are being replaced by flat panel fluoroscopy units. These are essentially the same as those used for **indirect digital radiography**.

Mechanism

1. X-ray photons strike a caesium iodide scintillator
2. Emits visible light
3. Emitted light converted to electrical charge by an amorphous silicon photodiode
4. Charge is stored in a flat panel array and read out
5. Detective quantum efficiency is comparable to an image intensifier (~ 65%)

Advantages of using a flat panel detector vs an image intensifier

- Less bulky
- **Better contrast resolution**
- **Better spatial resolution** (3 lp/mm vs 1 lp/mm)
- **Higher dynamic range**
- Can be used to provide **computed tomography**

Note that magnifying an image with a flat panel detector does not improve spatial resolution (like it does with an image intensifier) because the pixel size is fixed.

Automatic brightness control

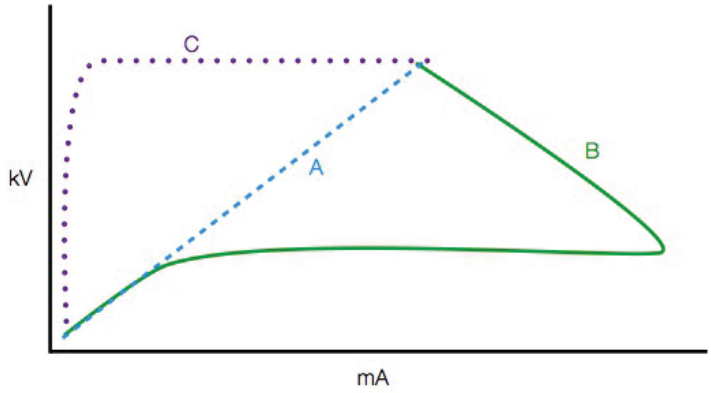
Overview

- Automatic brightness control (ABC) is essential during fluoroscopy as it is not practical to manually change the kV and mAs quick enough when contrast is added
- Measures the brightness in the centre of the output screen and **automatically adjusts kV and mA to provide a consistent brightness**

Dose control curves

- The kV and mAs adjustment follows pre-programmed brightness curves
- **Anti-isowatt curve** (curve A):
  - Increases both kV (to provide better penetration) as well as mA (increases number of photons)
  - This type of curve will increase input power with patient thickness up to a preset maximum
  - Is a **good compromise** between image quality and patient dose
- **High contrast curve** (curve B):
  - Used for iodine contrast studies
  - **Maximizes image quality at the cost of increased dose** (due to low kV)
  - Holds the kV at between 60–65 kV to provide the optimum spectrum for imaging iodine
  - With increasing patient thickness, mA is increased with the kV held at this value
  - Once the tube threshold for mA is reached, the system will increase kV while reducing mA to ensure that the maximum tube power rating is not exceeded
- **Low dose curve** (curve C):
  - **Minimizes dose at the cost of image quality**

- Increases kV rapidly as patient thickness increases up to the maximum kV. As patient thickness continues to increase, the maximum kV is used whilst increasing mA



Digital subtraction angiography (DSA)

DSA is used to produce images of contrast-filled vessels in isolation from other tissues. It allows **less contrast media to be used** and provides improved **clarity**.

Mechanism

1. A non-contrast image is taken before administration:
    - Commonly two are taken (the first to calibrate the exposure factors and the second is stored in memory as the **mask image**)
2. The contrast image is taken when the vessels have filled with contrast medium
3. The mask image is digitally subtracted pixel-by-pixel from the contrast image
4. Recording can continue to provide subtracted images using the initial mask image

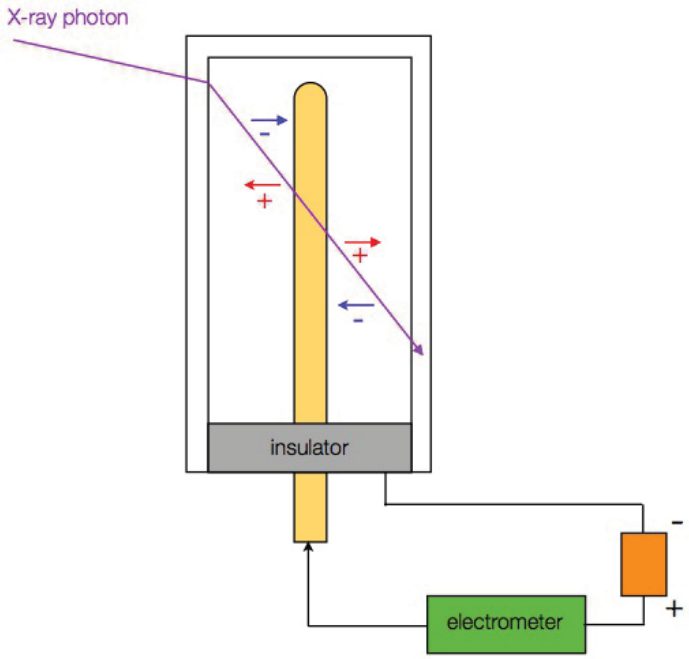
Chapter 5

Safety

Measurement of X-ray and gamma ray dose

Ionisation chambers

- Air kerma is calculated by measuring the amount of ionisation produced by a photon beam in air
- The instrument used is called an ionisation chamber
- The chamber consists of a thin plastic wall (lined with graphite) surrounding an air-filled cavity, separated from a central electrode by an insulator
- Whenever a photon is absorbed by the wall, it liberates an electron which produces ion pairs along its track
- For each coulomb of charge, 34J of energy is deposited
- To measure the charge:
  - Separate the ions by applying a polarising voltage (100 - 300V) between the outer wall and the electrode
  - The **ionisation current** is measured by an electrometer:
    - This current is **proportional to air kerma rate**
    - **Total charge is proportional to air kerma**
- Air is chosen as its effective atomic number ($Z = 7.6$) is close to that of tissue ($Z = 7.4$)


X-ray photon
insulator
electrometer
-
+
-
+
+
-
-
+

Radiation detectors and dose meters

Personal dosimetry systems (PDS)

- Dose limits are specified in terms of effective dose but this is difficult to measure directly
- Personal dosimeters measure the **personal dose equivalent** which is the equivalent dose at a depth d mm in a standard phantom
- There are three types of PDS in general use:
  1. **Thermoluminescent dosimeters** (**TLDs**):
     - Very common
     - Contain **lithium fluoride**
     - Z = 8.2 (close to tissue)
     - Contain filtration to allow measurements of shallow and deep dose
     - To read, it's heated to 250°C and the light output is proportional to dose
     - Accurate to within ± 5%
  2. **Film dosimeters**:
     - Increasing dose blackens the film
     - Provides a permanent record of exposure
     - Several disadvantages:
       1. The sensitivity is highly energy-dependent (silver and bromine are high Z elements)
       2. **Sensitivity is no better better than 0.1–0.2 mSv**
       3. Affected by the environment (i.e. heat)
  3. Optical stimulated luminescent dosimeters:
     - Not common
     - Contain aluminium oxide that has phosphor-like properties
     - Will emit light in proportion to dose when stimulated by a laser

    - Accurate to 0.01 mSv

Dose assessment

- The standard quantities used for dose assessment are:
    - **Entrance surface dose**
    - **Dose area product**

Entrance surface dose (ESD)

- Measured in Gray (Gy)
- Can be **measured directly using TLDs**
- Can also be calculated:
- We can calculate the air kerma from the kV and mA and applying an inverse square law correction
- To convert this to the absorbed dose on the skin we need to correct for back scatter (which we do with a fiddle factor)
- The accuracy of this calculation is critically dependent on the focus to object distance (FOD):
    - Often not recorded
- **Proportional to kV^2**

Dose area product (DAP)

- DAP meters measure the **product of dose and beam area**
- Unit is **Gy / cm^2** (or submultiples)
- Has become a standard on fluoroscopic equipment
- Is measured by an ionisation chamber mounted on the collimator of the X-ray tube

Below are typical entrance skin doses, dose area products and effective doses from an assortment of radiographic and fluoroscopic examinations:

Examination	ESD (mGy)	DAP (Gy/cm^2)	Effective dose (mSv)
CXR (PA)	0.15	0.1	0.01
Lumber spine (lateral)	10	2	0.3
Pelvis (AP)	3.5	2	0.7
Intravenous urogram	20	12	2.5
Barium enema	30	22	5

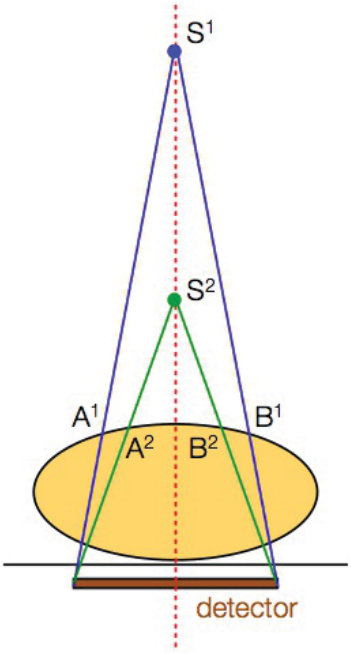
Factors affecting dose

Effect of altering either mAs or kV on patient dose

- Increasing mAs increases entrance and exit dose proportionately:
  - **Double the mAs = double the dose**
- Increasing kV hardens the beam (makes it more penetrating and increases the proportion of high energy photons that reach the detector:
  - A lower entrance dose is needed for the same exit dose
  - Thus, **increasing kV lowers dose**
- Remember that entrance surface dose **(ESD) is proportional to kV^2**

Focus-image distance (FID)

- Also known as focus-film distance (FFD)
- **Increasing FID deceases surface entrance dose** (and, to a degree, dose to deeper tissues)
- In order to keep the number of photons at the detector constant, we will need to increase the mAs but the dose saving from increasing the FID outweighs the dose increase from the mAs



At a shorter FID (S^2, photons are concentrated on a smaller surface area (defined by points A^2 and B^2) than they are at a longer FID (S^1). The larger surface area defined by A^1 and B^1 results in a lower skin dose.

Pregnant staff

Pregnant staff

- The **dose limit** for the foetus of a pregnant employee is equal to the limit for a member of the public (i.e: **1 mSv**)
- This limit applies over the declared term of pregnancy (the date the employee tells the employer)
- For diagnostic X-rays, the foetal dose is ~ 50% of the TLD dose
- For high energy radiation (e.g nuclear medicine) the dose is assumed to be the TLD reading

Female employees of reproductive capacity

- Have a **dose limit of 13 mSv (over any 3 consecutive month period) to the abdomen**
- This is to ensure that the employee does not receive a major part of the annual limit (20 mSv) over a short time period that coincides with conception and discovery of pregnancy
- Not really relevant in healthcare as the doses are generally much lower than this

Comforters and carers

Overview

- Exceptional circumstances occur in medicine where a member of the public might incur a radiation dose in excess of the dose limit
- An example is parents whose children are undergoing radio-iodine therapy
- **IRR99 permits the dose limit to be relaxed** for comforters and carers who knowingly and willingly are exposed to doses in excess of the limit
- The employer is required to:
  - Set a dose constraint (as much as 5 mSv)
  - Explain to the carer the dose and risks involved
  - Provide guidance on precautions to be taken to minimise dose

Practical aspects of radiation protection

Protection of staff

- Room design:
  - Exclusion from controlled areas when not required
  - Shielding:
    - Walls: **1 mm of lead = 120 mm of solid brick**. Usually 1 - 2 mm of lead is used
    - Doors incorporate lead
    - Lead glass screens
- Radiation sources:
  - **Primary beam**:
    - **Collimation** is crucial
    - Using an undercouch tube reduces both finger and whole body staff dose
  - **Transmitted radiation**:
    - Generally **<2% of the primary radiation** at the exit side of the patient
    - Biggest concern is finger dose
  - **Leakage**:
    - Restricted to **1 mGy/hr at 1m distance at maximum kV**
    - In practice is **generally <2%** (and this is due to scatter)
  - **Scatter**:
    - Caused by **Compton** interactions
    - The **main source of radiation to staff** and occurs in the part of the patient being imaged
- Practical aspects:
  - Inverse square law
  - Reduce exposure time

- Protective clothing:
  - **Lead aprons of 0.25, 0.35, 0.5 mm transmit 5%, 3% and 1.5% of radiation** respectively
  - **Thyroid collars** are usually **0.5mm lead** equivalent

Protection of the patient

- Must ensure the images are of sufficient quality so there is no need to repeat the examination
- **Collimate** to the area of interest to reduce dose
- Using a **magnified field of view** in fluoroscopy can reduce dose
- **Shielding** (e.g. gonadal protection in a hip radiograph)
- **Removal of the anti-scatter grid**:
  - Useful in fluoroscopic procedures
  - Reduces image quality (which could increase dose by increasing screening time)

Chapter 6

Radioactivity

Basics

Overview

- Nearly all the nuclides in the world are stable
- Apart from hydrogen, all the stable lighter nuclei contain equal numbers of protons and neutrons
- Heavier nuclei contain proportionately more neutrons
- Nuclides with the same atomic number (Z) but different atomic mass (A) are called isotopes
- Unstable nuclei (with a neutron excess or deficit) are radioactive and decay (transform) until they become stable with emission of any combination of radiation:
  - Alpha (α)
  - Beta (β^-)
  - Positron (β^+)
  - K-electron capture
  - Gamma (γ):
    - High energy photons released from an excited nucleus

Alpha decay

- Occurs in **heavier nuclei**
- Emission of an alpha particle (helium nucleus)

$$^{238}_{92}U \rightarrow {}^{234}_{90}Th + \alpha$$

- Due to their relatively high mass, electric charge and low speed, α particles readily interact with other atoms and are effectively stopped by a few centimetres of air

- **Z decreases by 2** and **A decreases by 4**. The daughter is therefore a **different element**

Beta decay

- Occurs in radionuclides with a **neutron excess**
- A **neutron changes into a proton**, **an electron** and an electron antineutrino
- The electron (the β- particle) is emitted immediately from the nucleus

$$n \longrightarrow p + \beta^- + \bar{\upsilon}_e$$

Electron antineutrino ($\bar{\upsilon}_e$)

- Think of it as the energy component
- Usually emitted immediately with the β- particle
- Some radionuclides (e.g. technetium) hold onto the antineutrino for a variable length of time before emitting it and are said to be **metastable**

$$^{99}_{42}Mo \xrightarrow{67\ h} {}^{99}_{43}Tc^m + \beta^- \xrightarrow{6\ h} {}^{99}_{43}Tc + \bar{\upsilon}_e$$

- **Z increases by 1** but **A stays the same**. The daughter is therefore a **different element**

Positron emission (B^+ decay)

- Occurs in radionuclides with a **neutron deficit**
- A **proton changes into a neutron and a positron**
- The positron is emitted from the nucleus with a high energy

$$p \longrightarrow n + \beta^+$$

- A clinical example is fluorine–18:

$$^{18}_{9}F \rightarrow {}^{18}_{8}O + \beta^{+}$$

- **Z decreases by 1** but **A stays the same**. The daughter is therefore a **different element**

K-electron capture

- Occurs in radionuclides with a neutron deficit
- The nucleus captures an electron from the nearest K-shell and combines this with a proton to form a neutron

$$p + e^{-} \rightarrow n$$

- The daughter nuclide:
  - Emits K-characteristic X-rays when an electron from the outer shell fills the hole left in the K- shell
  - May also emit gamma rays if left in an excited state

Measuring radioactivity

Activity

- The activity of a radioactive sample is the number of atoms that decay per second
- Unit is the **becquerel** (Bq):
  - **1 Bq = 1 disintegration per second**
  - Very small unit (natural radioactive content of the human body is ~2000 Bq)
  - We usually deal with megabecquerels (MBq)
- The count rate is the number of gamma rays that reach the detector per second:
  - Proportional to (but less than) the activity

Physical half life

- Half life ($\mathbf{t_{1/2}}$) is the time taken for a radionuclide's activity to **decay to half of it's original value**
- Is a **fixed** characteristic of a particular radionuclide:
  - Unaffected by temperature, chemical environment, etc

Important half lives

Half-life	Radionuclide	Clinical use
13 seconds	Krypton-81m	V/Q scans
2 hours	Fluorine-18	PET
6 hours	Technetium-99m	Bone scans
13 hours	Iodine-123	Thyroid investigation
67 hours	Molybdenum-99	Tc generator
73 hours	Thallium-201	Cardiac scans (defunct)
78 hours	Gallium-67	Being replaced by PET
8 days	Iodine-131	Thyroid ablation
200,000 years	Technetium-99	Tc-99m daughter product

Effective half life

- A pharmaceutical labelled with a radionuclide is called a **radiopharmaceutical**
- If the pharmaceutical alone is administered, it will eventually be eliminated from the body and, therefore, is said to have a **biological half life** (t_{biol})
- If a radiopharmaceutical is left in a vial, it will decay with its **physical half life**, t_{phys}
- If a radiopharmaceutical is given to a patient, both t_{biol} and t_{phys} impact on its **effective half life**:

$$\textbf{effective half life} = 1 / t_{biol} + 1 / t_{phys}$$

Radiopharmaceuticals

The "ideal" radiopharmaceutical

- Radionuclide component:
  - A physical half life of a few hours:
    - If it's too short then more activity would have to be prepared than is actually injected
  - Decays to a **stable daughter product** (i.e. one with a long half life)
  - **Emits gamma rays only**:
    - α and β particles would not form an image
    - In the **energy range 100–300 keV** (ideally 150 keV)
    - Ideally single energy only
  - Easily attached to a pharmaceutical without altering it's metabolism
  - **High specific activity** (high activity per unit volume)
  - Generator-produced
- Pharmaceutical component:
  - **Localise** specifically and quickly in target organ
  - Appropriate biological half life
  - **Low toxicity**
  - Cost effective

Producing radionuclides

- There are three main ways to produce radionuclides:
  - Radionuclide generator (e.g. $^{99}Tc^{m}$)
  - Cyclotron (e.g. ^{18}F)
  - Nuclear reactor (e.g. ^{99}Mo)

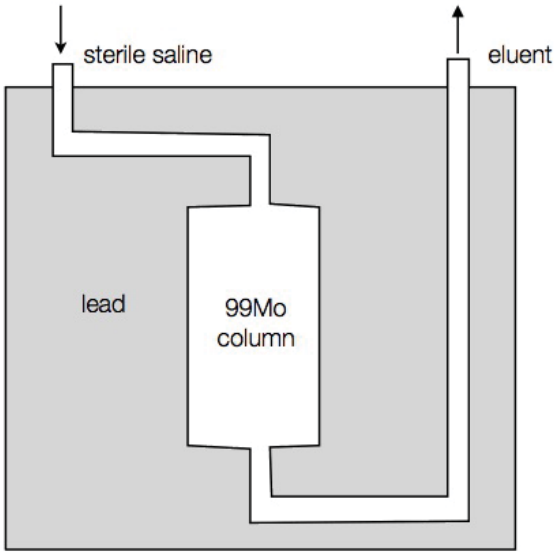
Technetium–99^m

- $^{99}Tc^m$ is used in 90% of radionuclide imaging
- Emits **140 keV gamma rays**
- Short half life of **6 hours**
- Produced on site from its parent molybdenum in a technetium generator

The technetium generator

- $^{99}Tc^m$ is produced in a generator from it's parent **^{99}Mo**
- ^{99}Mo is produced in a **reactor** and has a **67 hour half life**
- When the generator is left alone for a period of time, the level of activity from the $^{99}Tc^m$ reaches a **transient equilibrium**:
  - This means that the $^{99}Tc^m$ is decaying as quickly as it's being formed
- We **elute** the $^{99}Tc^m$ from the generator by washing it off the column with sterile saline:
  - This gives us **sodium pertechnetate** which has a physical half life of 6 hours
- **Technetium generators usually last about a week**

Schematic of a technetium generator



Cyclotron

- A type of particle accelerator
- Spirally accelerates ions (charged particles) within a magnetic field
- Ions are steered into a target to produce the desired radionuclide
- An example is the **creation of ^{18}F by bombarding ^{18}O with protons**

Nuclear reactor

- ^{235}U is split (undergoes fission) when bombarded by a neutron
- Fission results in the release of many more neutrons
- Suitable materials can be lowered into the reactor so that they are irradiated by the neutrons
- **Neutron capture** results and radionuclides are formed (e.g. ^{98}Mo converted to **^{99}Mo**)
- Some **fission by-products** are also useful in medicine (e.g. **^{131}I**)

Chapter 7

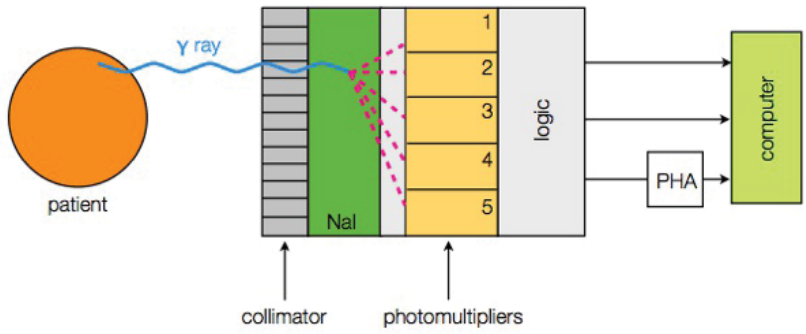
Radionuclide imaging

The gamma camera

Overview

- Patient is given a radiopharmaceutical, usually by injection
- The radiopharmaceutical concentrates in the target organ or tissue
- Gamma rays are emitted by the radionuclide and are detected by a gamma camera

Schematic of a gamma camera



The gamma camera

- The gamma camera is surrounded by **heavy lead shielding** to prevent interference
- Is comprised of several components:
  - Collimator
  - Crystal
  - Photomultipliers
  - Pulse logic
  - Pulse height analyser

Collimator

- A lead disc **25mm thick** and **400mm in diameter**
- Contains **20,000 closely packed hexagonal or circular holes 2.5 mm in diameter** separated by **0.3mm thick septae**
- The septae absorb almost all gamma rays attempting to pass through the holes obliquely
- Each hole effectively only absorbs gamma rays from within its direct line of sight

Crystal

- **500mm diameter**
- **10mm thick**
- Made of **thallium-activated sodium iodide** (NaI):
- High Z (good photoelectric absorption)
- Variable **efficiency** in absorption of gamma rays and emission of light photons:
  - **90% for $^{99}Tc^{m}$**
  - 30% for ^{131}I
- Fragile
- **Hygroscopic** (absorbs moisture from the air):
  - Is encapsulated within an aluminium cylinder for protection
- Each absorbed gamma ray produces **5000 light photons**:
  - ~4000 reach the photomultipliers (of which there are many closely packed together)

Photomultipliers

- Comprised of an evacuated glass envelope with a **photocathode** on the crystal side and a positive electrode (**anode**) on the other side

- Photocathode:
  - Absorbs the emitted light from the crystal and emits photoelectrons
  - **1 electron per 5–10 incident light photons**
- En route from photocathode to anode, the photoelectrons strike several **dynodes**:
  - Each dynode releases 3–4 electrons
  - Act as an **amplifier**
  - For each photoelectron originally emitted, about 1 million electrons reach the anode:
    - This generates a sufficient voltage to be measured
- **Typical tube voltage is ~1 kV**

Pulse arithmetic circuit (logic)

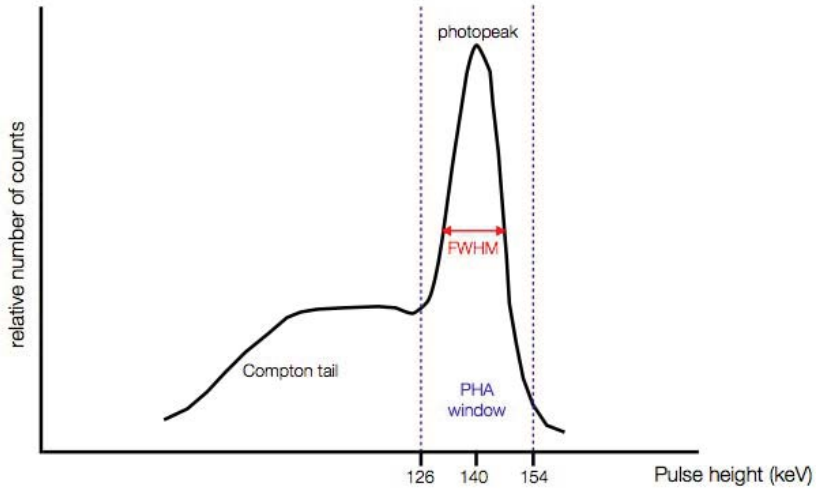
- Combines the pulses received from the photomultipliers and generates (based on in-built equations) three voltage pulses: X, Y and Z
- X and Y:
  - The horizontal and vertical coordinates of the light flash in the crystal
- Z:
  - The pulses from all the photomultiplier tubes (within a tiny space of time) are summed and treated as if they were one large photomultiplier to give us the total photon energy measured in the crystal within that small space of time
- The **height of the Z-pulse voltage is proportional to the gamma ray energy** (in keV)

Pulse height spectrum

- γ rays are scattered within the patient:
  - This means that γ rays that have originated outside the line of sight of the collimator can still enter the collimator
  - **Scattered γ rays have less energy**

- γ rays may lose energy through Compton interactions in the crystal before eventually escaping:
  - These rays will produce pulses of reduced height

$^{99}Tc^{m}$ Pulse height spectrum



- Since a large number of γ rays are emitted in succession from the patient, the Z-pulses vary in height. This is plotted on the **pulse height spectrum** graph above
- The **photopeak** corresponds to γ rays that have come from the patient and have not suffered Compton scattering
- Theoretically, the photopeak should be very narrow:
  - Due to transmission and detection factors, it actually has a measurable width expressed as the **full width at half maximum** (**FWHM**)
- The Compton tail on the left of the spectrum represents pulses of a lower energy that have suffered Compton interactions in the patient or the crystal
- **Only pulses in the photopeak are useful** for locating the position of radioactivity in the patient:
  - A **pulse height analyser** (PHA) is used to reject those within the Compton tail

Pulse height analyser (PHA)

- **The PHA only lets through pulses whose energy lies ± 10% of the photopeak**
- Those pulses that are let through are known as **counts**
- Pulses below the photopeak are useless as they are a result of Compton scatter
- There are two reasons for the very high energy pulses (those to the right of the photopeak):
  - Could be caused by the simultaneous detection of two or more γ rays from the patient:
    - Rejecting these does result in a loss of information
  - Could be caused by cosmic rays
- Using $^{99}Tc^{m}$ as an example:
  - **Window is set at 126–154 keV** (± 10% of $^{99}Tc^{m}$ γ ray energy of 140 keV)
  - Note that a 140 keV γ ray scattered at 45° will only lose 10 keV of energy:
    - This would be allowed to pass through the PHA and, therefore, degrade image quality

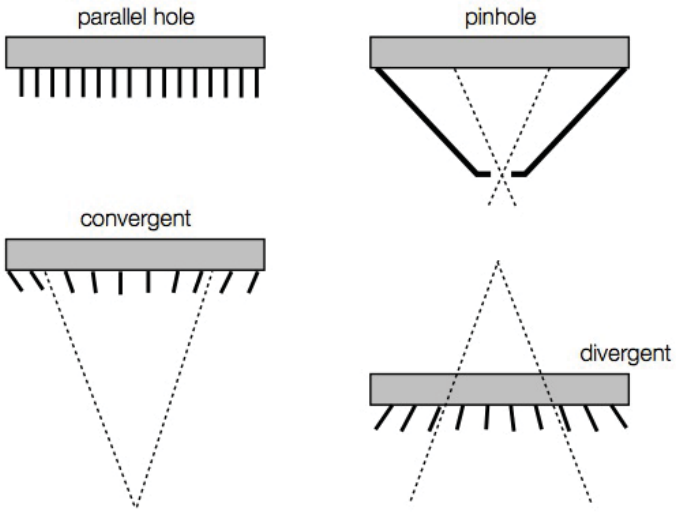
Construction of the image

- Counts are stored in a 2D matrix (X rows and Y columns)
- The more counts in a cell in the matrix, the darker final image in that pixel
- Each image frame usually contains 500,000 - 1 million counts

Gamma camera collimators

Overview

- Gamma cameras are usually supplied with a range of removable collimators
- Which collimator is used depends on the clinical application
- There are several different collimator designs:
  - Multi-hole:
    - Non-distorting:
      - **Parallel hole**:
        - Equal field of view (FOV) at all distances
    - Geometrically distorting:
      - **Divergent**:
        - **Minifies** the image permitting a larger FOV
        - Useful for imaging large areas with a small crystal
      - **Convergent**:
        - **Magnifies** the image (but reduces the FOV)
        - Useful for imaging small structures or children
  - **Pinhole**:
    - **Magnifies** and inverts the image
    - Useful for imaging small organs (e.g. the thyroid)
- **High resolution** collimators have **more numerous and smaller holes but lower sensitivity**
- High sensitivity collimators have fewer, larger, holes and a lower resolution


parallel hole
pinhole
convergent
divergent

Single photon emission computed tomography (SPECT)

Overview

- SPECT is a tomographic imaging technique with some similarities to conventional CT
- Rather than detecting X-ray attenuation through a projection, we collect gamma ray counts
- Essentially, a **gamma camera rotates around the patient** every 3° (or 6°), pauses for 30 seconds to acquire a projection and then advances another 3° (or 6°):
  - This is done 120 (or 60) times to cover a full 360° of the patient
  - We then reconstruct the data using either **filtered back projection** or **iterative reconstruction**:
  - Exactly the same process as for CT reconstruction
- **Noise is a major issue** with SPECT:
  - Largely as counts are only made for 30 seconds in each projection (low number of photons)
  - Movement artefact would occur if we waited longer than this
  - We usually use a **pixel matrix of 64x64** or **128x128** to reduce noise:
    - Gives SPECT a **low spatial resolution** (worse than planar imaging)

Types of SPECT scanner

- Single head:
  - Based on a standard gamma camera mounted on a rotating gantry
  - Largely superseded by multiple head scanners
- Multiple head:

- Still uses standard gamma camera technology but there are two (or even three) cameras orbiting the patient
- **Quicker acquisition**
- **Reduces motion artefact**

- SPECT/CT:
  - Have a dual head gamma camera as well as a fan beam X-ray source
  - Capable of normal CT and SPECT
  - Allows the **fusion** of SPECT (functional) and CT (anatomical) information to produce composite images

Positron emission tomography

Overview

- PET is a tomographic radionuclide imaging technique
- Signal is determined by the activity of radiopharmaceutical in a voxel of tissue
- Relies on the **detection** of the pair of photons resulting from the **annihilation** of a positron and an electron:
  - PET radiopharmaceuticals therefore decay by positron emission
  - Most commonly used is **^{18}F**

Positron annihilation:

- A positron (β^+) is the antiparticle of an electron:
  - Same mass but opposite charge
- Positrons do not exist long in our universe after creation:
  - Typically travel for **~2mm** before they **annihilate** with an electron
- Annihilation causes **destruction of the positron and the electron** and the release of **two 511 keV gamma photons in** (nearly) **opposite** (180°) **directions**

Fluorine–18

- The most widely used PET radiopharmaceutical is ^{18}F
- Produced in a cyclotron
- Half-life is ~2 hours
- Commonly administered as fluorodeoxyglucose (FDG), a glucose analogue
- Decays (via positron emission) to ^{18}O

The PET scanner

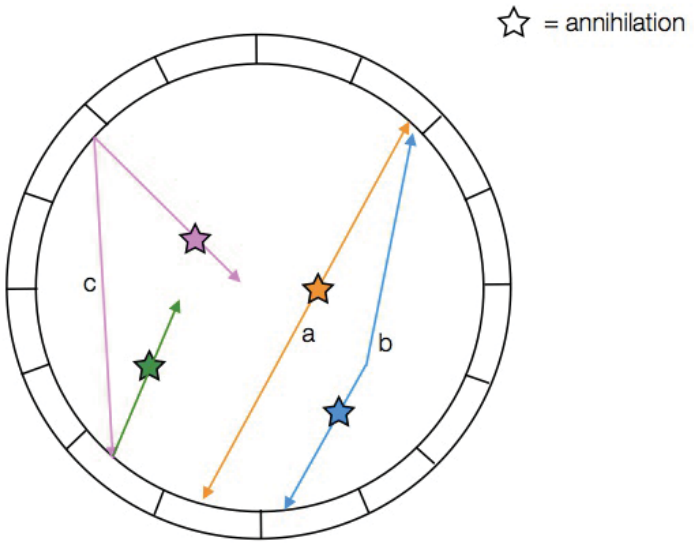
Scintillator material

- PET scanners comprise a **ring** of a large number of **solid scintillation detectors** (**10,000 - 20,000**)
- Requirements:
  - **High detection efficiency at 511 keV**
  - **Very short scintillation decay time**
  - **High energy resolution** (lots of scintillation photons per incident gamma photon)
  - High photoelectric : Compton scatter ratio
- Usually made from **BGO** (**bismuth germanate**):
  - Other materials include LSO (lutetium oxyorthosilicate) and GSO (gadolinium oxyorthosilicate)

Detector configuration

- Each scintillator block is coupled to photomultipliers which convert the emitted light into electrical charge
- Detectors can be separated by thin lead septae to reduce cross talk between them:
  - These are retractable on some scanners
  - Retracting the septae increases sensitivity but also increases the number of false coincidences

Coincidence detection



- A **coincidence** occurs when **two γ rays are detected along a straight line** (**line of response**, **LOR**) within a very narrow time frame (**~10ns**)
- Each detector can operate in coincidence with ~50% of the detectors that face it in the ring:
  - This means that the **patient is criss-crossed** by hundreds of LORs
- There are three types of coincidences:
  1. **True coincidence** (line a):
     - The simultaneous detection of two unscattered photons from a single annihilation
  2. **Scatter** (line b):
     - One or both photons are Compton scattered in the patient and both are detected
  3. **Random** (line c):
     - Photons from different annihilations happen to be detected simultaneously
- Thin lead septae may be used between each ring of detectors to reduce scatter and random events between rings

Acquisition of projection data

- In theory, there are as many LORs as there are pairs of detectors
- When a coincidence is detected between two detectors, the computer determines the LOR and stores one event in a pixel matrix that corresponds to that event
- Acquisition continues until a sufficient number of coincidences have been recorded (millions)
- A set of parallel LORs defines a projection
- In PET, all projections are acquired simultaneously

2D and 3D acquisition

- Data may be acquired in 2D or 3D:
  - Some PET scanners are designed for operation in only one of these modes whilst others can use either
- **2D acquisition**:
  - Uses thin **annular collimators** placed immediately in front of the detectors
  - Coincidences are confined annihilations that originated in one particular slice
  - **Reduces scatter and random coincidences**
- **3D acquisition**:
  - Does not use a collimator
  - Coincidences from a much greater volume of tissue are used
  - Greatly **increases total count** rate and thus **sensitivity**
  - Useful when there is little scatter produced or the amount of radiopharmaceutical administered is small (e.g. brain imaging, paediatrics)

Image reconstruction

- Uses filtered back projection, just like SPECT and CT
- The coincidence data are organised into a 2D matrix where each row represents a projection at a particular angle. This is the **sinogram**

PET image quality

Contrast

- PET (and SPECT) have much better image contrast than planar radionuclide imaging:
  - Largely due to the fact that tomographic imaging removes overlap
- Image contrast is reduced by scatter:
  - Greatest towards the centre of the image

Spatial resolution

- Typical **axial resolution is 4–6mm** (**better at the centre** than the periphery)
- This is determined by the following factors:
  - **Positron range**:
    - The distance from the site of creation to annihilation
    - The longer the range, the lower the resolution
    - Average range of a positron quoted as **0.6–2mm**
  - **Non–180° travel**:
    - If the positron and electron have residual **momentum** at the time of annihilation, the annihilation photons will not travel at exactly 180° to each other
  - **Detector element size**:
    - The **smaller** the detector, the **better** the resolution
    - 5mm is typical
  - **Detector depth**:
    - The **thicker** the detector, the **lower** the resolution
    - Resolution is therefore better at the centre of the detector ring

Quality assurance

Uniformity

- A conventional gamma camera should give a uniform response to a uniform field and is checked daily by **flooding**
- The flood plate is usually a sealed plate (larger than the field of view) containing **cobalt–57**:
  - Has similar gamma emission to, but longer half-life than, $^{99}Tc^{m}$
  - Test is performed with the collimator in place
- Area of **reduced counts** in the image = **defective photomultiplier**
- **Linear defect = cracked crystal**

Detector uniformity in PET scanners

- Checked in one of two ways:
  1. With a long lived source (^{68}Ge or ^{137}Cs) mounted on the gantry and rotated around the field to uniformly expose all of the detectors
  2. With a standard phantom with a centrally located positron source within it

Patient dose

Typical activity and effective doses for adults

Site	Radiopharmaceutical	Activity (MBq)	Effective dose (mSv)
Bone	Technetium-99m diphosphonate	600	5
Lung ventilation	Krypton-81m gas	6000	0.1
Lung perfusion	$^{99}Tc^{m}$ HSA macroaggregates	100	1
Kidney	$^{99}Tc^{m}$ DTPA	300	2
Kidney	$^{99}Tc^{m}$ MAG3	100	0.7
Thyroid	Iodine-123	20	4
Heart	$^{99}Tc^{m}$ MIBI	400	3
Brain	^{18}F FDG	400	8

It's clear from the table above that **most radionuclide studies are between 1–5 mSv** (except for PET FDG imaging which is higher).

Chapter 8

Radiation protection framework

Justification, optimisation and dose limitation

Overview

- UK legislation is based on recommendations made by the International Commission on Radiological Protection (ICRP)
- Latest guidelines are ICRP 103 but UK legislation based on ICRP 60
- ICRP introduced 3 basic principles:
  - **Justification**
  - **Optimisation**
  - **Dose limitation**

Justification

- Risk is proportional to dose and is greater for children than adults
- It's easier to justify the use of radiation for seriously ill patients than minor complaints
- The evidence must be strong to use radiation in healthy people, for instance in screening

Optimisation

- ICRP states that doses should be **as low as reasonably achievable** (**ALARA**)

Dose limitation

- In addition to justification and optimisation, there exist **dose limits**, an excess of which **cannot be justified** no matter how great the benefit
- Apply **only** to **staff** and the **general public**

- **Does not apply to patients** (their dose is regulated by justification and optimisation)
- Implicit in the limits is the concept of **tolerable risk**

Annual dose limits

	Employees (mSv)	**Public (mSv)**
Whole body (effective dose)	20	1
Lens of the eye	150	15
Skin	500 / cm^3	50 / cm^3
Extremities	500	50
Abdomen of fertile female employee	13	
Foetus of pregnant employee	1	

The ionising radiations regulations 1999

Overview

- Known as IRR99
- A **framework ensuring radiation exposure from work activities is as low as reasonably practicable**
- Imposes responsibilities on the employer who are liable under **criminal law**
- Employers can delegate tasks to employees but not legal responsibilities
- **Not applicable to patient dose** (that's IRMER)
- Consists of 8 major areas:
  1. Appointment of a radiation protection advisor
  2. Designation of work areas & local rules
  3. Appointment of radiation protection supervisors
  4. Dose limitation
  5. Staff classification and dose monitoring
  6. Provision of personal protective equipment
  7. Radiation incidents
  8. Quality assurance

Radiation protection advisor

- Almost invariably a medical physicist
- Responsible for **ensuring employers compliance** with IRR99
- Must be accredited by the Health & Safety Executive (HSE) and renew this every 5 years

Designation of work areas

- **Controlled area**:
  - A person working is likely to **exceed 3/10th of any dose limit**
  - **Special working procedures** are required to restrict exposure or accidents
  - The **external dose rate could exceed 7.5µSv / hr** averaged over the working day
  - Examples include the X-ray room or the injection room for radionuclides
- **Supervised area**:
  - If it's necessary to check that the exposure conditions aren't such that the area might need to be classified as a controlled area in the future
  - A person is likely to **exceed 1mSv / yr** (effective dose)
  - A person is likely to **exceed 1/10th of any dose limit**

Local rules

- **Describe safe methods of working with ionising radiation**
- Required for all controlled areas and some supervised areas
- Should be brief
- Must be brought to the attention of employees
- Usually consist of:
  - Name of the radiation protection supervisor
  - Description of controlled and supervised areas
  - Identification of responsibility for radiation protection
  - Contingency plans
  - Arrangements for pregnant staff
  - Practical instructions (e.g. where to stand in the room)

Radiation protection supervisors

- Ensure that the local rules are followed

Classified persons

- Applies when staff are **likely to exceed 3/10th of any dose limit** (i.e. 6mSv effective dose)
- Must be over 18
- Employee must be informed that they are a classified person
- Must be medically fit
- Rarely needed for healthcare workers as <1% receive >1mSv / yr

Dose monitoring

- IRR99 only requires employers to monitor classified worker's doses
- In practice, all employees working in controlled areas are given dosimeters
- Dosimeters should be provided by an HSE **approved dosimeter service** (ADS):
  - Records are sent to the HSE annually
  - Records are kept for 50 years
- **Outside workers**:
  - These are classified persons who work for **more than one employer**
  - ADS provides worker with a **radiation passbook** where doses are recorded

Personal protective equipment (PPE)

- The employer must provide adequate PPE (e.g. lead aprons) and the employee must use them
- It is the employer's responsibility to maintain and store the PPE

Radiation incidents

The employer must inform the HSE if any of the following occur:

1. An individual (including the public) receives a **dose greater than any relevant dose limit**
2. A **radiation source is lost**, **stolen** or **spilt** (causing significant contamination):
    - IRR99 specifies the levels of spill to be reported (e.g. >100MBq of $^{99}Tc^{m}$)
3. A patient receives a **dose much greater than intended** because of an equipment fault

Quality assurance progammes

- To ensure that the examination contains enough information to diagnose whilst resulting in the lowest amount of radiation and inconvenience to the patient and cost to the Trust
- Covers a number of areas:
    - Procedures (e.g. checking for pregnancy)
    - Equipment maintenance and calibration
    - Audit
- Multidisciplinary:
    - Users (e.g. radiologists, radiographers)
    - Support (e.g. physicists, service engineers)

The ionising radiation (medical exposures) (amendment) regulations 2006

Overview

- For every medical radiation exposure, IRMER makes the employer responsible for:
  - Providing a framework to work within
  - Producing referral criteria
  - Producing written protocols and procedures for the exposure
- IRMER also introduces three types of duty holders:
  - **Referrer**
  - **Practitioner**
  - **Operator**

Employer's responsibilities

- Correctly identify patients
- Define duty holder roles
- Check pregnancy and breast-feeding status of female patients
- Provide quality assurance programs
- Provide diagnostic reference levels for examinations
- Provide patient information (e.g. in the form of post-examination leaflets)
- Minimise risk of accidental or unintended patient exposure

Referrers

- The individual who **initiates the radiological request**

- The employer defines who may act as a referrer and any restrictions that may be imposed by way of a local protocol:
  - E.g. only consultants may request a CT scan or nurse practitioners can only X-ray limbs, etc
  - Usually based on RCR referral guidelines (iRefer)
- The referrer **must provide sufficient clinical information** for the IRMER practitioner to justify the examination

Practitioners

- The individual who **justifies the examination**
- Require theoretical and practical training
- Usually radiologists and (for some examinations) radiographers

Operator

- Very broad definition
- **A single examination may have several operators**
- Encompasses individuals who:
  - Check the patient ID
  - "Press the button" to deliver the exposure
  - Maintain the equipment
- Operators do not need to be registered healthcare professionals

Diagnostic reference levels (DRLs)

- These are **doses for typical examinations for standard-sized patients**
- Are used as an aid to optimisation
- Are set locally (by the employer) and reflect local practice

- The local value should not be greater than the national DRL unless it can be clinically justified
- Must be able to **audit** the levels
- Usually defined in terms of **dose area product** (DAP), **screening time** or **radioactivity administered**

Exposures for research, health screening and medico-legal purposes

- I.e. examinations where the patient is unlikely to receive any direct health benefit
- Included in IRMER but really only to make sure that there are local procedures in place
- Research examinations must be approved by an ethics committee and be voluntary
- Patients must be informed of the risks
- Must be dose constraints in place for healthy volunteers

The MARS regulations and ARSAC

The medicines (administration of radioactive substances) regulations

- Also known as the MARS regulations
- Applies only to doctors and dentists
- A collection of regulations designed to **protect patients**
- Similar to IRMER but concerns the administration of **radioactive substances**
- All MARS practitioners must have an **ARSAC certificate**

Administration of radioactive substances advisory committee (ARSAC)

- This committee was set up under the MARS regulations
- Issues:
  - Certificates to practitioners
  - Guidance on normal and maximum levels of activity to be used
- Part of the Department of Health

ARSAC certificates

- Specify:
  - **Location** (e.g. a specific hospital)
  - **Radiopharmaceutical**
  - **Purpose**
- Require renewing every 5 years

The radioactive substances act 1993

Overview

- Now superseded by the **environmental permitting regulations 2010**
- Concerned with the **protection of the population as a whole and the environment**
- Enforced by the **Environment Agency**
- Controls the amount of radioactive substances per location and their disposal:
  - Need to know the activity of each radionuclide disposed per month:
  - Liquids (poured down sink, into sewers)
  - Solids (disposed of by outside contractors)
  - Gaseous waste

Medical and dental guidance notes

Overview

- Published by the Institute of Physics and Engineering in Medicine
- Concerns radiation protection regulations and the practical application of radiation protection
- Are not legal requirements but represent standards compliant with IRR99
- Extensive. Below are some examples

The X-ray tube

- **Leakage < 1 mGy / hr 1m from the focus**
- **At least 2.5mm aluminium equivalent filtration** (1.5mm for < 70kV dental equipment)
- The position of the focus should be marked on the tube casing

Warning signals

- There should be an indicator light on the control panel to show when the X-ray beam is switched on

Exposure switches

- They should require that constant pressure is maintained for continuous exposure:
  - CT scanners are an exception (single press starts the sequence)
- For mobile equipment:
  - The position of the exposure switch should be positioned such that the operator is > 2m from the X-tube:

        - In practice, they are usually on a cable
    - There should be a key-operated switch to prevent unauthorised use

Shielding

- **Image intensifier housing should have > 2mm lead equivalent shielding**
- For an undercouch fluoroscopy system, there should be a lead apron suspended from the bed

Fluoroscopy dose rates

- **Skin entrance dose rates should be < 100mGy / min for any field size**
- Remedial action is needed if the dose is > 50mGy / min for the largest field size

Chapter 9

X-ray computed tomography

The CT scanner

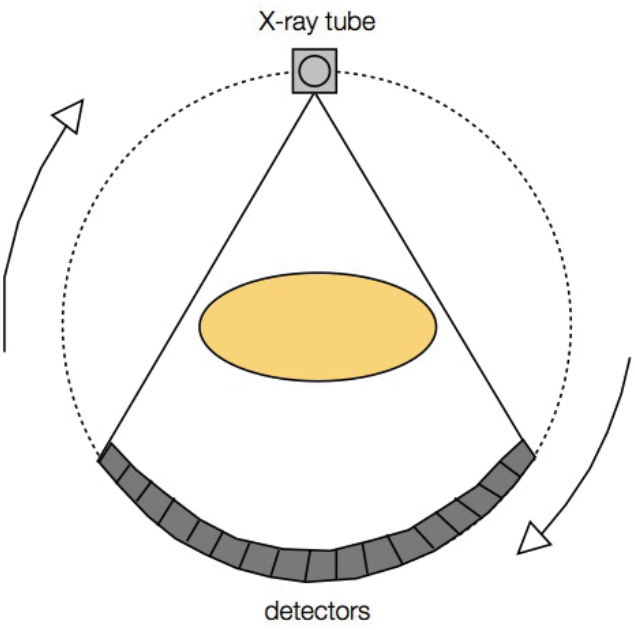
Scanner generations

- First and second generation:
  - Known as the rotate-translate type
  - Single X-ray source and either one (first) or a bank of up to 30 (second) detectors
  - Both moved around the patient in 1° increments
  - Data was collected through a 180° rotation
- **Third generation**:
  - The **rotate-rotate** type
  - Most modern scanners
  - Large number of small detectors arranged in an arc covering a complete patient cross section
  - Allows **continous data collection through a full 360° rotation**
  - Permits scan times of less than 0.4s
- Fourth generation:
  - Detectors are arranged in a stationary ring around the patient and the tube rotates
  - The outer part of the fan beam is always outside the patient and can be used to measure unattenuated radiation and self-calibrate
  - Requires vastly increased number of detectors which is prohibitively expensive
- Fifth generation:
  - Also known as an **electron beam scanner**
  - Electrons focussed on and swept round a high voltage target ring to produce X-rays
  - No mechanical parts thus rapid scan times possible

Configuration of a typical CT scanner

- The **rotational axis is the Z-axis**
- X-ray beam is collimated as a wide fan shape big enough to cover the patient in cross section:
  - Has a narrow width parallel to the Z-axis
  - For a single slice scanner this width defines the slice thickness
- Behind the patient is an arc of **500–1000 detectors**:
  - The **radius of the arc is equal to the focal distance** (thus each detector is the same distance from the source)
- Patient lies on a couch that can be moved longitudinally through the gantry aperture
- The gantry is normally perpendicular to the couch but can be tilted up to 30°:
  - Mainly used for head scanning so the scan plane can be made parallel to the skull base

Simple third generation scanner



The CT X-ray tube

- Tubes for CT scanners have to be capable of producing prolonged exposure times at high mA
- Usually have two **focal spot sizes**, the smallest being ~ **0.6mm**
- Usually operate at 120 kV but the range is between 80 - 140kV
- Have heat capacities in excess of 4MJ

Filtration

- CT algorithms rely on the X-ray beam being **monoenergetic** which, of course, it isn't
- To approximate a monoenergetic beam more closely, the X-ray beam is deliberately hardened by adding filtration:
  - Usually **0.5mm copper** (**equivalent to 8mm aluminium**)
  - Produces a mean energy of ~ 70 keV from 120 kV
- Since the patient cross-section is usually elliptical, X-rays at the periphery tend to pass through less tissue:
  - To compensate for this difference in attenuation across the field of view, some CT scanners have a **bow tie filter** after the X-ray tube:
    - A filter that is thin in the centre and thick at the edges to artificially harden the beam at the edge

Collimation

- A collimator is mounted on the X-ray tube
- The beam is collimated to a fixed width (usually ~ 50cm)
- In the case of single slice scanners, collimation also defines the slice thickness (0.5 - 20mm)
- Single slice scanners also have a post-patient collimator:

- Mounted in front of the detectors and used to reduce scatter reaching the detectors when the slice thickness is less than the detector width
- Not needed for multi-slice scanners as the full width of the detectors in each row is used to form the image and shielding from scatter would also shield from direct radiation

Detector requirements

- Must be **small** to allow good spatial resolution (single slice scanners have 600–900 in an arc)
- **High detection efficiency** for X-rays in the CT energy range
- **Fast** response time (with negligible afterglow)
- **Wide dynamic range**:
  - Wide range of X-ray intensities (e.g. from zero attenuation when the beam passes to the side of the patient to very high attenuation in a lateral projection of a heavy patient)
- **Stable**, noise-free response

Types of detectors

- Older single slice scanners used ionisation chambers:
  - Filled with high atomic number gases (xenon or krypton) at high pressure (20 atm)
  - Incident X-rays ionise the gas and produce a charge at the collection electrode
  - Only ~ 60% efficient
  - Not suitable for use in multislice scanners
- All new scanners use **solid state detectors**:
  - Consist of a scintillant (e.g. **bismuth germinate**) and embedded photodiode to detect output
  - **Very high detection efficiency** (**~98%**)

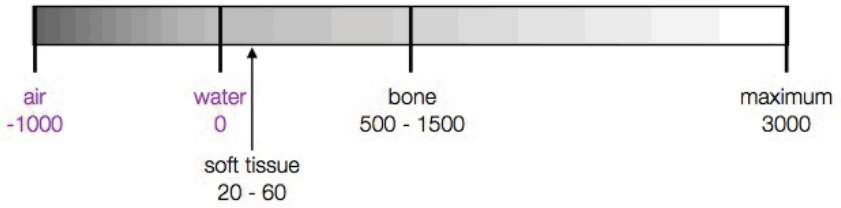
The CT image

The CT number

- Most commonly a 512x512 matrix (although 1024x1024 is sometimes used)
- Technically composed of voxels (volume elements) rather than pixels:
  - Each voxel has three dimensions with a depth equal to slice thickness
- Each displayed pixel stores a value known as the CT number (CTn):
  - The **average linear attenuation coefficient of the tissues within the voxel**
  - Also known as the **Hounsfield unit**

$$\mathbf{CTn = 1000 \times (\mu_T - \mu_W) / \mu_W}$$

(where μ_T = average linear attenuation coefficient of the tissues, μ_W = linear attenuation coefficient of water)



Windowing

- The CT number scale typically covers a 12 bit range:
  - 12 bits = 2^{12} = 4096 shades of grey (–1024 to 3071)
- The human eye cannot distinguish this many shades of grey (its range is about 50)
- If we displayed –1000 as black and +3000 as white then very little differentiation would be seen between soft tissues (whose CT numbers are within a limited range)
- We therefore use windowing:

- Many different presets depending on the region of interest
- We set a CT number level that we're interested in and a range either side. Anything below this window is black and above is white

Image reconstruction

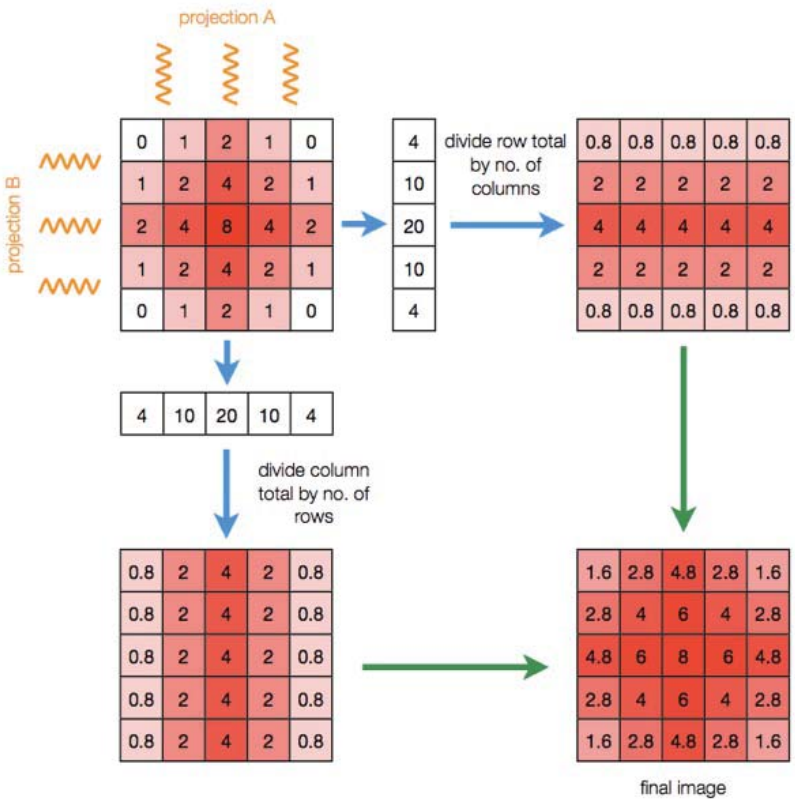
Overview

- In a modern scanner, during acquisition, X-rays are produced continuously
- A series of projections through the patient are collected at discrete angles (e.g. every 0.5°)
- If there are 900 detectors, that's 900 * (360 * 2) = 648,000 measurements per rotation
- Think of the fan beam as composed of many **narrow pencil beams**, one to each detector:
  - Each detector, therefore, samples a point on the projection that represents the total attenuation along the pencil beam path from source to detector
- For every 360° rotation, an individual voxel is traversed by one or more pencil beams for every measurement taken:
  - The attenuation of each voxel, therefore, contributes to the measured transmission for a large number of the ray sums
- In theory, the CT numbers for each voxel could be calculated from a series of simultaneous equations but, for a number of reasons, this is not done and instead we must use a mathematical **algorithm**

Back projection

- Imagine our voxel matrix looking like a grid composed of rows and columns
- Each cell in the grid represents a pixel
- The value of each cell is the attenuation of that voxel
- The CT scanner can't give us the value of each cell but instead gives us the total of each row and column
- It is from these row and column totals that we need to calculate the individual cell values

Illustration of back projection



Back projection disadvantages

- Back projection does not return a "true" image of the scanned object
- We can see that the final image above is blurred:
  - Partly because **too few projections** (2) have been used:
    - In real life, we take about 1000 projections per rotation
  - Secondly because **all the points along the pencil beam contribute to the projection values**
- If back projection alone was used, the resulting image would always be blurred, even if a large number of projections were taken. The solution is to use **filtered back projection**

Filtered back projection

- The most common method used for image reconstruction
- For each pencil beam measurement, the values that are projected back use the data not just from the pencil beam itself but also from the neighbouring pencil beams
- Similar to back projection but each projection is first multiplied by a filter or **kernel**
- Different kernels exist (e.g. to **enhance edges** or to **detect low contrast detail**)
- Advantages:
  - **Large reduction in blurring**
- Disadvantages:
  - **Noise is amplified**
  - Sharp edges in the scanned structure (e.g. a fluid surface) can cause line artefacts

Helical (spiral) CT

Overview

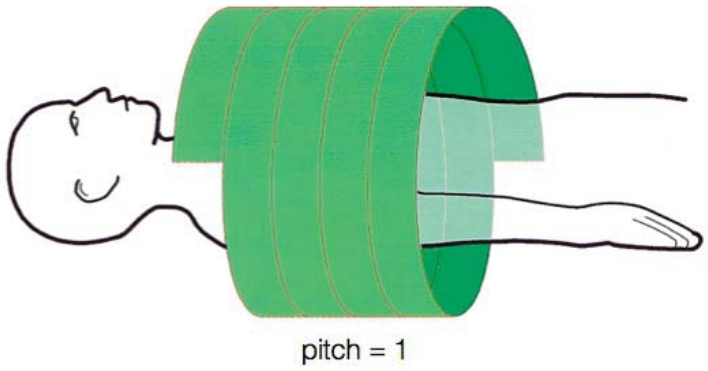
- The first helical scanner became available in 1990
- A major advance in scanner design was **slip ring technology**:
  - Permits **continuous**, **unidirectional**, **rotation** of the gantry
  - **Faster** rotation times
- A helical CT scan works by continuously exposing X-rays as the gantry rotates and the patient moves through it
- Helical CT has largely replaced conventional axial scanning

Pitch

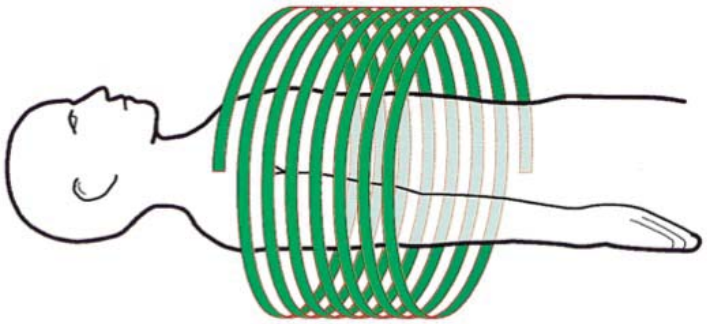
- Picture a helical scan as a ribbon wrapped around the body
- If the scanner had a slice thickness of 10mm, a rotation time of 1 second and the table moved at 10mm / sec then the edges of a 10mm wide ribbon would be touching and there would be complete coverage of the body. This would be a pitch of 1

$$\text{pitch} = \text{table movement per rotation} / \text{slice width}$$

- If we increased the speed the table moved then the ribbon would be "stretched out" and would no longer completely cover the body. This is a pitch > 1
- **Pitch is normally kept less than 1.5** (greater pitch requires more interpolation and, thus, reduces resolution)



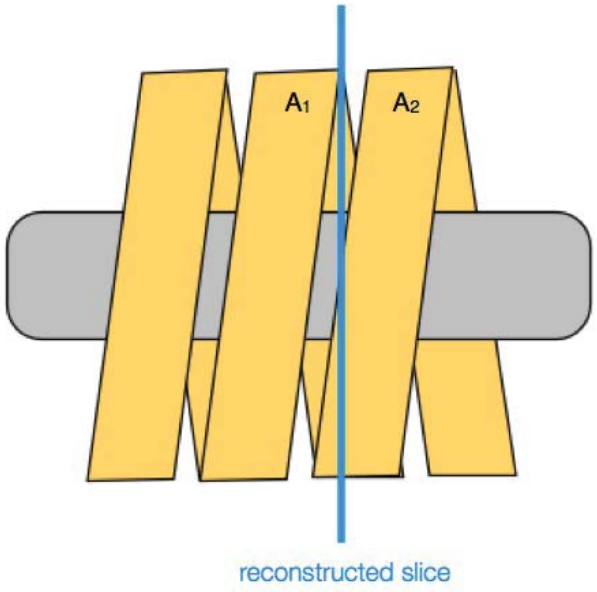
pitch = 1



pitch > 1

Linear interpolation

- Only 1 of the 1000 projections in a full rotation goes exactly through the centre of the reconstructed slice



- The data to be used to reconstruct the slice above would be interpolated from data collected at positions A_1 and A_2:
  - the relative weight given to each position depends on the relative distances from the plane of reconstruction

Advantages of helical CT

- **Faster** scan times
- **Less motion artefact**
- Retrospective slice **reconstruction**
- **3D** reconstruction
- **Less partial volume effect** (due to retrospective slice reconstruction)

Multislice CT

Overview

- Allows for **true 3D reconstruction** as obtained voxels are **isotropic** (w = d = h)
- Multislice scanners use multiple rows of solid state detectors
- Each row contains 600–900 detectors (as in a single slice scanner)
- Each row is narrow because the width of the detector determines the reconstructed slice width (as opposed to a single slice scanner where the collimated fan width determines slice width)
- Detector rows may be used **separately** or in **combination** to provide different slice widths
- 64 and 128 slice scanners are now commonplace

Multislice pitch

- Some controversy over definition
- Preferred (and most common) definition is **beam pitch**:

beam pitch = table movement per rotation / beam width

- An alternative definition is slice pitch:

slice pitch = table movement per rotation / slice width

The cone beam effect

- The principles behind multislice and single slice helical reconstruction techniques are the same:
  - Involve interpolation of measured data between adjacent data sets
- Standard reconstruction algorithms assume that the X-ray fan beam is non-

divergent in the Z-direction:
    - A reasonable assumption for single slice scanners
- For multislice scanners, the **X-ray beam is not a flat fan but a cone**:
    - This leads to artefacts and cone beam algorithms have been devised to minimise these

Image quality

CT contrast

- CT improves image contrast because it **separates structures** (no overlapping)
- kV affects contrast in CT less than in plain films due to heavy filtration
- Maximum **spatial resolution is 2 lp/mm**

Noise

- Reduces contrast resolution of small objects
- Reduces spatial resolution of low-contrast objects
- Three sources of noise in CT:
  1. **Quantum noise** (due to number of photons detected)
  2. **Structural noise** (due to the reconstruction technique used)
  3. Minimal **electronic noise** (due to the internal components of the scanner)

Quantum noise

- Remember, **noise = 1 / √N** (where N = number of photons)
- Can be **decreased** by increasing the number of photons detected:
  - **Increasing mAs** per rotation (increase mA or scan duration)
  - **Increasing slice width**:
    - Decreases image quality due to decreased spatial resolution and increased partial volume effect
  - **Increasing kV**:
    - Increases penetration
    - Would increase dose but mA can be lowered to compensate

        - Reduces contrast

Structural noise

- Increasing pixel size reduces noise:
    - The larger the voxel, the more photons detected and the lower the noise
- Reconstruction algorithm (kernel):
    - Remember that the kernel is the filter that is applied to each point of the measured data before back projection (resulting in filtered back projection)
- Smoothing kernels will decrease noise but reduce resolution

Pitch and noise

- **Single slice scanners**:
    - **Increasing pitch alone has no effect on noise**:
        - This is because the image is reconstructed from the same number of projections
        - Despite dose being reduced (which would normally increase noise due to quantum mottle)
        - Caveat is that slice width is increased so:
            - Increased partial volume effect
            - Reduced contrast of small objects
- **Multislice scanners**:
    - Algorithms only take projections from a defined "slab" of the patient of a fixed width
    - **Increasing pitch alone** decreases number of photons and **increases noise**
    - Multislice scanners, therefore, automatically increase mAs to keep noise constant

Image artefacts

Motion artefact

- Patient, respiratory and cardiac motion cause blurring
- The algorithm is misled by a moving structure occupying different voxels during the scan

Streak artefact

- **High attenuation objects** such as metal implants cause streak artefacts:
  - Streaks appear as **dark and light lines** emanating from the object
- Accentuated by motion

Photon starvation

- A variation of streak artefact
- A typical example would be the lateral projection of the chest through both shoulders:
  - The **X-ray attenuation through the increased tissue volume is high and may exceed the dynamic range of the processing unit**
- Modern scanners are capable of delivering more photons across this area (determined by the protocol and the topogram)

Beam hardening

- Also known as “cupping”
- Can lead to **dark streaks** in the image
- Remember that the reconstruction algorithm assumes a monoenergetic beam
- As the beam passes through the patient it is hardened (the low energy

components having been attenuated by earlier tissue):
    - Causes the attenuation coefficient (and therefore CT number) of a given tissue to decrease along the path of the beam
- The tissues in the centre of the patient are crossed by harder beams than those at the surface
- The result is that the **CT numbers are lower in the centre of the patient than they should be**
- Can be reduced by:
    - A **bow tie filter**
    - Beam-hardening algorithms

Ring artefact

- Caused by failure of a single detector
- Produces a light or dark ring in the image depending on whether the detector is giving a high or low signal

Partial volume effect

- The CT number stored in each voxel represents the average attenuation coefficient in that voxel
- A high contrast object that is smaller than the voxel may therefore be seen even if it's dimensions in the transaxial plane are less than the pixel size
- This means that a high contrast structure that crosses the transaxial plane can be visualised in several adjacent pixels
- **Results in high contrast structures appearing larger than they really are**
- **Reduces the visibility of low contrast structures**
- Effect can be reduced by decreasing slice width

CT fluoroscopy

Overview

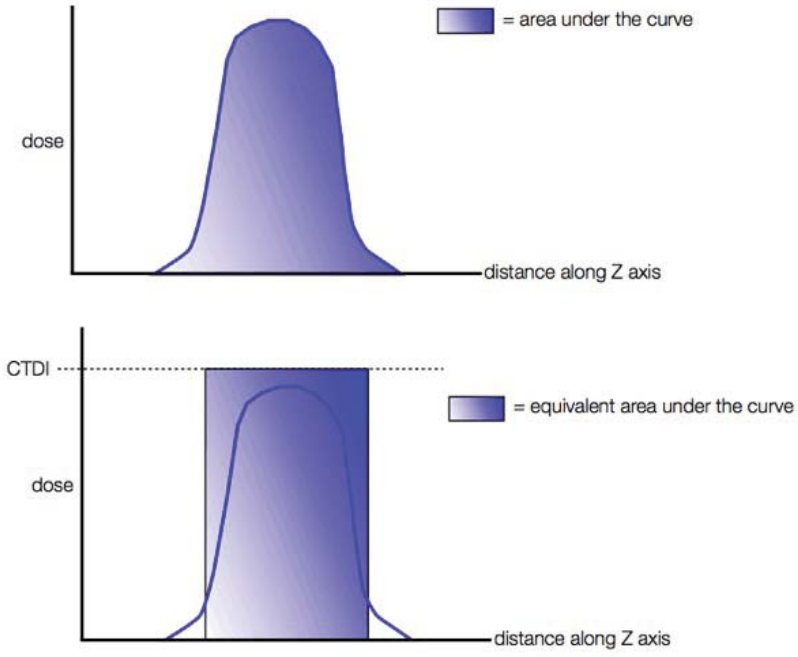
- CT fluoroscopy is the display of a CT image in real time
- Gantry continuously rotates whilst the couch remains stationary
- Uses a **180° rotation dataset**
- **5 frames / sec** can be achieved
- Generally a lower effective dose than a diagnostic scan but **high skin doses** can occur (as scanning is reduced to a narrow region of the body)

Measuring CT dose

CT Dose Index (CTDI)

- A measure of **dose from a single rotation** (slice)
- Unit is **mGy**
- Best explained diagrammatically

Dose from a single rotation (dose profile curve)



Dose profile curves

- As you can see from the diagrams above, CTDI is the height of a rectangle with the same area under the curve as the single slice dose profile curve
- The **peak dose is independent of slice width**, depending only on hardware factors:
  - Filtration

    - Collimation
    - Focal spot size
    - kV and mAs
- CTDI is therefore scanner-dependent

Weighted CTDI ($CTDI_W$)

- For patient dosimetry purposes, $CTDI_W$ is often quoted
- This takes into account the **spatial distribution** of dose within the patient in the scan plane:
    - Remember beam hardening. The dose is greatest at the surface and less in the centre
- Calculated using perspex phantoms and an ionisation chamber
- Unit is mGy
- **Typical $CTDI_W$** would be **10–20 mGy (head)** and **5–10 mGy (body)**

Volume CTDI ($CTDI_{vol}$)

- This is simply the **CTDI divided by the pitch**

$$\mathbf{CTDI_{vol} = CTDI / pitch}$$

- Increasing pitch lowers dose (as we'd expect)
- Measured in **mGy**
- Is an approximation of the **average absorbed dose within the scanned volume**

Dose length product (DLP)

$$\mathbf{DLP = CTDI_{vol} \times irradiated\ scan\ length}$$

- More useful than CTDI for comparing doses to individual patients
- DLP **unit is mGy/cm**
- Allows us to estimate effective dose by multiplying by a fiddle factor (EDLP):

Effective dose = DLP x EDLP

- Effective dose unit is mSv
- The fiddle factor varies depending on the region of the body scanned

Typical doses for CT

	$CTDI_w$ (mGy)	DLP (mGy/cm)	Effective dose mSv
Head	60	700	1.5
Chest	14	400	7
Abdo / pelvis	16	500	8.5

Factors affecting CT dose

In the below examples, we are altering just a single parameter at a time. In reality, other parameters may need to be altered as well to maintain adequate image quality.

Adjustable factors affecting patient dose

- kV
- mAs (mA combined with rotation time)
- Pitch

kV

$$\text{tube output } \alpha\ kV^2$$

- We can see that radiation output increases massively with kV
- Increasing kV increases dose but because more photons penetrate the patient **we can usually decrease mA** to give the same level of noise:
  - In reality, therefore, increasing kV leads to a decrease in dose
  - **If we do not alter mA though, increasing kV would increase dose**

mAs

- **Increasing mAs increases dose** but it decreases noise

$$\textbf{dose } \alpha \textbf{ mAs noise } \alpha \textbf{ } 1 / \sqrt{\textbf{mAs}}$$

- Scanners use mAs modulation to keep a constant detector signal so that changes in patient size do not significantly affect the level of noise:
  - Dose savings of 10–40% compared to using a constant mAs
  - Different manufacturers use different names, e.g Siemens is called CARE Dose

Pitch

- **Doubling the pitch halves the dose**

$$\text{dose} \propto 1 / \text{pitch}$$

Gated CT imaging

Overview

- Allows imaging of the heart by using the ECG to time projections to reduce motion artefact
- There are two main types:
  - Retrospective gating
  - Prospective gating

Retrospective cardiac gating

- Patient is hooked up to an ECG machine
- Tube current is on continuously during helical acquisition
- The ECG data is used to select projections to be used for the image reconstruction
- Doses can be very high

Prospective cardiac gating

- Patient is, again, hooked up to an ECG machine
- ECG data is used to turn mA on only when the heart is relatively still
- Requires a stable, predictable, heart rate:
  - Patients are usually **beta-blocked**
- **Significantly lower dose than retrospective gating**

Chapter 10

Ultrasound

Basic principles

Overview

- Ultrasound refers to sound waves whose frequency is too high to be heard by the human ear:
  - Audible sound: 20 Hz–20 KHz
  - Ultrasound: > 20 KHz
  - **Diagnostic medical ultrasound: 2–20 MHz**

The properties of sound waves

- Sound is a **mechanical**, **longitudinal**, **pressure wave**
- The speed of sound varies depending on the medium it is propagating through:
  - **Air**: **330 m/s**
  - **Water**: **1480 m/s**
  - **Soft tissue**: **1540 m/s** (an approximate average)
- Sound waves can be reflected, refracted and focused
- The speed of sound is affected by the density and stiffness of the material:
  - **High speed of sound = high stiffness and low density**
  - Low speed of sound = low stiffness and high density
- Thus, **speed is proportional to stiffness over density**:

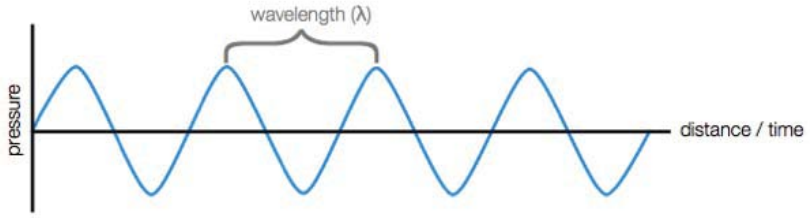
$$\mathbf{c = \sqrt{(k / \rho)}}$$

(where c = speed, k = stiffness, ρ = density

- And **wavelength is proportional to speed** (the wave equation)

$$\mathbf{\lambda = c / f}$$

(where c = speed, f = frequency, λ = wavelength)



Attenuation

- When travelling through a material, sound is attenuated exponentially for 2 reasons:
  1. Energy is **absorbed** and converted to heat
  2. Energy leaves the forwards-travelling beam due to **scatter** and **partial reflection**:
     - the **higher the frequency, the higher the attenuation**
- Attenuation is measured in decibels (dB)
- In soft tissue, **sound loses 1 dB / cm for every 1 MHz**
- In air: 40 dB / cm at 1 MHz
- Since there is little absorption or scatter in water, a full bladder can aid penetration
- As a rule of thumb: **penetration (cm) = 40 / frequency (MHz)**

The piezoelectric effect

- Certain crystalline materials demonstrate the piezoelectric effect:
  - Undergo **compression & expansion when subjected to an electrical current**
  - Produce an **electrical current when subjected to vibrations**
  - Act as both **transmitters and receivers** of sound waves
  - When heated above their curie temperature, they lose their piezoelectric properties:
    - This is why ultrasound transducers should not be autoclaved

Scatter

- Occurs when sound encounters a structure that is much smaller than its wavelength
- The wave is scattered more or less equally in all directions

Beam behaviour at material interfaces

Overview

- Two things happen when an ultrasound beam strikes the boundary between two materials at a right angle:
  1. Some energy is reflected (the **echo**)
  2. The remainder of the energy is **transmitted**

Acoustic impedance (Z)

- The proportions of energy reflected and transmitted depend on the acoustic impedances of the two materials
- Depends on **density** and the **speed of sound**

$$\mathbf{Z = c\rho}$$

(where Z = acoustic impedance, c = speed of sound in the tissue, ρ = density)

- **Independent of frequency**
- It's the difference in Z between the two materials that's important:
  - **Large difference in Z = more energy reflected**
  - **Small difference in Z = more energy transmitted**
  - No difference in Z = 100% transmission

Examples of acoustic impedance (Z)

Tissue	Z
Air	0.0004
Lung	0.18
Soft tissue (fat, water, muscle)	1.3 - 1.7
Bone	7.8

Gaseous interfaces

- Since Z is negligible for air or gas, **total reflection occurs** with the following consequences:
  - Gas-filled organs cast a shadow and structures underneath cannot be imaged:
    - Normal lung cannot be penetrated
    - The bowel wall can be seen but not the lumen
  - It is impossible to get sound from the transducer to the patient if there is air trapped between:
    - This is why we use coupling gel

Typical reflection values

Interface	% energy reflected
Gas / tissue	99.9
Tissue / transducer	80
Bone / muscle	30
Fat / muscle	1
Blood / muscle	0.1
Liver / muscle	0.01

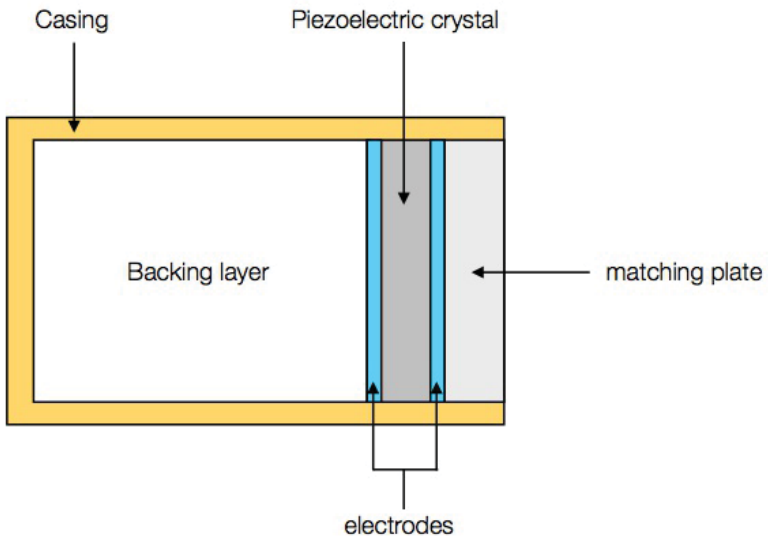
Ultrasound transducers

Probes

- There are several types of probes used in diagnostic ultrasonography
- A probe may contain one or many transducers:
  - Single transducer (e.g. a continuous wave pencil probe)
  - Array of transducer elements:
    - Linear
    - Curvilinear
    - Phased
    - Annular

Components of a transducer

1. A piezoelectric crystal sandwiched between two electrodes
2. A backing layer:
   - Matched to the transducer so as to permit sound waves to travel backwards and be scattered and absorbed within the probe itself without reflection
3. A matching plate:
   - Is a quarter of a wavelength thick
   - Made from plastic with an acoustic impedance (Z) between the transducer and skin

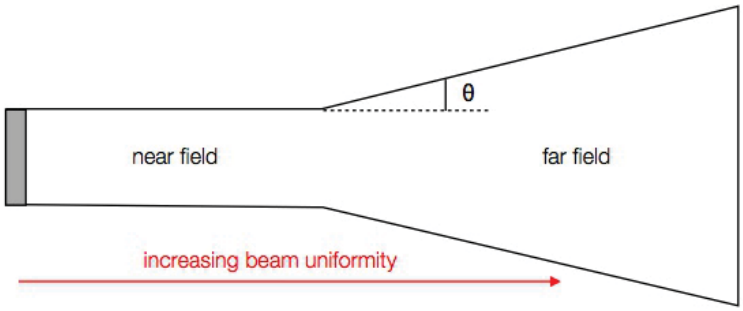

Casing
Piezoelectric crystal
Backing layer
matching plate
electrodes

Beam shape

Overview

- If the transducer diameter < wavelength (λ):
  - Waves would be emitted in all directions
  - No overall beam direction
- If **transducer diameter > λ** we get a **near and a far field**

Near and far fields



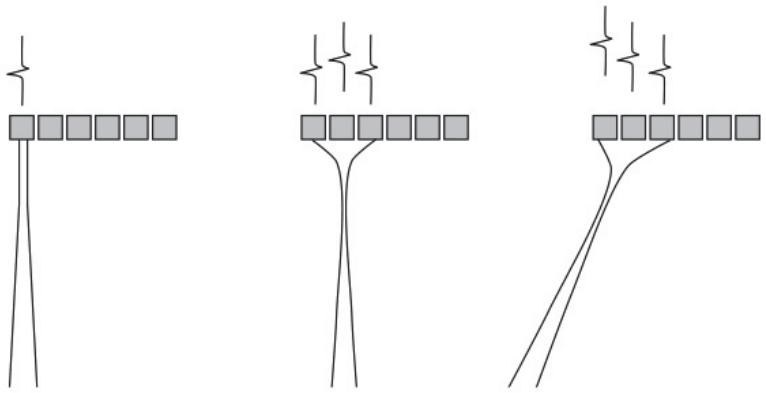
- The proximal, almost parallel, part of the beam is called the **near field** (or Fresnel region)
- The distal, divergent, part of the beam is the **far field** (or Fraunhofer region)
- The area just beyond the near field/far field interface is the area of **maximum beam uniformity**
- As **frequency increases**:
  - Near field length increases
  - Angle of divergence (θ) decreases
- As **transducer diameter increases**:
  - Near field length increases
  - Angle of divergence decreases

Focusing

- Transducers are able to focus the ultrasound energy to a particular depth by increasing beam intensity at that point
- **Produces stronger echoes** from structures at that depth
- **Improves lateral resolution**
- There are three ways to focus a beam:
  1. Use a curved piezoelectric element
  2. Cement a plastic acoustic lens to the face of the transducer
  3. Electronic focusing

Electronic beam formation

- It's possible to control the direction of the ultrasound beam (beam steering) and its focus by varying the number and the timings of excitation of different transducer elements



A-mode and B-mode imaging

A-mode imaging

- Stands for **Amplitude mode** imaging
- Simplest form of ultrasound imaging
- Generates a **graph of echoes plotted as a function of depth**
- Largely superseded by B-mode imaging

B-mode imaging

- Stands for **Brightness mode** imaging
- Images a *slice* of the patient
- Generally we have an image matrix of 512 x 512 pixels
- Mechanism:
  1. The ultrasound beam scans back and forth across a 2D section of the patient in a linear or sector pattern
  2. Each pulse gives rise to a train of echoes
  3. Amplitude is converted to grey scale value
  4. Time taken to receive the echo correlates to depth (row in the image matrix)
  5. Grey scale value is written to the matrix
  6. For any pixel not covered by a scan line, a grey scale value is **interpolated** from adjacent pixels
  7. The sequence is repeated at a sufficiently high frame rate to provide a real time image

Dynamic range

- Every component in an ultrasound system has a dynamic range

- It's the ratio of the maximum intensity of the signal to the minimum that can be detected
- This ratio is generally 70–80 dB
- After time gain compensation, the ratio is about 40–50 dB
- The monitor only has a grey scale range of about 25 dB:
  - Within which the eye can only distinguish about **30 levels of grey**
- The system therefore compresses this 40–50 dB range down to a 20–30 dB range:
  - We can choose to enhance low, medium or high level signals as we desire

Ultrasound pulses and the range equation

- Each ultrasound pulse (one scan line) is approximately:
  - **2–5 cycles in duration**
  - **1mm long**
  - **3mm wide**
  - Occupies a volume the shape of an onion
- The **range equation**:
  - We can calculate the distance (d) of a reflecting surface if we know:
    - The time taken between transmission of the pulse and reception of the echo (t)
    - Speed of the ultrasound beam (c, assumed to be 1540 m/s in soft tissue)

$$\mathbf{d = (t * c) / 2}$$

(divide by 2 as t is the time to go back *and* forth)

Frame rate

- This is the number of whole images (frames) that are displayed per second (measured in Hz)

- The image will appear smooth if the frame rate is > ~20 Hz
- To calculate the frame rate, we calculate the time taken to record one scan line in ms (using the range equation), multiply that by the number of scan lines and then divide that value into a second (1000ms)
- **Frame rate decreases**:
  - As **depth increases**
  - As **number of scan lines increases**
  - If colour flow **doppler** is used
- Since we are limited by the number of scan lines we can use in order to produce a real time image, modern scanners introduce additional computed lines in between the measured lines in a process called **interpolation**

Time gain compensation

- The **amplitude of the sound pulse decreases the further it travels** into the body due to attenuation by tissue
- Similarly, the echo is attenuated as it travels back towards the transducer
- This means that a reflector deep in the body produces a weaker echo than an identical reflector closer to the surface
- We compensate for this attenuation electronically with time gain compensation:
  - As soon as the transducer is pulsed, **the gain of the amplifier is steadily increased in proportion to the time elapsed** and, thus, the distance travelled by the sound
- Can be varied by the operator

Spatial resolution

Overview

- Resolution is the ability to identify that two structures close together are separate
- Three dimensions:
  1. Axial
  2. Lateral
  3. Slice thickness

Axial resolution

- Also known as **depth resolution**
- The minimum distance between two reflectors along the axis of the beam that can result in two distinguishable echoes
- The axial resolution is about half the pulse length
- **High frequency = high axial resolution**

Lateral resolution

- The ability to **differentiate two reflectors side by side at the same depth in the same scan plane**
- For high lateral resolution:
  - **Focus the beam**
  - **Increase number of scan lines**

Slice thickness resolution

- The ability to **differentiate two reflectors at the same depth in different scan**

planes

- Dictates the thickness of the section of tissue that contributes to echoes on the image
- Increased by **focusing** the beam

The doppler effect

Overview

- The doppler effect is the **change in frequency of a wave for an observer moving relative to the source of the wave**
- For sound waves, the effect can result from movement of either the source, the observer or the medium
- As the wave source moves towards the observer:
  - Each successive wave crest is emitted from a position closer to the observer than the previous wave
  - Thus, each successive wave takes less time to the reach the observer than the previous
  - This results in an increase in frequency
- Conversely, as the wave source moves away from the observer:
  - Each successive wave crest is emitted from further away from the observer
  - Thus, each successive wave takes longer to reach the observer and the frequency decreases

The doppler frequency

- Regarding ultrasound, this is the **difference between the transmitted and received frequency**
- Depends on 4 factors:
  1. **Emitted frequency**
  2. **The speed of sound**
  3. **The speed of the object** struck by the wave
  4. The angle the wave strikes the object (**angle of insonation**, Cos θ)
- Doppler ultrasound provides us with a means of assessing speed and direction of

movement

- **Angle of insonation = 90 degrees, doppler frequency = zero**

$$F_{dop} = \frac{2 \times F_{transducer} \times V_{object} \times \cos\theta}{c}$$

(where F_{dop} = doppler frequency, $F_{transducer}$ = transducer frequency, V_{object} = object velocity, θ = angle of insonation, c = speed of sound)

Doppler ultrasound

Continuous wave doppler ultrasound

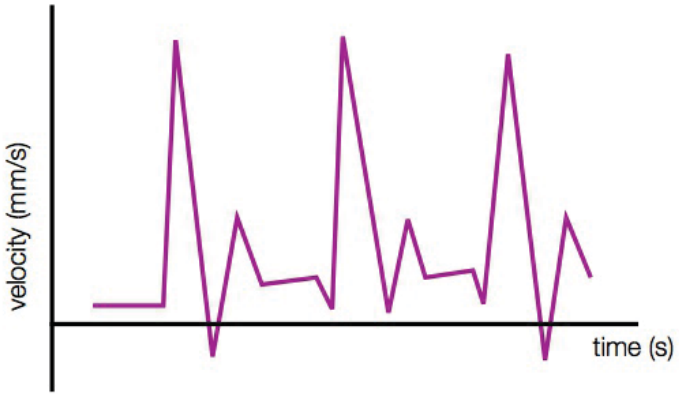
- **Blood flow velocity is measured by the doppler shift of ultrasound back-scattered by blood cells**
- The probes uses two differently angled transducers (one to transmit and one to receive)
- Transmitted frequency range is 2–10 MHz (dependent on vessel depth)
- Mechanism:
  1. Transmitter transmits continuously at frequency F
  2. Receiver receives continuously waves of frequency F*
  3. The original frequency (F) is suppressed
  4. We extract the doppler signal (F - F*) electronically
- For an 8 MHz transmitted pulse, the doppler shift is approximately:
  - **5 KHz for arteries**
  - **1 KHz for veins**
  - Both these frequencies happen to be within the audible range
- The **higher the frequency** of the doppler shift, the **higher the velocity**
- With continuous wave doppler, it is not possible to locate the moving reflector or to distinguish between flow in two overlapping vessels at different depths in the beam

Pulsed doppler

- Provides a spectral display of blood velocity from a region of interest
- Doppler measurement is combined with a real time B-mode image (**duplex imaging**)
- Mechanism:

1. From a B-mode image, a line of sight and sampling volume is chosen for the doppler beam
2. The normal short B-mode imaging pulses are interspersed with bursts of doppler ultrasound (approximately 10 cycles)
3. A range gate is set to accept only those echoes that arrive within a short time interval at a specific time such that they could only have arisen from the selected sampling volume

- The spectral graph produced displays the velocity against time:



- The area under the curve is filled by pixels:
  - Each pixel has a grey level that represents the number of blood cells in the sampling volume that have that velocity at that moment in time
- From the spectral graph, we can calculate many things such as peak and mean velocity

Colour flow imaging (CFI)

- Also known as colour doppler imaging
- We know that a B-mode image shows the **strength** of an echo coming from each pixel
- A colour flow doppler mapped image shows the direction and velocity of movement occurring in each pixel by means of an arbitrary colour code:
  - Flow towards the transducer = red

- Flow away from the transducer = blue
- Turbulence (variations in flow direction) = green / yellow

Power doppler

- **Used to provide information on the amplitude** of the returning doppler signal
- Ignores the velocity
- The amplitude of the signal **depends on the number of red blood cells** within the imaged volume
- All movement, regardless of phase, contributes to the amplitude:
  - Thus power doppler **emphasizes the quantity of blood flow**
- **Increased sensitivity compared to colour doppler**
- Signal is less dependent on insonation angle
- Not subject to aliasing (as power doppler is non-directional)

Harmonic imaging

Overview

- As the ultrasound propagates through the tissue, its **wave becomes distorted** because the speed of sound varies with tissue type
- The sound wave **ceases to be a perfect sinusoidal wave**
- The wave distortion becomes more pronounced as depth increases and reduces image quality
- The distorted wave is made up of several **harmonic frequencies**
- This **distortion only occurs in the central, high amplitude, part of the beam** (not the weak side lobes)

Harmonic frequency

- The fundamental frequency (transducer frequency) is termed the **first harmonic**
- Subsequent harmonics are **integral multiples** (i.e. whole number multiples) of the first harmonic
- E.g:
  - First harmonic = 2 MHz
  - Second harmonic = 4 MHz
  - Third harmonic = 6 MHz, etc
- There are two ways to detect tissue harmonics:
  1. Harmonic band filtering
  2. Pulse inversion

Harmonic band filtering

- Harmonics are produced after a few centimetres of penetration

- Harmonics exist in the returning echoes but are not produced within it:
  - Unless the returning echo is from a contrast agent
- We **remove the the fundamental harmonic from the returning echo using a filter**:
  - This leaves an echo composed of tissue-generated harmonics that we can process
- We need to ensure that the transmitted pulse does not contain any high frequencies that may be present in the harmonics as this would corrupt the signal:
  - Use a **narrow bandwidth**
  - Use a **long pulse duration**
- Using a long pulse duration reduces axial resolution:
  - Somewhat negated by the fact that the image quality is better

Pulse inversion

1. **Invert** (i.e. 180° out of phase) **every other transmitted pulse**
2. **Sum the echoes received** from each pair of transmitted pulses:
   - All odd harmonics (including the first harmonic) are removed
   - The amplitude of all even harmonics (including the second harmonic that we're interested in) are doubled:
     - Enhances the signal to noise ratio
   - Advantages:
     - No filtering is used so broad bandwidth and short pulse length can be used:
       - **Better axial resolution**
   - Disadvantages:
     - Multiple pulses used so takes longer:
       - Movement artefact
       - **Lower frame rate**

Advantages of harmonic imaging

- **Less noise**:
  - Weak signals do not produce harmonics as they are not subject to non-linear propagation
  - Better visualisation of low contrast lesions and liquid-filled cavities
- **Improved lateral resolution**:
  - The effective width of the harmonic beam is narrower than the main beam

M-mode imaging

Overview

- Also known as **motion mode** imaging
- Primarily used for **cardiac** imaging
- The heart valves and heart wall move too quickly to be followed with a normal real time scanner
- A B-mode image is frozen on the screen and is used to direct the beam from a stationary transducer along a line of interest:
  - The beam should intersect the moving structure as close to a right angle as possible
- Echoes are displayed on the screen as a line of moving bright dots
- Quantitative analysis is possible
- Often combined with 2D colour doppler

Artefacts

Overview

Image formation assumes that sound travels a straight line, with a constant velocity and attenuation and is reflected only once from each surface. None of these assumptions are correct and as a result, artefacts occur.

Speckle

- Caused by interference from **many small structures** (too small to be resolved) within tissue
- Produces a **textured appearance**
- Echo pattern is random and unrelated to the actual pattern of scatterers within the tissue

Reverberation

- Caused by multiple reflections to and fro between the transducer face and a relatively **strong reflector near the surface**
- Produces a **series of delayed echoes** equally spaced in time that falsely appear as distant structures

Double reflection

- E.g. the diaphragm acts like a **mirror** and structures in the liver can appear to lie within the lung

Acoustic shadowing

- **Strongly attenuating** or reflecting structures **reduce the intensity of echoes** from the region behind them and cast shadows

- E.g: bowel gas, lung, bone and gallstones
- Made worse by time gain compensation

Acoustic enhancement

- **Fluid-filled structures** (weakly attenuating) **increase the intensity of echoes** from the region behind them
- E.g: cysts and full bladder
- Made worse by time gain compensation

Refraction

- Distortion of the image due to a **markedly different velocity through a tissue** than the assumed 1540 m/s
- E.g: high velocity in gallstones and low velocity in the lung

Aliasing

- Caused by **under-sampling**
- Refer to earlier article on nyquist frequency
- Affects colour doppler imaging leading to wrap around:
  - Means that flow towards the transducer is represented as flow away from the transducer and vice-versa
- If you have a scale from 0 - 5 and the true value is 6 how do you represent this? The scanner will represent this as 1 (6 wraps around to the beginning of our scale)

Contrast agents

Overview

- Act by **increasing the reflections** from the tissue containing the agent
- Generally based on **microbubbles** (**< 4µm**) or **nanoparticles** (**< 1µm**)
- Bubble diameters are much less than the pulse wavelength but they can **resonate** at the ultrasound frequency
- Are destroyed:
  - By very high frequency ultrasound:
    - This property is exploited in reperfusion imaging
  - Within a few hours by the body

Types

- Air-filled microspheres encapsulated in an albumin shell:
  - Increase back scatter from ventricular borders to improve visualisation and flow evaluation
  - Also adhere to thrombi and aid in the diagnosis of DVT
- Low solubility gas encapsulated in a lipid shell:
  - Used for general vascular work
- Perfluorocarbon nanoparticles:
  - Long half life
  - Slowly taken up by the liver and spleen
  - Good for imaging of metastases
- Gold-bound colloidal microtubes:
  - Can be immunologically targeted by combining with antibodies

Ultrasound safety

Overview

- Ultrasound is non-ionising
- There is **no solid evidence that ultrasound is harmful at diagnostic energies**
- **Intensity** of the beam is greatest in the focal region:
  - **Averages 0.1 mW/mm^2**
  - Peak is 0.1 W/mm^2 (during the brief pulse)
- There are agreed safety guidelines in ultrasonography:
  - Time averaged intensity should never exceed **100 mW/cm^2**
  - Total sound energy (intensity x dwell time) should never exceed **50 J/cm^2**

Potential harmful effects

- **Local heating** (this can be used therapeutically)
- **Acoustic streaming** of cellular contents in the beam direction:
  - Takes a few seconds to occur
  - Can be diagnostic (seen in the bladder and cysts)
- **Cavitation**:
  - The process whereby small bubbles in a liquid are forced to oscillate in the presence of an ultrasound beam
  - When oscillating, the bubbles expand and contract and alter the flow of liquid in their immediate vicinity:
    - Possibility here to affect cell function
  - A bubble can also burst whilst oscillating:
    - Releases lots of energy in a small space that could, potentially, be damaging

Mechanical index (MI)

$$\mathbf{MI = p / \sqrt{f}}$$

(where MI = mechanical index, p = peak negative pressure in MPa, f = frequency in MHz)

- Displayed on all scanners
- Estimates the degree of bio-effects a given set of ultrasound parameters will produce
- The higher the MI, the greater the degree of bio-effects
- The FDA stipulates that **scanners cannot exceed a MI of 1.9**
- There is a potential hazard with gas contrast agents if MI > 0.7

Thermal index (TI)

- The TI **estimates the temperature rise in the patient**
- There are **three indices** depending on the region being scanned:
  - Soft tissue thermal index (TI_s)
  - Bone thermal index (TI_b) for scanning through soft tissue into bone
  - Cranial thermal index (TI_c) for scanning through bone into soft tissue
- Thermal indices are pre-calculated by the manufacturer and so are a best guess only
- Rule of thumb:
  - The "real" temperature rise is ~ 2 x TI but guidelines are based on TI alone
- FDA states **TI < 6** (except **ophthalmology** where it's **< 1**)
- BMUS states **TI < 3 for fetal scanning**

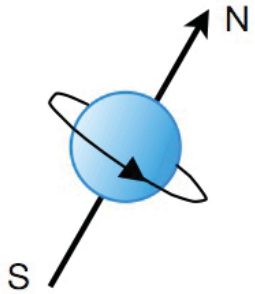
Chapter 11

Magnetic resonance imaging

Protons and their magnetic fields

Spin

- Every proton has a single positive charge
- Protons spin about a central axis (much like the Earth)
- This spinning charge is essentially a small current and, therefore, induces a magnetic field
- Since **protons have their own magnetic field**, we can think of them as little bar magnets with a north and a south pole
- Confusingly, protons are referred to as spins in many text books



Proton alignment

- Normally, the alignment of our proton magnets is random
- When placed inside a strong external magnetic field (B_0), the proton's magnetic field lines up in one of two orientations:
  - **Parallel** with the direction of B_0 (**spin-up**)
  - **Anti-parallel** to B_0 (**spin-down**)
- **Slightly more protons align spin-up** (because it requires less energy to do so)
- How many more protons align spin-up **depends on the strength of B_0**:
  - **4 per million in a 1.5T magnetic field**

Precession

- Within a strong external magnetic field, protons not only line up but they **rotate elliptically** like a spinning top or a Weeble:
  - Known as precession
- How many times a proton precesses per second is known as the **Larmor frequency**:
  - This **increases as B_0 increases**

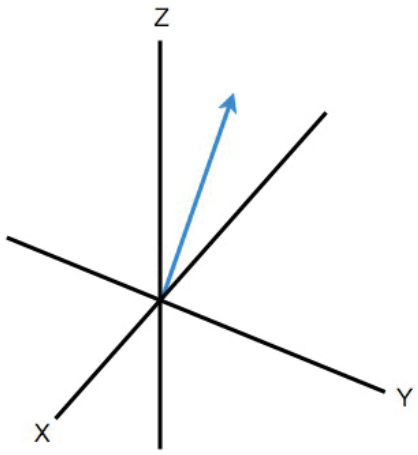
$$\omega_0 = \gamma * B_0$$

(ω_0 = Larmor frequency in MHz, γ = gyromagnetic ratio and B_0 = magnetic field strength in T)

- Gyromagnetic ratio (γ):
  - Material specific (42.5 MHz / T for protons)
- Therefore, **if B_0 = 1T then the Larmor frequency is 42.5 MHz**

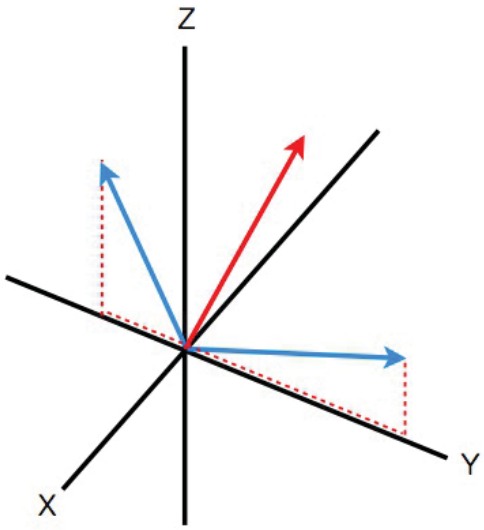
Vectors

- A vector has both size and direction and is represented graphically by an arrow
- A vector's size is represented by it's length
- In MRI physics, the size represents the magnetic field of a proton
- Each proton's magnetic field can be represented by a vector in a 3D coordinate system where:
  - Z = direction of B_0 = **longitudinal component**
  - X and Y = **transverse component**

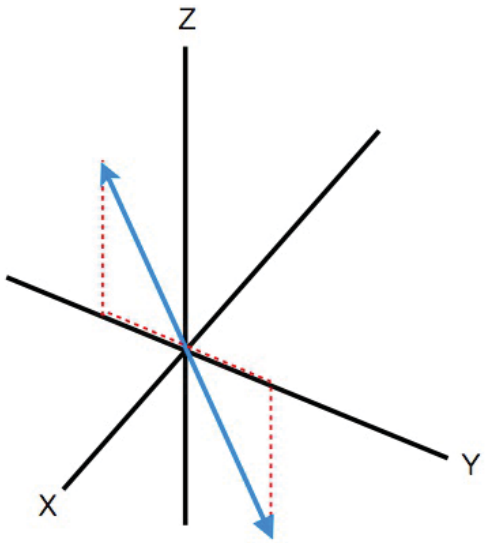


Vector summing and cancellation

- We add vectors together by adding their Z component and then their transverse component
- If two vectors are equal in size but opposite in direction, they **cancel each other out**
- Remember that there is a slight preponderance for protons to align spin-up compared to spin-down when placed inside B_0:
  - This means that **most of the proton magnetic fields will cancel each other out**
  - There will be a few spin-up protons which are not cancelled out:
    - This will be in the Z (longitudinal) direction and gives us a **net longitudinal magnetization**
- We are **unable to measure** this net magnetization because it is very tiny and in the same direction as B_0 (which is very large)
- In order to measure this net magnetization (which will give us our signal) **we need to "flip" it into the transverse direction**



The red vector is the result of adding the 2 blue ones. We add the Y components (a small negative and a larger positive value) to get an overall positive Y component. The Z components are both positive so we end up with a larger longitudinal component than we started with.

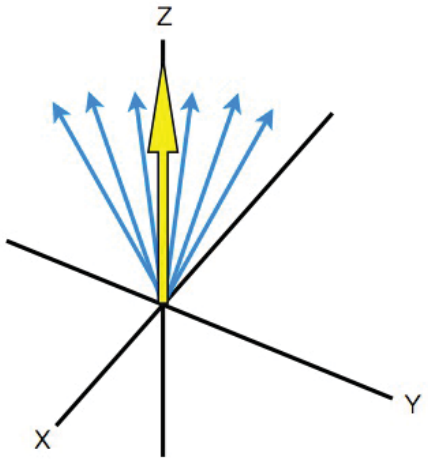


These 2 vectors cancel each other out. They have equal but opposite longitudinal (Z) and transverse (Y) components.

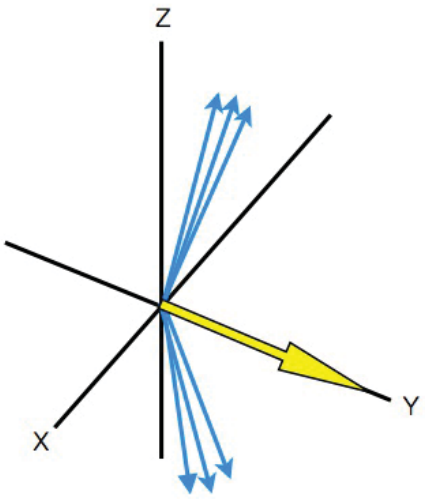
The radiofrequency (RF) pulse

Overview

- We said earlier that we are unable to measure the net magnetisation of the patient when it's in the longitudinal direction
- In order to measure the net magnetisation, we need to flip it 90° into the transverse direction
- If we send in an **RF pulse at the Larmor frequency** the protons pick up some energy:
  - Known as **resonance**
  - Some of the spin-up protons will become spin-down:
    - Has the net effect of **reducing the net longitudinal magnetisation**
  - All of the photons will be **in phase**:
    - Gives us our transversal magnetisation



Here we have 6 spin-up protons. Their transverse (X, Y) components cancel each other out giving us no transversal magnetisation. They do, however, posses a net longitudinal magnetisation (yellow arrow) which we can't measure.



The RF pulse has given 3 protons enough energy to become spin-down. The longitudinal components of the vectors now cancel each other out leaving us with **no** longitudinal magnetisation. However, we do have a net transversal magnetisation as the protons are now in phase (yellow arrow).

After the RF pulse

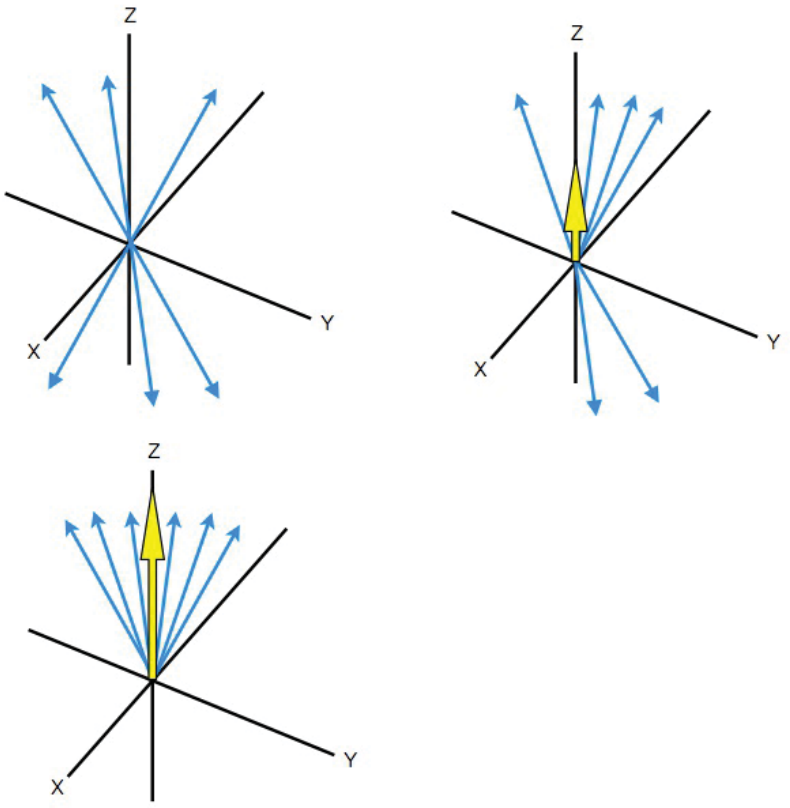
- As soon as the RF pulse is switched off, the protons start doing two things simultaneously:
  1. Losing energy and **returning to spin-up**:
     - Longitudinal magnetisation increases (longitudinal relaxation)
  2. **Dephasing**:
     - Transversal magnetisation decreases (transversal relaxation)

T1 & T2

Longitudinal relaxation time (T1)

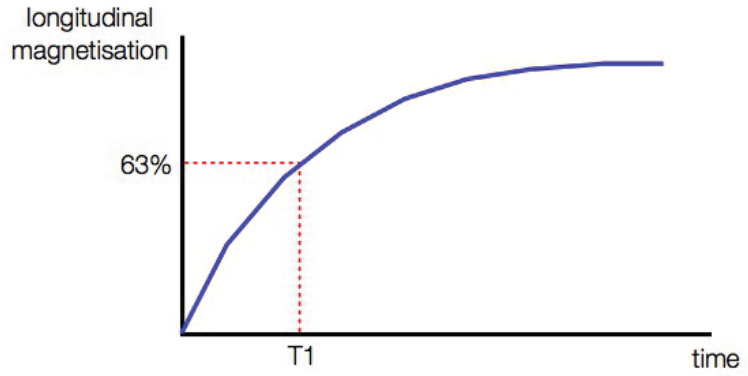
- The longitudinal relaxation time is the **time taken for all of the protons that became spin-down to become spin-up again**
- It's a continous, gradual, process
- It's a reflection of how easy it is for the protons to transfer the energy they gained from the RF pulse to their surroundings (also known as the lattice)
- Longitudinal relaxation time is known as T1 or **spin-lattice relaxation**

Diagram illustrating the gradual recovery of the longitudinal magnetisation (yellow arrow)



The T1 curve

- **T1 is the time taken for the longitudinal magnetisation to recover to 63% of its maximum value**
- After three T1 intervals, recovery is 95%



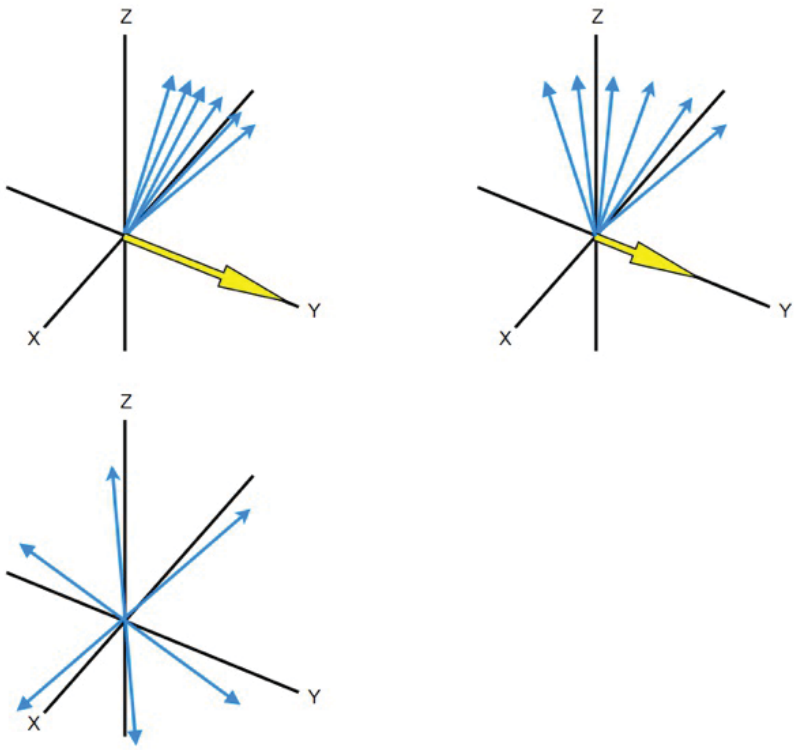
What does T1 depend on?

- T1 depends on the surrounding tissue composition and structure
- The **shorter the T1**, the **quicker the protons exchange thermal energy** with the lattice
- **Liquids** have a **long T1**:
  - Difficult to hand over thermal energy as the surrounding molecules are moving too rapidly
- **Fat** has a **short T1**:
  - The carbon bonds at the ends of fatty acids have frequencies near the Larmor frequency:
    - Thus energy transfer is easier
- **T1 increases as** the strength of the external magnetic field (**B_0**) **increases**:
  - The protons precess faster (Larmor frequency increases)
  - Faster moving protons are less efficient at transferring energy to the slower moving lattice

Transversal relaxation time (T2)

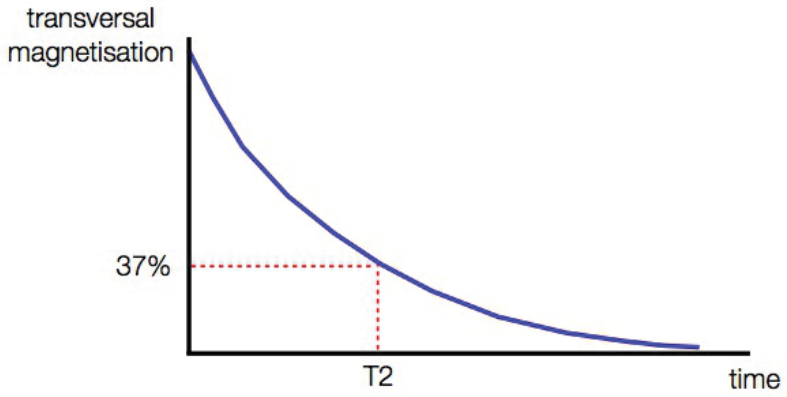
- The transversal relaxation time is the **time taken for the protons to completely dephase**
- It is a continous, gradual, process
- Protons precess at the same rate in an identical magnetic field:
  1. B_0 is not identical throughout (there are tiny field inhomogeneities)
  2. Each proton is affected by the tiny local magnetic fields from nearby nuclei:
     - These are not distributed equally
     - Are characteristic of a tissue
- Transversal relaxation time is known as T2 or **spin-spin relaxation**

Diagram illustrating the gradual loss of the transversal magnetisation (yellow arrow)



The T2 curve

- **T2 is the time taken for the transversal magnetisation to decay to 37% of its maximum value**



What does T2 depend on?

- **Dephasing is caused by inhomogeneities in**:
  - The **external magnetic field** (B_0)
  - **Local magnetic fields** in the surrounding tissue
- The **shorter** the **T2**, the **more inhomogeneous** the local magnetic field is
- **Liquids**:
  - The molecules within liquids move relatively quickly
  - This means the local magnetic fields of those molecules also move quickly
  - The magnetic fields "average out" to give a relatively homogeneous magnetic field
  - Protons, therefore, stay in phase for longer
  - **T2 is long**
- **Impure liquids**:
  - Large molecules move relatively slowly
  - This means that the local magnetic field is more inhomogeneous

- Protons, therefore, dephase more quickly
- **T2 is short**

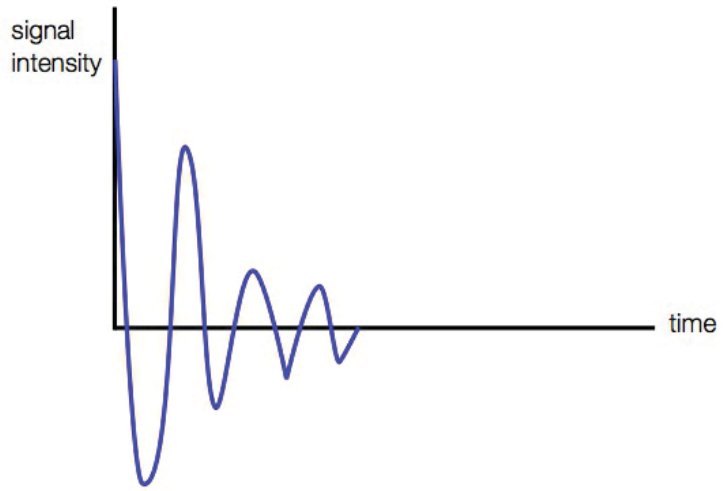
Free induction decay

Overview

- We know that applying a 90° pulse flips the longitudinal magnetisation into the transverse plane
- When the 90° pulse is switched off, two things happen (simultaneously):
  1. Transverse magnetisation decays to zero
  2. Longitudinal magnetisation recovers to it's original value before the 90° pulse was applied
- Since the net magnetic field is changing, an electrical charge is induced in our receiving equipment:
  - This is our raw MR signal
- The **frequency** of this raw signal **remains constant** (as the net magnetic vector is rotating at the Larmor frequency) but its **intensity decreases with time**

Free induction decay

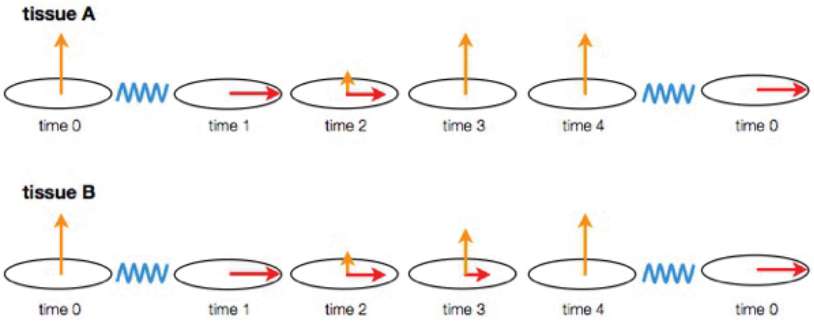
- This raw signal is known as the **free induction decay (FID) signal**



T1 weighting

T1 weighting

- A T1 weighted image is an MR image in which the T1 (or spin-lattice) relaxation times of the tissues provide most of the contrast in the image



Two tissues start with maximum longitudinal magnetisation (MZ, orange arrow). A 90° RF pulse flips MZ into the transverse plane (MXY, red arrow).

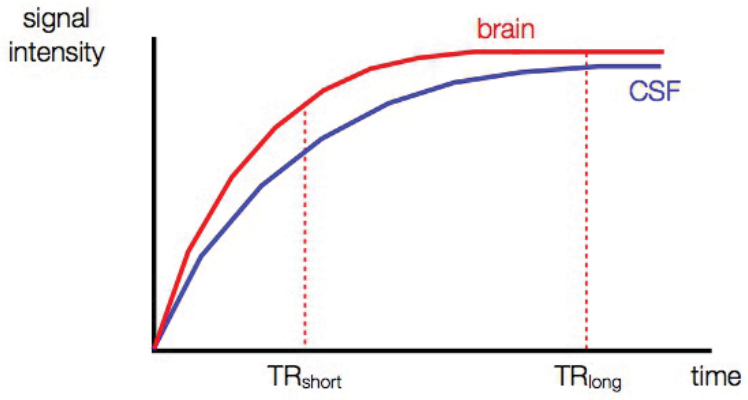
At time 3, tissue A has fully recovered MZ whereas tissue B does not fully recover MZ until time 4.

If we send a second 90° pulse in after time 4, we will not be able to differentiate tissue A from tissue B because both will have equal MZ. This means MXY will be equal after flipping.

If, however, we send a pulse in after time 3, tissue A will yield a bigger MXY than tissue B because MZ did not fully recover in tissue B before the second pulse.

Time to repeat (TR)

- **TR is the time between 90° RF pulses**
- TR varies depending on the study and can be set by the operator:
  - **Long TR > 1500 msec**
  - **Short TR < 500 msec**
- Consider the T1 curves for brain and CSF



- We can see that brain has a shorter T1 than CSF
- If we wait a long time between RF pulses (TR_{long}) there is very little difference in signal intensity between brain and CSF
- If we repeat the RF pulse sooner (TR_{short}) there will be a greater difference in signal intensity because the longitudinal magnetisation of brain will have recovered and will lead to a greater transverse magnetisation after the flip following the RF pulse

Localising the signal

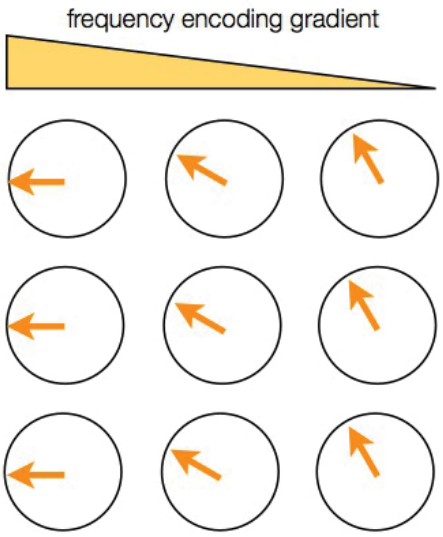
Slice selection

- Essentially, the external magnetic field generated by the MR scanner is homogeneous
- If a patient lies in a MR scanner, all of the protons in their body will precess at the same frequency (the Larmor frequency) regardless of where they are in the body:
  - This means that every proton is susceptible to excitation by the incoming RF pulses
- To examine a specific slice of the patient we **superimpose** a second magnetic field:
  - Known as the **slice selecting gradient** (**SSG**)
  - Produced by **gradient coils**
- Once the SSG has been applied, protons will precess at slightly different frequencies depending on where they are in the SSG
- It's possible to **orientate** the SSG **in any plane** without moving the patient
- Slice thickness is determined by the slope of the gradient:
  - **Steeper slice selecting gradient = thinner slice**

Frequency encoding gradient

- After slice selection, the signals that are emitted by the patient all come from a particular slice
- We will be splitting the slice into a matrix of columns and rows so we can reconstruct an image
- We figure out which column a signal comes from by applying a second magnetic field gradient:
  - The **frequency encoding gradient** (**FEG**)

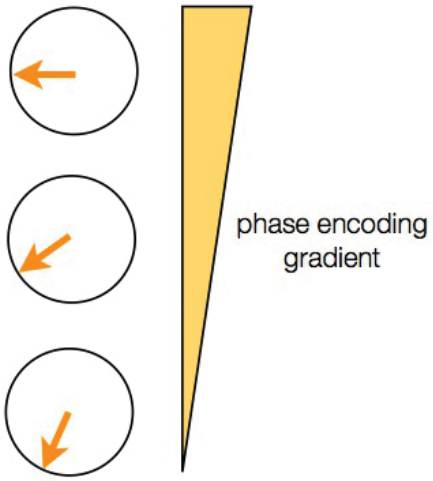
- The FEG causes protons to precess at different frequencies depending where along the gradient they lie:
  - **All protons in the same column of the slice precess at the same frequency and are in phase**



Protons in the same column precess at the same frequency and are in phase. Protons in different columns are precessing at a very slightly different frequency to the other columns (but are still in phase with each other)

Phase encoding gradient

- To figure out from which row a signal originates we need to apply a third gradient known as the **phase encoding gradient** (**PEG**)
- The PEG is applied for a short time, perpendicular to the FEG
- Whilst the PEG is on, protons in a particular column precess at different speeds depending on which part of the gradient they are in
- When the PEG is switched off, the protons in that column start to precess at the same frequency as before but they are now **out of phase with each other**
- We must apply a different PEG for each row we want in our matrix



Consider a single column. Before the PEG was switched on, protons were in phase. Whilst the PEG is on, the protons in this column precess at different frequencies depending on where they are along the gradient. When the PEG is switched off, the protons begin to precess at their original frequency (thus we still know what column they are in) but they are now out of phase.

Acquisition time

- The acquisition time can be calculated from the following equation:

$$\textbf{acquisition time} = \textbf{TR} \times \textbf{N} \times \textbf{N}_{\textbf{ex}}$$

- TR = time to repeat (msecs)
- N = number of desired rows in our matrix (i.e. the number of phase encoding gradients used)
- N_{ex} = number of excitations:
  - This is how many times we sample a particular signal before averaging it
  - Increasing N_{ex} will increase the **signal-to-noise ratio** (SNR)

Multislice techniques

- Most of TR is wasted if the scanner has to wait for up to 2 seconds before repeating pulses
- We can use this wasted time to deliver a succession of 90° and 180° pulses (each

of different frequency) to excite a series of slices before going back to the first one

- This allows us to **speed up acquisition time**

Summary

- We apply three magnetic field gradients (SSG, FEG and PEG):
  - From the patient we get a number of signals with different frequencies
- Signals with the same frequency will have come from a particular column in the slice of interest
- If we look at all the signals with the same frequency, we can determine their row by looking at their phase
- A computer can determine the components of our signal using a **Fourier transform** algorithm

K-space

What is K-space?

- It's a region of computer memory that **stores the raw signal data** from the scanner
- Once K-space is full (end of the scan) the data is mathematically processed into our image
- There's a lot of complex maths underlying this but for practical purposes, we don't need to know about it

Fourier transformation

- Fourier transformation is a mathematical technique that takes a complex wave (our MR signal) and breaks it down into several simple wave components:
  - **Any wave can be broken down into a collection of sine waves**
  - Any group of sine waves can be reconstructed into a complex wave
- Why Fourier transformation is so important is that it **allows us to move between two domains**:
  - E.g. from the time domain (complex MR signals generated with time) to the frequency domain (the complex wave is broken down into many waves each with a single frequency)
- K-space actually stores the **wavelength** and Fourier transformation allows us to move between this wavelength domain and the spatial domain (i.e. position)

The structure of K-space

- **Number of K-space rows = number of phase encoding gradients**
- **Number of K-space columns = number of frequency encoding gradients**
- Typically one row of K-space data is acquired per TR (not always the case)

- Although K-space is a matrix (rows and columns) its coordinate system does not start at the top left. Instead **(0,0) is at the centre**
- K-space centre:
  - Long wavelength components
  - Higher signal intensity
  - **Provides image contrast**
  - Derived from the shallow encoding gradients
- K-space **periphery**:
  - Short wavelength components
  - Lower signal intensity
  - **Provides fine detail (e.g. edges)**
  - Derived from the steep encoding gradients
- Try not to think that the top left corner of K-space corresponds to the top left corner of our final image because it does not:
  - **Every point in K-space provides some form of information for every point in the image**
- This fact is taken advantage of by advanced scanning techniques

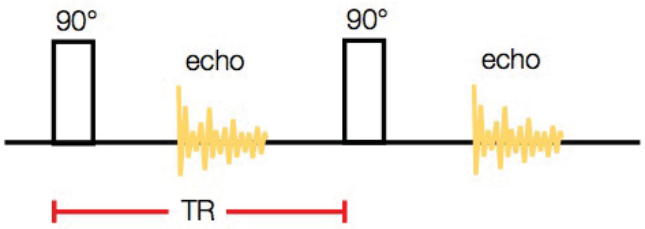
Saturation and inversion recovery pulse sequences

Overview

- A pulse sequence is a combination of RF pulses of a particular frequency and timing or a combination of RF pulses and magnetic field gradients
- They are responsible for determining which tissue characteristics contribute most to our image
- Can be split into standard and fast sequences

Partial saturation / saturation recovery

- The simplest type of pulse sequence
- Used to get faster T1 weighted images than is possible with inversion recovery sequences
- Give a 90° pulse followed TR msecs later by another 90° pulse and so on
- **Short TR**:
  - Known as **partial saturation recovery**
  - **T1 weighted** images
- **Long TR**:
  - Known as **saturation recovery**
  - **Proton density weighted** images



Inversion recovery sequences

- Used to provide **heavily T1 weighted images**
- Start with a **180° pulse**:
  - This flips all protons spin-down
  - They gradually regain their longitudinal magnetisation
  - After a certain number of msecs (TI, inversion time) we send in a 90° pulse:
    - This flips the (antiparallel) longitudinal magnetisation into the transverse plane, where it can be measured

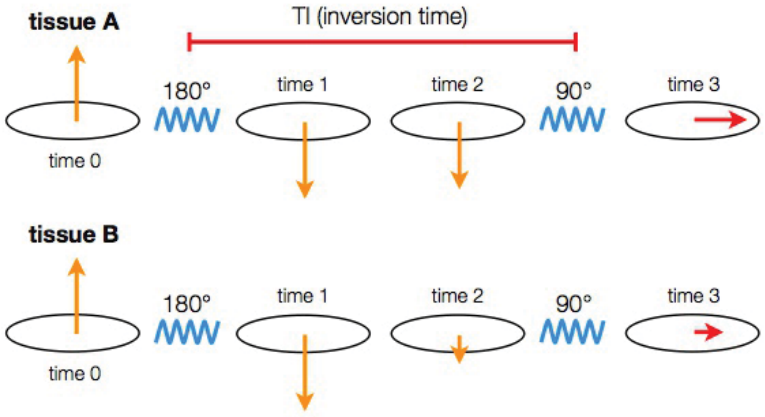
STIR (Short TI Inversion Recovery)

- Used to **suppress the signal from fat**
- Since fat has a very short T1, it will recover it's longitudinal magnetisation quicker than other tissues after the 180° pulse:
  - If we apply the 90° pulse after a short TI (~ 125 msecs), there will be little longitudinal magnetisation from fat (because M_z has to go through zero before becoming spin-up)
  - This means that there will be no M_z from fat to flip transversely and thus, we will get little or no signal
  - However, other tissues, such as water, will have some degree of M_z which will be flipped by the 90° pulse and, thus, will give a measurable signal

FLAIR (FLuid Attenuated Inversion Recovery)

- Used to **suppress the signal from fluid**
- Very similar to STIR but the TI is chosen such that there is no M_z from fluids at the moment the 90° pulse is given

Schematic of simple inversion recovery pulse sequence



Two tissues start with maximum longitudinal magnetisation (M_z, orange arrow). A 180° pulse is applied which flips M_z spin-down.

Gradually, the protons become spin-up again and M_z regrows.

A 90° pulse is applied and the magnetic vector is flipped into the transverse plane (M_{xy}, red arrow).

We can see from the illustration that tissue B recovers it's M_z quicker than tissue A (because it has a shorter T1). This means that when we flip M_z with a 90° pulse, the signal intensity from the two tissues are different.

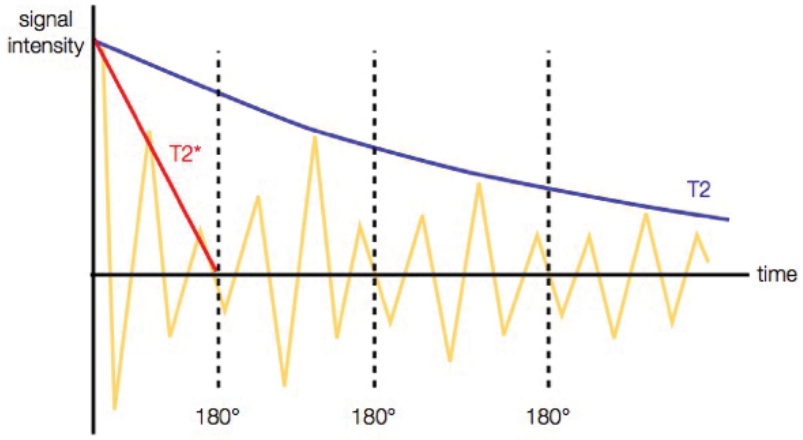
These differences in signal intensity are purely due to differences in T1.

The spin echo sequence

The 180° RF pulse

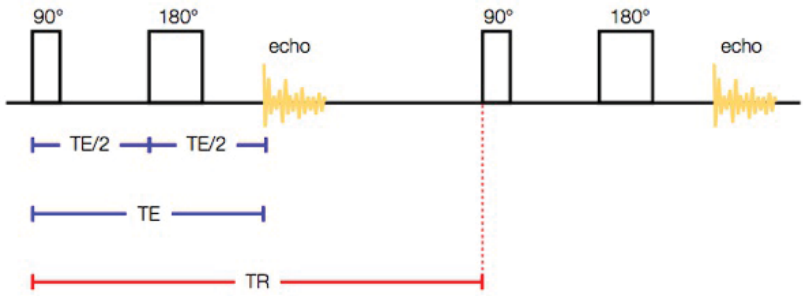
- Straight after the 90° pulse is given, all the protons are in phase
- The protons quickly begin to dephase once the 90° pulse has been turned off
- We then wait for a certain amount of msecs (TE/2) and send in a 180° pulse:
  - This reverses the direction of precession putting the protons back in phase after a further number of msecs (TE/2)
  - So, TE msecs after the 90° pulse, our protons are back in phase and we get a signal called the **spin echo**:
    - **TE stands for time to echo**

The T2 and T2* curves



- If we didn't use a 180° pulse, the protons would dephase very quickly because they are affected by both external and local magnetic field inhomogeneities:
  - This is represented by the red curve and is known as the **T2* curve**
- The **180° pulse serves to remove the effect of external field inhomogeneities**:
  - The resultant curve (blue line) should look familiar (it's the T2 curve):
    - Only affected by inhomogeneities in the local magnetic field

Schematic of the spin-echo sequence



TE and TR

- The spin-echo sequence is one of the most important pulse sequences in MR
- By altering TE and/or TR we can create T1, T2 or proton density weighted images

	Short TE < 20 msec	**Long TE** > 80 msec
Short TR < 500 msec	T1 weighted	Not used
Long TR > 1500 msec	PD weighted	T2 weighted

- **T1 weighted image**:
  - By using a short TR we do not allow full recovery of the longitudinal magnetisation at the point of applying the 90° pulse
  - Using a short TE reduces the influence of T2 factors
  - This means that differences in T1 can be maximised
- **T2 weighted image**:
  - By using a long TR, we can make sure that the longitudinal magnetisation has fully recovered before applying the 90° pulse
  - This means that differences in T1 times are irrelevant
  - We need to use a reasonably long TE to make sure that differences in T2

characteristics have time to become apparent

- **Proton density (PD) weighted image**:
  - A long TR and a short TE means that full longitudinal recovery has occurred (T1 effects minimised) and that T2 effects have not had a chance to become apparent
  - The signal is, therefore, dependent only on the number of protons present in the tissue

Echo trains

- We can make use of the relatively long TR used with the spin echo sequence
- Following each 90° pulse, two or more **successive 180° pulses produce successive echos**, each with an **increasing TE**
- The first echo may produce a PD-weighted image (since TE is short, say 20ms) whereas subsequent echoes produces ever more T2-weighted images (as TE is longer)
- ETL = echo train length
- There is a limit to the ETL (because the signal is undergoing T2 decay):
  - We typically use ETLs of 2 - 16

Turbo spin echo sequence

- We create an echo train by starting with a 90° pulse and then a series of 180° pulses
- **After each echo we apply a phase encoding gradient**
- This allows us to **reduce the acquisition time** by large factor (e.g. if ETL = 8 then our acquisition time is decreased by a factor of 8)
- Since scan time is reduced, we can increase the matrix size (since increasing matrix size by using more phase encoding gradients takes time) to **improve spatial resolution** (smaller voxels)

Gradient echo sequences

Overview

- Saturation, inversion and spin-echo sequences take a relatively long time to acquire
- **Gradient echo sequences are faster**:
  - Reduces motion artefact (patient movement, respiration, cardiac motion, etc)
  - Increases patient throughput
- There are several variations with different acronyms (e.g. FLASH or Fast Low Angle SHot)

Basic principles

- The **most time consuming part of standard pulses sequences is the TR** (time to repeat)
- With a spin echo sequence, we can't shorten TR too much for two reasons:
  1. A **180° pulse takes time to deliver**:
     - If TR is too short there would be insufficient time between the 90° pulses
  2. There would be **very little longitudinal magnetisation recovery**:
     - Consequently, there would be very little transversal magnetisation (Mxy) after the 90° pulse
- Therefore we need:
  1. A way to refocus the dephasing protons without using a slow 180° pulse
  2. Some way to preserve the longitudinal magnetisation before flipping

How do we refocus the protons?

- Instead of a 180° pulse, we use a **magnetic field gradient** (MFG):

- This is superimposed over the main magnetic field

- **Switch on** the MFG for a very short period of time:
  - Causes lots of inhomogeneities in the imaged slice:
  - Leads to **rapid dephasing** of the protons and a quicker loss of M_{xy}

- Switch off MFG and switch it back on but in the **opposite** direction:
  - Has a similar effect to using a 180° pulse and leads to **partial rephasing**

- We get a transient increase in signal intensity called the **gradient echo** (which we measure)
- It's important to note that because we are not using a 180° pulse, the **signal intensity is affected by T2* effects** and **not** T2 effects

How do we overcome the small M_{xy} signal caused by the short TR?

- Using a 90° pulse completely (albeit transiently) abolishes the longitudinal magnetisation (M_z)
- Instead, we use a **smaller flip angle** (**10 - 35°**):
  - This means that there is always a substantial amount of M_z left that can be tilted by the next flip pulse

Summary

- Why are gradient echo sequences faster?
  - Smaller flip angles mean the RF pulse duration is shorter
  - There is no time consuming 180° pulse
  - We can use a short TR
- Larger flip angle = more T1 weighting
- Longer TE (time to echo) = more T2* weighting
- Gradient echo sequences provide intense signal from vessels

The magnet

Permanent magnets

- As the name suggests, these are **always magnetic**
- Advantages:
  - Do not require energy input to generate a magnetic field
- Disadvantages:
  - **Thermal instability**
  - **Limited magnetic field strength**
  - **Heavy** (a 0.3T permanent magnet can weigh 100 tons)
- They are occasionally used in low field strength scanners

Resistive magnets

- Also known as electromagnets
- An **electrical current is passed through a coil of wire and generates a magnetic field**
- Advantages:
  - Can generate a higher field strength than permanent magnets
- Disadvantages:
  - Are only magnetic when a current is flowing, therefore, **require a lot of energy**
  - Get very **hot** (as an electrical current is constantly flowing):
    - This becomes impracticable at high field strengths

Superconducting magnets

- These are the **most common** type of magnet used in MR scanners today
- An electrical current is passed through a superconducting material:
  - **Superconductors have zero electrical resistance**
  - Once an electrical current is sent in, it flows permanently, creating a magnetic field
- Superconducting materials need to be cooled to near absolute zero (4°K or –269°C):
  - This is done with a **cryogen** (either liquid helium or nitrogen)
- The magnetic field can be turned off by **quenching**:
  1. Cryogen is released
  2. Superconducting material warms up and ceases to be a superconductor
  3. Electrical resistance rises, the current stops flowing and the field disappears
- Advantages:
  - **High field strength**
  - **Excellent magnetic field homogeneity** (**10–50 ppm** over a 45cm diameter region)
- Disadvantages:
  - High **cost**
  - Expensive cryogens

The coils

Overview

- RF coils send in the RF pulse to excite the protons and receive the resulting signal
- The same or different coils can be used for transmitting the RF pulse and receiving the signal

Gradient coils

- There are three sets of gradient coils:
  - **Slice selecting** gradient coil
  - **Frequency encoding** gradient coil
  - **Phase encoding** gradient coil
- The rapid switching of these coils (< 1 ms) is the source of the loud noise the MR scanner emits

Shim coils

- These coils permit **fine adjustments** to be made to the main magnetic field
- **Improves** the **homogeneity** of the field

RF coils

- These transmit the RF pulse and receive the emitted signal
- They should be as close to the region of interest as possible
- **Body coil**:
  - Usually a permanent part of the MR scanner

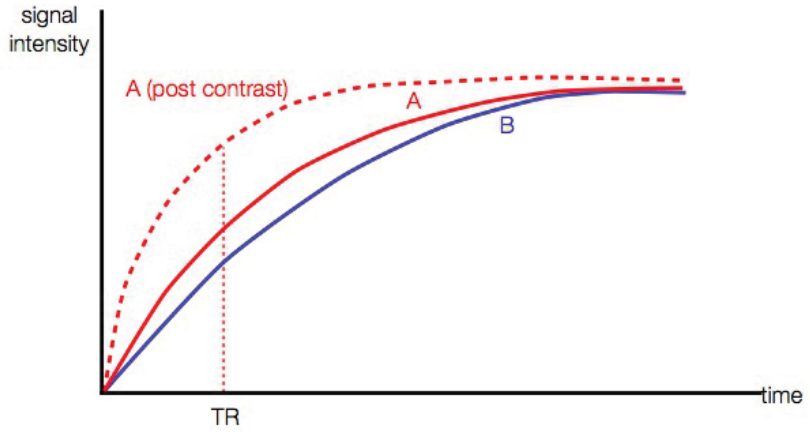
- Used to transmit the RF pulse for all types of scan
- Receives the MR signal when larger parts of the body are imaged

- **Head coil**:
  - Transmits and receives
  - Used for brain scanning
- **Surface coils**:
  - Receive only
  - Are separate coils that are moulded to cover a certain anatomical region
  - Receive signals from a depth equal to the coil radius
  - **Improve resolution** (by permitting smaller voxels) and have a **better signal-to-noise ratio**
  - Have a **smaller FOV**

MR contrast media

Overview

- MR contrast media are **paramagnetic** substances:
  - **Shorten the T1 and T2** of tissues that take them up:
    - For a given **TR** there is **more signal** (T1 curves shifted to the left)
    - For a given **TE** there is **less signal** (T2 curve shifted to the left)
- The most common contrast agent used is **Gadolinium** (Gd):
  - Gd is a toxic rare earth that is chelated with DTPA to remove its toxicity
    - Well tolerated
    - Does not pass through the intact blood-brain barrier (will traverse if damaged)
- Since loss of signal (T2 shortening) is harder to appreciate than enhancement (T1 shortening), **T1 weighted images are most commonly used after contrast administration**:
  - A shortening of T1 results in a higher signal (and hence brighter image) of the contrast-enhanced tissue
- As Gd shortens T1, it is also possible to reduce TR and still achieve the same signal:
  - Can result in **faster acquisition times** (as TR is the rate limiting factor)



Here are the T1 curves of two tissues (A & B). Gd contrast is administered and **only** taken up by tissue A. The result is that the T1 of tissue A is shortened and its curve shifts to the left. This means that for a given TR, the difference in signal intensity between A and B is greater, thus providing us with better contrast between the two tissues.

MR angiography

How vessels appear in a MR image depends on many factors including the type of vessel (vein or artery), the sequence used and the imaging parameters.

Spin echo (SE) sequences and blood flow

- **Flow void**:
  - This is the term given to the phenomenon that causes a vessel with **fast-flowing blood** within it (e.g. arteries) to give no signal and, thus, appear **dark** on the image.
  - Protons in blood within the vessel in the slice get flipped by the 90° pulse
  - The blood moves along the vessel and out of the slice, being replaced by unexcited blood from outside the slice
  - Because the unexcited blood is not in the transverse plane we get no signal from the vessel
- **Flow enhancement**:
  - This is the term given to the phenomenon causing a vessel with **slow-flowing blood** within it (e.g. veins) to give a high signal and appear **bright** on the image.
  - Protons in the slice are flipped by the 90° pulse
  - Protons in the slice slowly regain longitudinal magnetisation
  - Unexcited blood from outside the slice moves into the vessel and possesses full longitudinal magnetisation
  - A second RF pulse flips the protons and we see a greater degree of transverse magnetisation in the vessel than surrounding it and thus get a stronger signal

Gradient echo sequences and blood flow

- **Flowing blood and CSF usually appear bright**
- This is because the gradient pulse (unlike the 180° pulse in SE sequences)

rephases all spins (even those outside the slice):

- Essentially fast moving blood causes flow enhancement much like slow moving blood in SE sequences

Diffusion weighted imaging

Overview

- **Requires a fast imaging sequence** such as gradient echo
- Two large gradient pulses are applied after each RF pulse:
  - If the **protons are moving** then the protons undergo a **phase shift**:
    - Reduction in signal
  - If they are static, their phase remains unaltered:
    - Increase in signal
- We therefore get a **high signal from tissues with restricted diffusion**
- Diffusion-weighted images are all T2-weighted to varying degrees

Apparent diffusion coefficient (ADC)

- How much an image is diffusion-weighted is measured in terms of it's **b-factor**:
  - 0 = no diffusion weighting (T2 image)
  - 1000 = heavily diffusion weighted
- ADC measures the magnitude of diffusion and removes the effects of T2:
  - **Low ADC = restricted diffusion** (high signal)
  - High ADC = normal diffusion

MR artefacts

Aliasing

- Happens when the **field of view (FOV) is too small**
- Occurs when protons outside our FOV are excited by the RF pulse:
  - The resulting signal detected by the body coil is then falsely allocated to a pixel in our image matrix (when really it shouldn't be in the image)
- Produces **image wrap-around in the phase encoding direction**
- Can be reduced by:
  - Using a **surface coil** that more closely matches the FOV
  - **Increase FOV**

Motion artefact

- Usually occurs in the **phase encoding direction** (because frequency encoding is very quick)
- Can be reduced by:
  - Immobilising or sedating the patient
  - Swap the frequency and phase encoding direction to remove artefact from area of interest

Chemical shift

- The **precession frequency differs very slightly for protons in different chemical environments**
- During the frequency encoding step, fat precesses slightly slower than water:
  - ~ 3.5 ppm (although this increases as B_0 increases)
- The scanner cannot know about these tiny differences:

    - Artefact occurs in the **frequency encoding direction**
    - Causes a **dark or bright band at fat : water interfaces**
- Can be reduced by:
    - Using a **wider receiver bandwidth**
    - Using a **steeper gradient**

Ringing

- **Parallel stripes at high contrast interfaces** (e.g. between CSF and the spinal cord)
- Caused by under-sampling in the **phase encoding direction**:
    - The pixel size is too large to represent the boundary accurately
- Can be reduced by:
    - **Increasing the number of phase encodings**
    - **Reducing the FOV**

MR safety

Overview

- MRI does not involve ionising radiation
- In current clinical practice very little, if any, ill effects are seen
- Each of the components of the MR scanner has the potential to cause harm

The static magnetic field

- A voltage could be induced in flowing blood if the static magnetic field is very large
- The MHRA (Medicines and Healthcare products Regulatory Agency) have provided guidance on the safety of the main magnetic field:
  - **Pregnant patients should never be exposed to > 2.5 T**
  - **No patient should be exposed to > 4 T**
  - If a patient is exposed to > 2.5 T they must have a panic button and be in constant visual contact
  - Over 24 hours, the **average staff exposure should be < 0.2 T**
  - Patients with a **pacemaker should not be exposed to > 0.5 mT** (note millitesla)

The gradient fields

- Electric fields are produced perpendicular to the gradient fields
- These electric fields induce **eddy currents** in conductive tissues causing stimulation:
  - An eddy current is an electric current induced in a conductor when that conductor is exposed to a changing magnetic field
  - Can cause unpleasant effects:

        - **Peripheral nerve stimulation**
        - **Muscle contraction**
        - **Cardiac arrhythmias** (very rare)
        - **Flashing visual lights**
        - Metallic taste sensation
- These effects are **not seen below 20 T/sec**. Generally start at around 60 T/sec
- As a precaution, **MRI is avoided during the first trimester of pregnancy**
- **Acoustic noise**:
    - Most MR scanners do not exceed 120 dB
    - **Hearing protection is required > 90 dB** to prevent irreparable hearing damage
    - Ear plugs reduce noise by only 10 - 30 dB
    - There are concerns about the effects of loud noise on the **fetus**

Radiofrequency fields

- **Microwave heating** is the main concern although this is alleviated somewhat by vasodilatation:
    - The cornea (avascular) and testes (small blood supply) are higher risk
    - Heating of metallic implants can occur
    - **Skin temperature rise should not > 1 °C**
    - Specific absorption ratio (SAR):
        - This is the **RF energy deposited per unit mass of tissue**
        - Unit is **W/Kg**
        - SAR is greater for larger body parts and for the 180° pulse compared to the 90° pulse
- The vast majority of adverse incidents in MR arise from burns secondary to contact with monitoring leads and electrodes