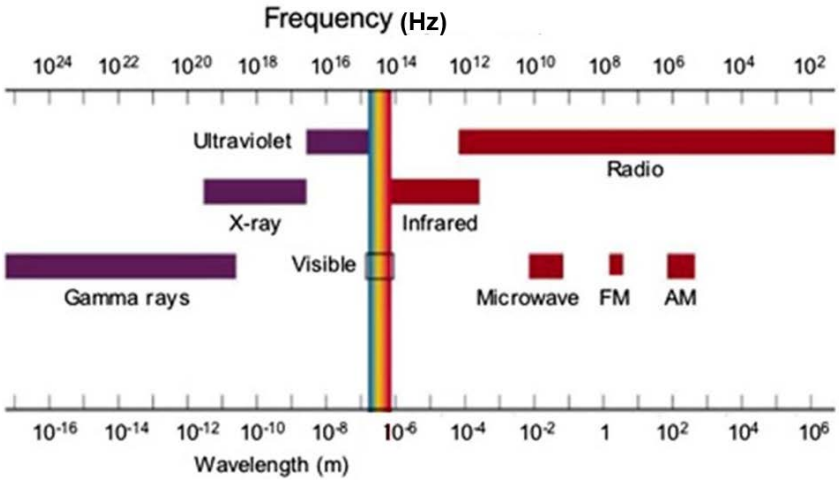
OCR A Physics A level

Topic 6.5: Medical Imaging

(Content in italics is not mentioned specifically in the course specification but is nevertheless topical, relevant and possibly examinable)

X-ray Production

X-rays are produced when charged particles are rapidly decelerated (or accelerated) and their kinetic energy is transformed into high frequency photons of electromagnetic radiation. X-rays and gamma rays have overlapping frequency spectra i.e. a gamma ray and an X-ray can have the same frequency (and so equal wavelength and energy). The only distinction between these gamma and X-rays is then their history. Gamma rays are produced via radioactive decay or during particle collisions with a mass defect e.g. electron-positron annihilation or nuclear fission of uranium-235 (see Nuclear and Particle Physics 6.4). X-rays are produced by Bremsstrahlung or braking radiation which is when radiation is given off by charged particles due to their acceleration. X-rays used in medical imaging are often referred to as soft X-rays as they have energies generally lower than gamma rays.



X-ray Tubes

X-ray tubes produce X-rays by accelerating electrons in a high-voltage electric field then rapidly decelerating them via collisions with a hard metal anode (positive electrode) e.g. tungsten. Electrons are first emitted from a heater or filament (cathode or negative electrode) into a vacuum tube via thermionic emission. Thermionic emission is the process by which electrons are emitted from a heated source, in this case a high resistance coiled wire. The vacuum tube is needed to prevent the electrons from colliding with air molecules before they have acquired enough energy to emit X-rays.

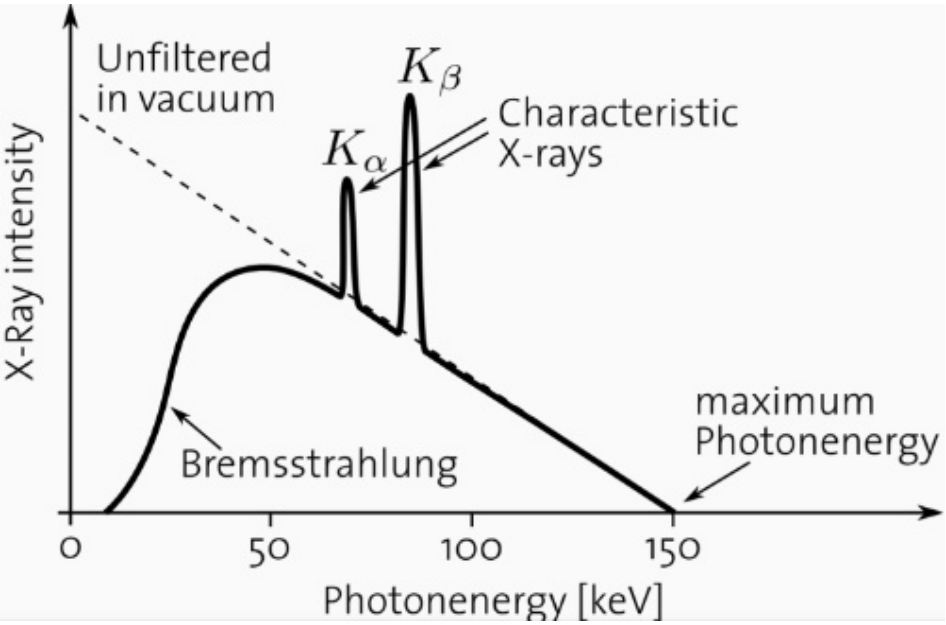
An external power supply produces a potential difference between the cathode and the anode of up to 200kV. Therefore, the electrons gain a kinetic energy of up to 200keV (see Nuclear and Particle Physics 6.4). Upon collision, the electrons decelerate rapidly and some of their kinetic energy (~1%) is emitted as X-rays. The rest is lost to thermal energy in the anode. To prevent the anode from overheating, the anode is either rotated so a new area is constantly exposed to the electron beam or it is cooled with circulating water supply.

The X-rays are emitted in every direction from the anode. However, a straight and parallel or collimated beam is much more useful as it can be directed at specific parts of a patient's anatomy, e.g. a broken limb, and minimise exposure. Therefore, the vacuum

tube is encased in a material which is thinner in one place than the rest, the window, so that X-rays only emerge in one place outside the tube. The beam is then directed into a series of straight and parallel metal tubes otherwise known as a collimator that further collimate the beam by absorbing any rays that are not parallel to the axis of the tubes.

X-ray Spectra

Braking radiation produces a broad range X-ray wavelengths with a hump-shaped intensity profile as seen below. However, there are also a few sharp lines of characteristic radiation that are not due to decelerating electrons. These lines are instead caused by incident electrons knocking out bound low energy level electrons in the anode atoms. Higher energy electrons will then transition down to the unoccupied shell and their excess energy will be emitted as radiation. Photons produced via this process have specific wavelengths and so the number of X-rays at these energies will be greater and so the intensity here will be higher.



Ionising Radiation

X-rays (as well as gamma-rays and UV rays) have such high energies that they can ionise matter i.e. cause electrons to be emitted from atoms (e.g. the photoelectric effect). In living cells, X-rays can ionise DNA and other tissues causing damage to the organism and possible harmful mutations to the genes. Thus, to minimise this damage, living tissue must be exposed to only low intensity beams for with short periods (exposure times). The ability of X-rays to destroy cells can be utilised in the treatment of cancer (radiotherapy). A beam of X-rays is directed towards cancerous cells in the hope that it will reduce the number of cancer cells without causing too much damage to healthy tissue.

X-ray Attenuation Mechanisms

As X-rays ionise the matter they pass through, they lose some or all of their energy to the atoms or molecules in their way. This causes a gradual decrease in the energy or

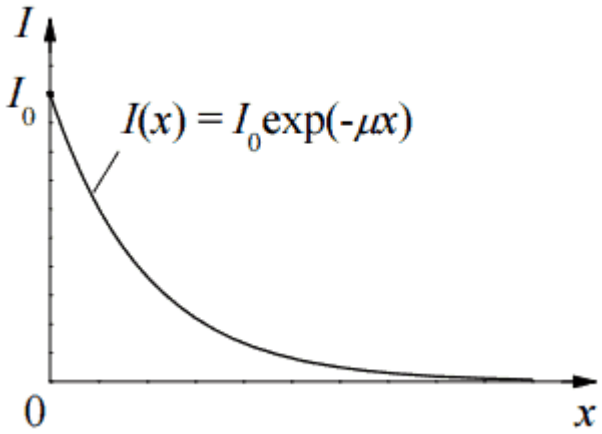
intensity (power per unit cross-sectional area i.e. $I = \frac{P}{A}$ in units of Wm^{-2}) of the X-ray beam known as attenuation. Different materials attenuate X-rays to a different extent so tissues can be contrasted by measuring the intensity of the attenuated beam once it has passed through the patient. For example, bone attenuates X-rays to a greater extent than flesh or other soft tissues. Therefore, when a limb is exposed to an X-ray beam the X-rays that collide with bone are more likely to be absorbed and the beam shows more attenuation directly behind the bone. If a photographic film is held behind the patient's limb then it will be blackened less if it is in the direct path of X-rays that passed through bone. This is clearly seen as a white outline of the patient's skeleton. Nowadays, digital detectors are used as the images are easier to process, store and transfer.

The intensity of a collimated beam of X-rays (collimated so there is no intensity decrease due to the spreading out of the beam) decreases exponentially. For example, if the beam intensity halves in 1cm of bone, then in 2cm it will quarter and in 3cm it will be an eighth of its original value.

The attenuation of an X-ray beam in a medium can be evaluated by

$$I = I_0 e^{-\mu x}$$

where I_0 is the initial intensity before entering the medium, I is the attenuated intensity after passing through a thickness x (m) of the medium and μ (m^{-1}) is a property of the material known as the attenuation or absorption coefficient which describes how well a medium absorbs X-rays i.e. the attenuation coefficient for bone will be higher than that of muscle.



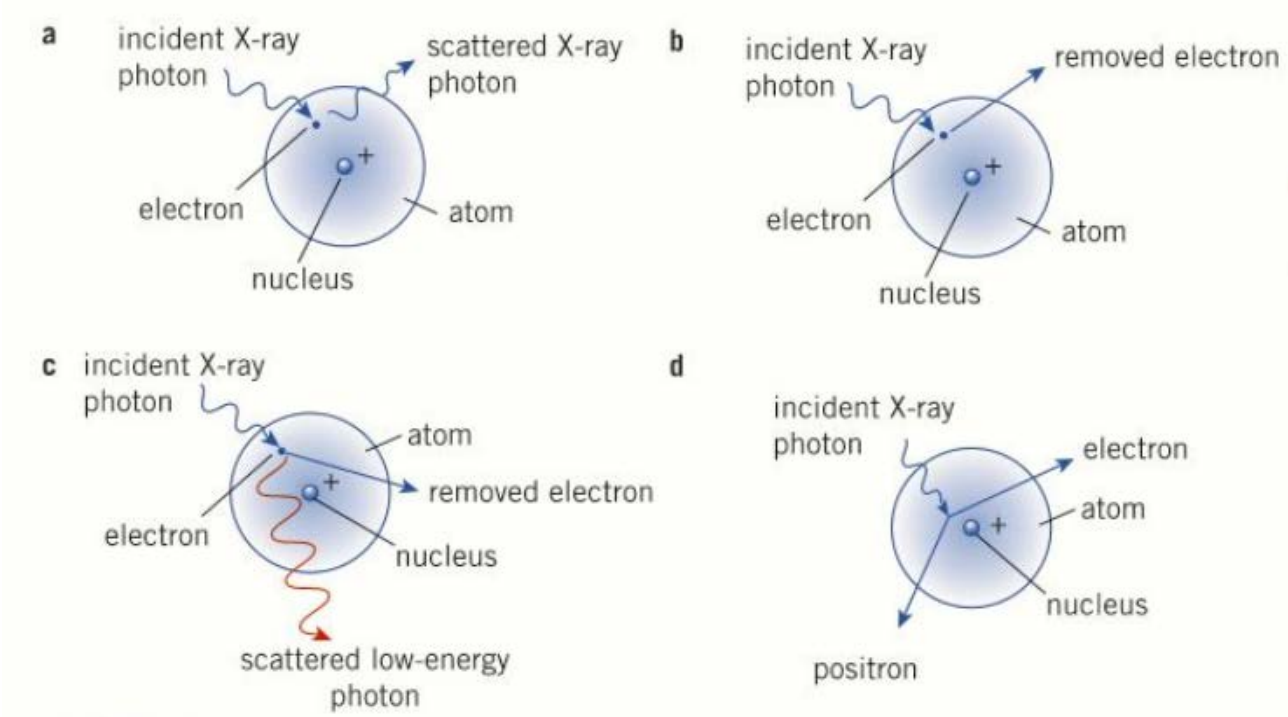
Absorption Mechanisms

There are four important mechanisms by which X-rays are absorbed and beams are attenuated.

- Simple Scattering: X-rays of energy 1-20 keV will reflect off layers of atoms or molecules in the material as they do not have enough energy to undergo more complex processes.
- Photoelectric Effect: X-rays of energy less than 100 keV can be absorbed by electrons in the material as they have the same energy as the ionisation energies of the atoms. When an X-ray is absorbed by an atom, a photoelectron is released

and another electron may transition down to the lower energy level emitting a scattered photon in the process.

- Compton Effect: X-rays of 0.5 to 5 MeV lose only a fraction of their energy to electrons in the absorbing materials. This is due to an inelastic interaction between the photon and the electron. The scattered X-ray photon will have less energy than before, and so its wavelength will be greater. The Compton electron will be scattered in a different direction as momentum must be conserved.
- Pair Production: When X-ray energy is greater than 1.02 MeV passes through the electric field of an atom it will spontaneously produce an electron-positron pair via the mass-energy relation. The positron will then go on to collide with another electron and annihilate producing photons. This process is not very important in medical X-rays as the photon energies are usually not high enough to produce an electron-positron pair.



▲ **Figure 2** *Attenuation mechanisms: (a) simple scatter – the X-ray photon is scattered elastically by an electron; (b) photoelectric effect – the X-ray photon disappears and removes an electron from the atom; (c) Compton scattering – the X-ray photon is scattered by an electron, its energy is reduced, and the electron is ejected from the atom; (d) pair production – the X-ray photon disappears to produce an electron–positron pair*

X-ray Imaging

Contrast media are high attenuation coefficient materials that have heavy atoms with a large proton number and so a large number of electrons. Contrast media are easily identified on digital X-ray images as, like bone, they cause a lower detected intensity to be seen in the direct path of the contrast media. These materials, such as barium, $Z = 56$, or iodine, $Z = 53$, and are far better at absorbing X-rays than soft tissues as soft

tissues low average proton numbers ($Z \approx 7$) and so low attenuation coefficients. The relationship between the absorption coefficient μ and the proton number is largely defined by the photoelectric effect as this is the absorption mechanism that occurs most frequently at soft X-ray frequencies. This relationship can be summarised as

$$\mu \propto Z^3$$

Therefore, contrast media can be around 500 times better absorbers than the surrounding soft tissue as the attenuation coefficient is proportional to the proton number cubed.

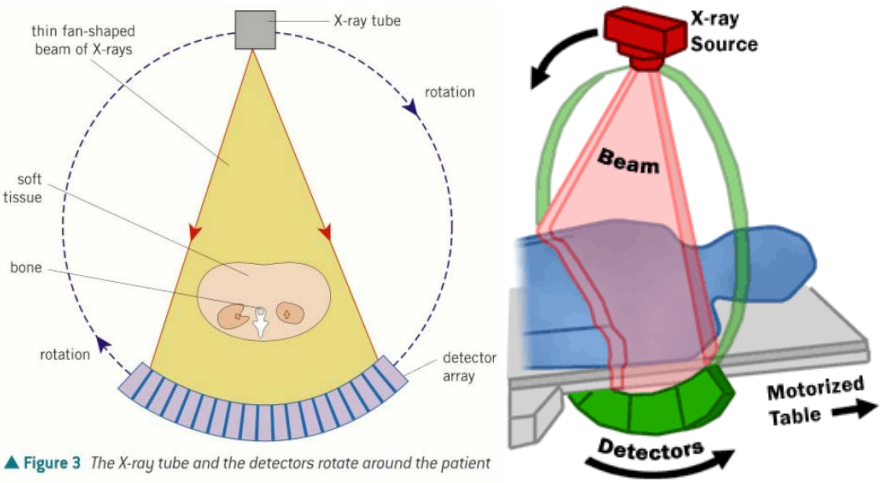
Iodine is used as a contrast medium in liquids i.e. to observe blood flow. An organic iodine compound is injected into the blood stream and the patient is then exposed to X-rays. Areas with low attenuated intensities identify regions of healthy blood flow and areas that show little attenuation locate poor blood flow.

Barium sulphate is used as a contrast medium in the digestive system. The patient is given a white liquid mixture (a 'barium meal') which the patient swallows before the X-ray image is taken. Similarly, areas with lower measured intensities outline the intestines and can be used to locate blockages.

Computerised Axial Tomography (CAT)

A conventional X-ray image is a useful diagnostic tool. It is cheap and quick however it only provides a two-dimensional image and cannot distinguish overlapping bones or different soft tissues.

A computerised axial tomography or CAT scan is an effective way of examining the internal three-dimensional structure of a patient using X-ray imaging. The CAT scanner records a large number of 2D X-ray images then assembles them into a 3D image with the help of computer software. The resolution of the image is greater than the conventional X-ray and the CAT scan can distinguish between differing soft tissues. However, CAT scans take a significantly longer time and so expose the patient to a far greater dose of ionising radiation.



▲ Figure 3 *The X-ray tube and the detectors rotate around the patient*

CAT scanners contain an X-ray tube that generates a fan-shaped beam. This is directed onto the patient whilst lying on their back. A ring of electronic detectors opposite detect

the X-ray beam intensity. This information is then converted into electrical signals and processed to reconstruct the tissues that the beam has passed through. The X-ray tube and the detectors can then rotate about the patient and move up and down their length to create a full 3D image of the patient's body when all images of each slice are stitched together. The image can then be displayed on a computer monitor and analysed.

Medical Tracers

In medicine, radioactive isotopes are combined with specific elements to form compounds that collect in particular locations in the body. These compounds are known as medical tracers. Radiopharmaceuticals such as these are used in both diagnosis and therapy.

In non-invasive diagnosis, these sources have to be placed inside the patient's body and their emissions detected from the outside. This makes gamma-emitters most useful as they are least ionising and most penetrative. Beta and alpha emitters are more ionising so would cause significant damage.

Radioisotopes used in medicine also tend to have high activities and short half-lives so that the imaging can be achieved quickly, the patient is exposed to a minimal dosage of harmful radiation and only small amounts of the radioactive source are needed.

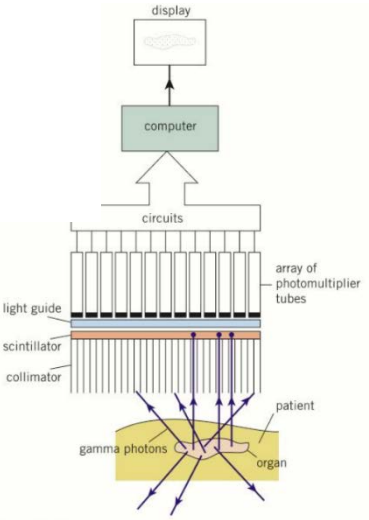
Many of these radioisotopes are produced artificially on-site due to their short half-lives. This is because these radioisotopes must be utilised almost immediately or their activity will decay below measurable levels.

Flourine-18, a source commonly used in Positron Emission Tomography (PET) scans. It undergoes beta plus decay releasing a positron from a proton and forming a neutron in the nucleus (see Nuclear and Particle Physics 6.4). The positron then annihilates with an electron in the patient's body to form a pair of gamma photons which are detected to locate the F-18 source in the patient's tissue. F-18 has a half-life of approximately 110 minutes and is produced through the nuclear transformation of oxygen-18 in a particle accelerator i.e. O-18 is bombarded with protons. Tracers of F-18 include sodium fluoride which is used for skeletal imaging as it displays rapid bone uptake.

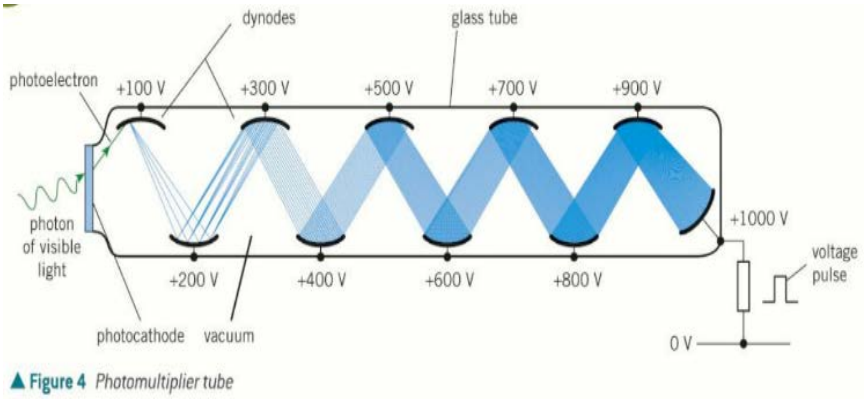
Technetium-99m is a versatile radioisotope that is used to monitor many major organs. The 'm' stands for metastable which means that it remains in a high energy state for prolonged periods of time. When it decays via gamma emission (half-life ~ 6 hours) it then forms Tc-99 which is stable with a half-life of 210,000 years. This ensures that the patient receives limited ionising radiation from any residual Tc-99 left in their body after the procedure. The photon is released with an exact energy of 140 keV and so can be identified easily. Tc-99m can be combined to form $NaTcO_4$ which when injected into the patient will target brain cells and so can be used to image the brain.

Gamma Cameras

Gamma cameras detect gamma photons emitted from medical tracers within the body. When gamma rays are emitted from the body they travel in every direction. This makes tracing the location of their emission difficult. Therefore, a collimator is used so that only photons travelling in one direction are detected. The collimator is made of a mesh of parallel honeycomb-shaped tubes so that photons travelling in any direction other than that of the axis of the tubes is incident upon the walls of the collimator and absorbed. The collimator has to be made of a high density metal to ensure that the gamma is absorbed.



▲ Figure 2 *The components of a gamma camera*



▲ Figure 4 *Photomultiplier tube*

The collimated photons are then incident on a scintillation crystal (e.g. sodium iodide) which is a material that will emit many photons when a high energy photon is incident upon it. Approximately a tenth of the gamma photons are absorbed onto the scintillator but each photon produces thousands of visible photons. The visible photons are then directed onto a photocathode which produces an electron for each visible photon detected.

This electron then passes into a photomultiplier tube. Photomultipliers contain a set of dynodes (intermediate electrodes which emit additional electrons) which are kept at high voltage so that as the initial electron hits them a cascade of electrons is generated amplifying the signal. The position of the impact in the scintillator is used to locate the emission site of the original gamma photon. This signal is finally detected by a computer and displayed on a screen.

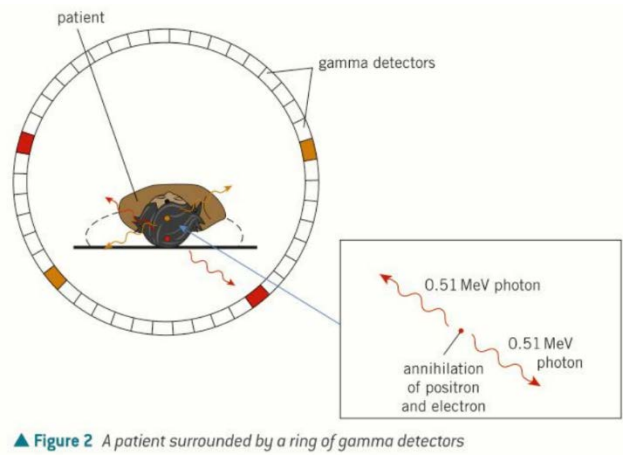
Positron Emission Tomography (PET)

As with conventional X-rays and CAT scans, a gamma camera produces only a 2D low resolution compared with a PET scanner. A PET scanner is a ring of gamma cameras placed around the patient so that an accurate 3D image can be generated from the emission site of the gamma photons.

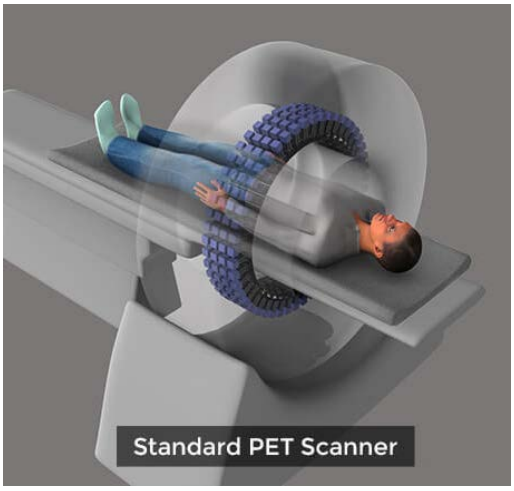
When positron-emitters such as F-18 are used, a pair of gamma photons are produced from the electron-positron annihilation. These two gamma photons are emitted in opposite directions to obey the conservation of momentum. Each of the photons is detected at one of two diametrically opposed detectors in the ring. Their arrival times are recorded. Based on these arrival times, the exact location of the annihilation event can be calculated as the speed of the photons is known. As annihilation occurs soon after beta emission, the site of the tracer can be estimated. This is repeated until a 3D model of the tracer locations can be produced. The tracer density can be determined from the rate of photons emitted in each region and so the tracer uptake in these regions can be known.

The tracer fluorodeoxyglucose (glucose substituted with F-18) is used in PET scanners to locate areas in the body with high rates of respiration such as cancerous tumours or active parts of the brain.

PET is a non-invasive technique which accurately demonstrates organ function and can be used to observe the effects of various medications. However, it is expensive and requires tracers to be synthesised on-site.



▲ Figure 2 *A patient surrounded by a ring of gamma detectors*



Ultrasound

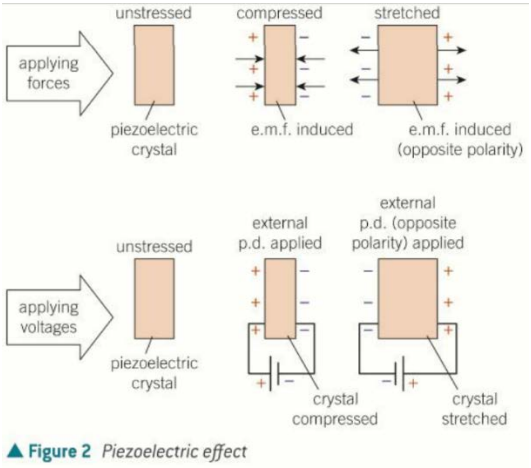
Ultrasound is a longitudinal sound wave with a frequency greater than human hearing range i.e. greater than 20 kHz (a typical frequency in medical diagnosis is 5MHz). The sound waves can be refracted, reflected, Doppler shifted and diffracted. Using these wave behaviours, properties of the media under study can be determined from measurements of the waves after they have been in contact with the medium i.e. diffraction can be used to identifying apertures or features of a few millimetres in size.

Ultrasound is a non-ionising and non-invasive technique that is quick and affordable. It is particularly useful for finding the boundary between two media. A transducer in the ultrasound device is used to produce an electrical signals from the soundwaves. This can then be analysed by computer software and an image can be generated and displayed on a screen.

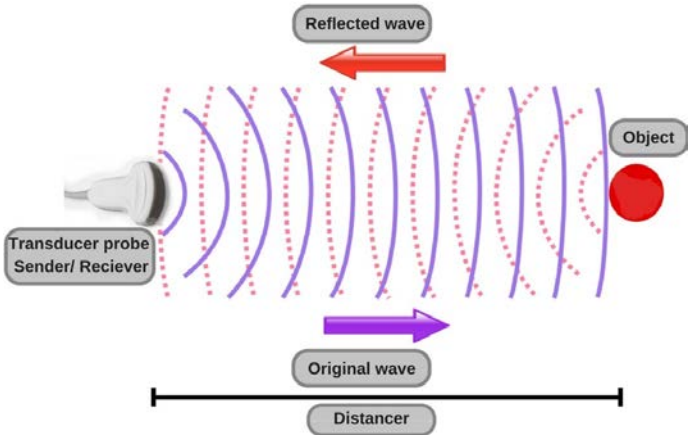
Piezoelectric Effect

A piezoelectric material generates a voltage when it is contracted or expanded or will contract and expand if a voltage is applied. Therefore, by applying a voltage to a piezoelectric crystal we can produce ultrasound vibrations and a piezoelectric crystal absorbing ultrasound will produce an alternating voltage. Piezoelectric crystals tend to be made from quartz, polymeric or ceramic materials.

An ultrasound transducer has an alternating potential difference that causes repetitive compression and stretching of the crystal. A resonant frequency of the crystal is chosen to increase the intensity. Once the ultrasound has been created the potential difference is turned off and the reflected signal is read.



▲ Figure 2 *Piezoelectric effect*



There are two types of ultrasound scan:

- A scan: The simplest type of scan uses a single transducer to emit a signal and then later receive the reflected signal back. It is used to determine distances from the ultrasound device to the point of reflection (usually the boundary between two media). This is achieved by measuring the time delay between generating and receiving the signal and using the speed of sound in the media to approximate the distance.
- B scan: A more complex scan that produces a 2D image. This is accomplished by moving the transducer over the patient's skin. At each position, the scan produces a measure of the time interval and so the distance to the reflection point between signal production and reception. The B scan is a series of A scans that are stitched together to form an image.

Ultrasound waves are pulsed to allow time for the reflected waves to be received. Smaller wavelengths give more detailed images as they will allow the soundwaves to diffract around finer points of detail on the object under inspection.

Acoustic properties of Ultrasound

The acoustic impedance, Z (measured in units of $kgm^{-2}s^{-1}$), of a sound wave is defined as the product of its density, ρ, and the speed of sound in that medium, c.

$$Z = \rho c$$

When ultrasound hits a boundary between two media, a fraction of the wave energy (or intensity) is reflected and the rest passes though and is transmitted. The fraction reflected depends on the acoustic impedances of the media.

If a sound wave travelling through a medium of impedance Z_1 is then incident on a boundary with a second medium of impedance Z_2, the fraction of the original wave intensity, I_0, that is reflected, known as I_r, depends on the impedances according to the following equation

$$\frac{I_r}{I_0} = \frac{(Z_2 - Z_1)^2}{(Z_2 + Z_1)^2}$$

where $\frac{I_r}{I_0}$ is known as the reflection coefficient or the fraction of the intensity reflected at the boundary. This equation applies when the angle of incidence is 0 i.e. the soundwaves are travelling perpendicular to the boundary so no refraction occurs.

When Z_1 is similar to Z_2 the fraction of intensity reflected is small so that most of the wave energy is transmitted. When Z_1 is very different from Z_2 the fraction of intensity transmitted is small so that most of the wave energy is reflected. Therefore, it is clear that if an ultrasound transducer was operated in air next to the patient's skin there would be a large amount of reflection due to the differences in the impedances of the two media (99.9% reflection). To maximise the transmission of the ultrasound into the patient (and maximise the reflected intensities and level of detail), an impedance matching gel with a very similar impedance to skin is used between the transducer and the skin.

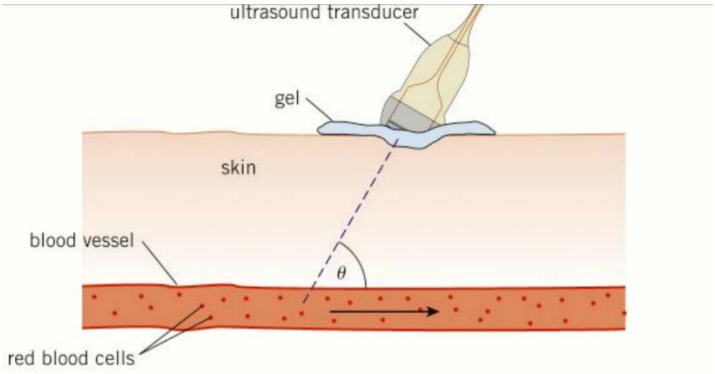
Doppler Effect

The Doppler Effect is the change in frequency of a wave when it is reflected or produced by a moving source. Doppler imaging is used as a non-invasive technique to measure the speed of blood flow. Ultrasound waves are sent into a blood vessel. The blood flowing past the transducer contains iron that reflects the wave back to the transducer. Depending on the direction and speed of flow, the ultrasound frequency is either shifted up or down as shown by the following

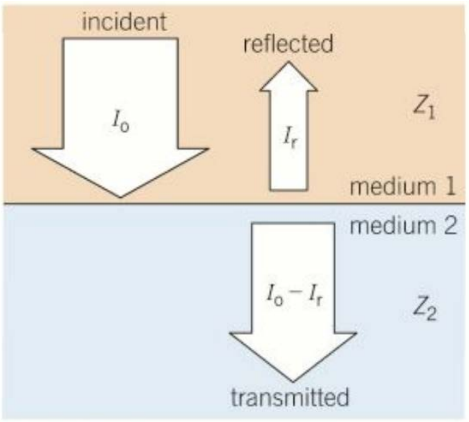
$$\Delta f = \frac{2fv\cos\theta}{c}$$

where Δf is the observed frequency shift, f is the original frequency, v is the speed of the flow, c is the speed of ultrasound in blood and θ is the angle between the probe and the direction of blood flow. A typical shift for a 5-15 MHz ultrasound is around 3 kHz which can be detected easily. This technique can be utilised to demonstrate blood flow

through arteries and veins and reveal blood clots, narrowing artery walls and calculate the volume of blood flow.



▲ **Figure 3** *Ultrasound transducer used to determine the speed of blood flow*



▲ **Figure 2** *Reflection and transmission of normally incident ultrasound at a boundary between media*