Teaching guide: Medical physics

This teaching guide provides background material for teachers preparing students for the Medical physics option of our A-level Physics specification (7408). It provides more detail on specification topics that teachers may not be familiar with and should be used alongside the specification. This guide is not intended as a set of teaching notes.

Contents | Page

Section 1: The eye

a) Structure

The human body is sensitive to most parts of the electromagnetic spectrum. The whole body feels warmth when irradiated with infrared radiation, whilst ultraviolet radiation may cause damage in the form of burning (reddening of the skin and sun tan) and skin cancer. The eye, however, is very sensitive to wavelengths in the approximate range 380 nm to 760 nm and can perceive objects which emit or reflect radiation in this range. The eye converts the patterns produced by changes in intensity (brightness) and wavelength (colour) into electrical signals which are then interpreted by the brain.

The eye is surrounded by a tough, fibrous coating, called the sclera, which is transparent at the front to form the cornea but opaque elsewhere (see **Figure 1**). The sclera is lined with a layer of tissue, called the choroid, containing a network of blood vessels supplying food and oxygen to the eye. It also contains black pigmentation which reduces the reflection of light within the eye and prevents blurring of the image. The thickened edge of the choroid in the region round the lens is called the ciliary body and contains circular muscle fibres, the ciliary muscles, responsible for altering the shape of the lens. Inside the choroid, at the back of the eye, is the retina, the light–sensitive layer, on which light is focused by the lens.

Figure 1: the eye
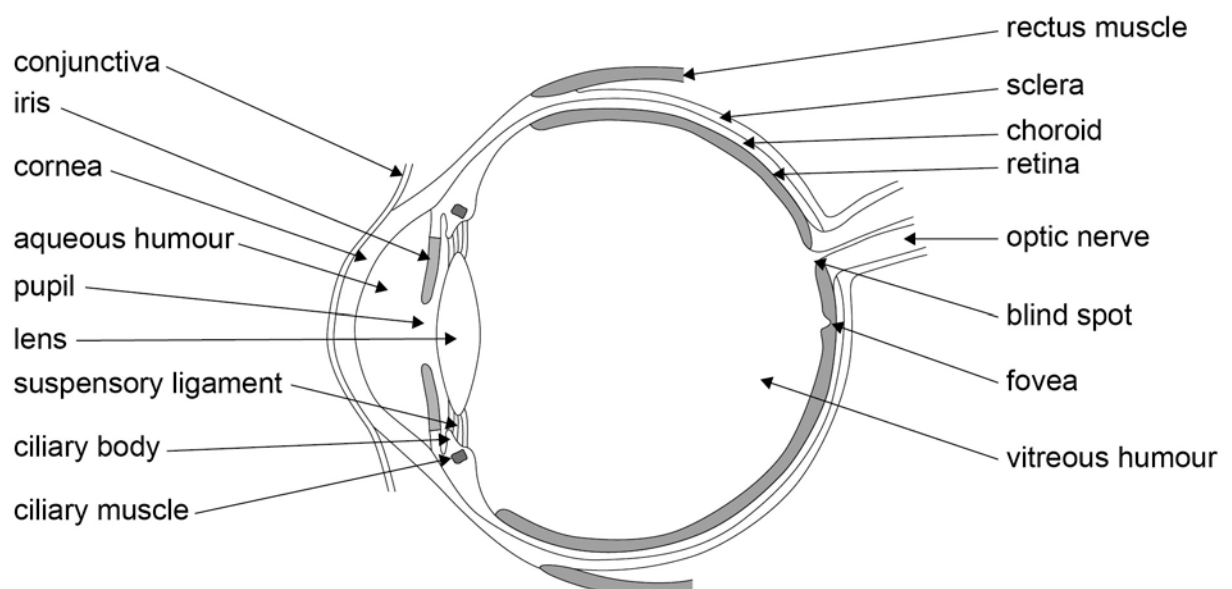
The lens is held in position by a circular band of tough, fibrous material, the suspensory ligament, whose outer rim is attached to the ciliary muscle. Even when this muscle is relaxed there is a degree of tension in the ligament, keeping the lens slightly curved; in this position the lens has maximum focal length, and a normal eye will focus quite clearly objects at distances from 5 metres to infinity. Objects closer than 5 metres are brought into focus by contraction of the ciliary muscle, causing a reduction in

tension in the lens and therefore an increase in its curvature and a consequent decrease in its focal length. This focusing process is referred to as accommodation.

The two chambers into which the eye is divided by the lens are filled with transparent fluid, the gelatinous vitreous humour between lens and retina and the watery aqueous humour between lens and cornea. The pressure of these fluids on the sclera maintains the approximately spherical shape of the eye.

Most of the refraction and image formation occurs when light first enters the eye at the air/cornea interface. The presence of the aqueous and vitreous humours reduces the amount of refraction taking place at the lens, the main purpose of the lens being to adjust the overall focus of the eye, ie to permit accommodation.

The aperture to the lens (pupil), and therefore the intensity of light falling on the retina, is controlled by a coloured extension of the ciliary body known as the iris. The diameter of the pupil decreases when near objects are viewed, thus removing the outermost rays of light which may not be refracted sufficiently to obtain a sharp image; hence the size of the pupil is also used to control spherical aberration.

Figure 2: section through the retina
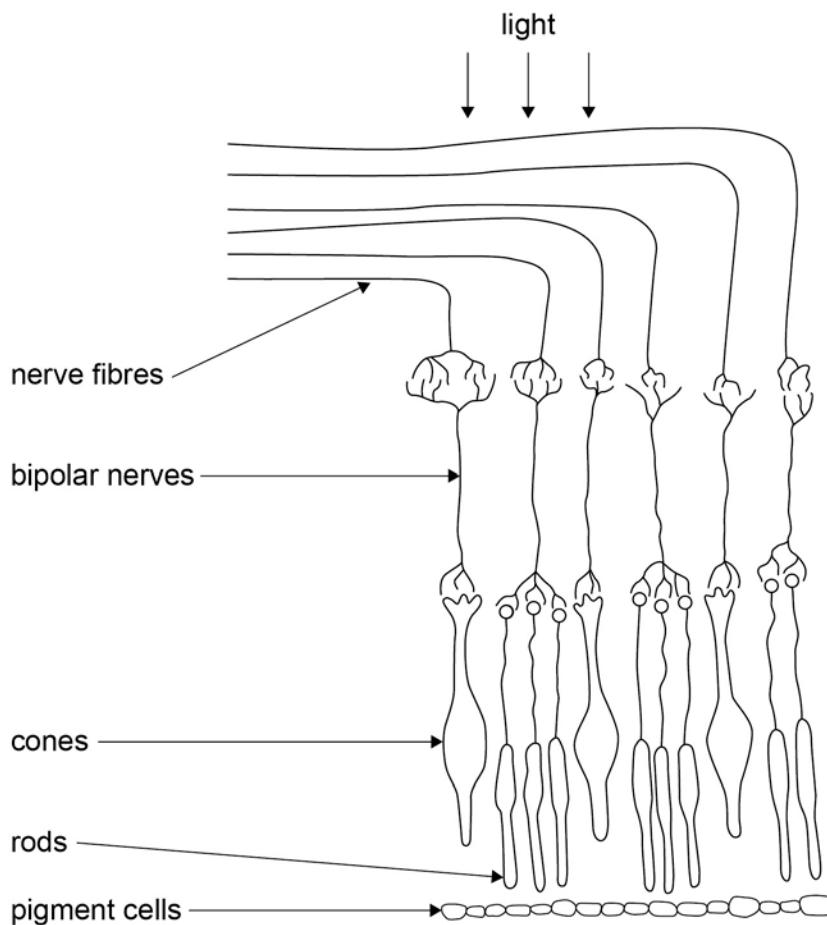
The retina, which when illuminated produces electrical signals, consists of 2 types of light-sensitive cells called rods and cones (see **Figure 2**); there are about 20 times more rods than cones. Rods are used in low-intensity light detection, and several can be joined to one fibre of the optic nerve; thus they give only simple light perception, with gray scale images and little detail. Cones, on the other hand, are sensitive to high intensity light and each is connected directly via one nerve fibre to the brain. Cones give more detailed vision in colour as the cones are also sensitive to different wavelengths within the visible spectrum.

The outer ends of the rods and cones (cylindrical in rods and conical in cones) are very close together and are light-sensitive along their length; hence the eye can focus simultaneously without adjustment of the lens on objects between infinity and about 5 metres, ie without accommodation. The nerve fibres which become the optic nerve lie on the surface of the retina and light must penetrate this layer before activating the rods and cones.

Each photodetector, ie rod and cone, can be thought of as a photoelectric cell because light energy is converted to electrical energy. When light falls on the rods and cones, the associated optic nerve fibres are stimulated electrically, and the electrical pulses are conducted to the brain where they produce the sensation of vision. Rods contain

visual purple (rhodopsin), a compound containing a form of vitamin A and a protein. Light causes this visual purple to bleach and produce small emfs; the bleaching is reversed by enzyme action and at low light-intensities this reversal occurs more rapidly than the bleaching. There is therefore a greater concentration of visual purple at low intensities, and the rods become more sensitive under these conditions. In order to achieve maximum sensitivity at low light-intensities, the visual purple must be allowed time to regenerate completely. This process, called dark adjustment, can take up to 45 minutes. Cones, although known to contain other photopigments, do not contain visual purple. The structure and function of the cone photopigments is not as well understood as that of visual purple.

The ratio of rods to cones is not uniform over the retina. The region directly opposite the pupil, the fovea, consists entirely of slender, close-packed cones, so giving good spatial resolution. As these cones have a yellowish colour this region is also known as the yellow spot. Further away from the fovea the cone density rapidly decreases and the rod density increases, so that the periphery of the retina consists mainly of rods.

When the eye looks directly at an object, the image is formed on the fovea, and, because of the large number of cones, each connected directly to the brain, the object is seen in great detail. Objects which are focused on other parts of the retina are not clearly seen and sometimes are only noticed 'out of the corner of the eye' when they move. In poor light however, because of the larger number of rods, it is these outer parts of the retina which are used; for this reason, objects sometimes apparently disappear when looked at directly in poor light.

Just below the fovea is a region known as the optic papilla or blind spot. This is the area of the retina where nerve fibres from all parts of the retina converge and leave the eye to form the optic nerve. There are no rods or cones in this area and so it is not sensitive to light.

b) Colour vision

Electromagnetic radiation of wavelength approximately 380 nm to 760 nm is detected by the eye as different colours. All these colours can be produced using mixtures of the primary colours, red, green and blue, and so in principle the retina need only be sensitive to these three colours. Modern ideas concerning colour vision suggest that cones contain three types of light-sensitive material, each able to detect one of the primary colours. When light falls on the retina, each of these materials will be affected to an extent depending on the intensity of the primary colour present in the incident light, and each will produce a corresponding stimulus which is passed on to the brain.

The section on the eye also expects knowledge of ray diagrams for both convex and concave lenses which can be found in most GCSE textbooks. The three main defects of vision, myopia or short sight, hypermetropia or long sight and astigmatism should be understood as should their correction using a suitable lens. The students are expected to be able to use suitable equations to find the image distance, the object distance or the focal length/power of a correcting lens given other values for the correction of

either myopia or hypermetropia. The calculations based on a correcting lens may either be done for the lens alone, or might be done considering the eye alone and then the eye and lens combined. In calculations using the correcting lens alone, the object is at the corrected point, is always real and has a positive value for the object distance, whilst the image is at the uncorrected point, is always virtual and has a negative value for the image distance. In the calculations involving the eye lens, it is usual that the eyeball is regarded as having a diameter of $2.0 \, \text{cm}$, which is the then taken as the image distance from the eye lens for a focused image on the retina. For astigmatism, no calculations are expected, but the students should know that the prescription for the correcting lens should state both a power and an axis angle for correction.

Section 2: The ear

The ear is divided into three regions, the outer ear, the middle ear and the inner ear.

a) The outer ear

The outer ear consists of a tube called the external auditory meatus, whose inner end is closed by the tympanic membrane (eardrum) which is about $1 \, \text{cm}$ in diameter. The outer end of the external auditory meatus is surrounded by the pinna (ear flap), located at the side of the head. The walls of the ear canal (external auditory meatus) have many ceruminous or wax glands, whose secretion helps to protect the tympanic membrane and keep it pliable.

b) The middle ear

Figure 3: the middle ear
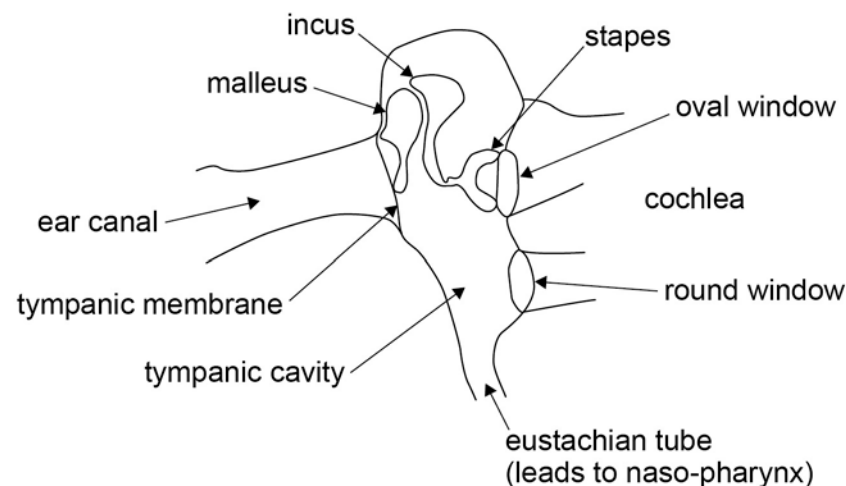
The middle ear, shown in **Figure 3**, consists of the tympanic cavity which is air filled and communicates with the naso–pharynx (throat) through the Eustachian tube. This allows the passage of air when a person swallows to equalise the pressure on the two

sides of the tympanic membrane. The middle ear contains three small bones, hammer (malleus), anvil (incus) and stirrup (stapes), called the auditory ossicles (ear bones). One end of the hammer is attached to the inner surface of the eardrum; its other end is connected by a small joint to the anvil which, in turn, is connected to the stirrup. The stirrup is attached to the oval window (a membrane covered opening leading into the inner ear). A second membrane covered opening, called the round window, also connects the middle ear to the inner ear, thus giving the inner ear two flexible ends which allow movement of the enclosed liquid.

The ear bones rock like levers and transfer the energy of the sound wave across the tympanic cavity. As the energy is transferred, the amplitude of the wave is increased by a factor of about 1.5. The large difference in the areas of the eardrum and the oval window causes the pressure factor to be further increased by a factor of about 15 so that the pressure at the oval window is over 20 times higher than the sound pressure at the eardrum.

c) The inner ear

The inner ear is a complex, bony chamber, containing a membranous labyrinth filled with fluid, made up of the cochlea (hearing organ) and the balance organs. As well as being concerned with orientation of the body, the balance organs, detect changes in velocity, but they play no part in the hearing mechanism. The cochlea is a helical, spiral-shaped cavity within the solid bone. Inside the cochlea are three membranous charnels which run the full length of the cochlea. The arrangement of the channels is best understood by unwinding the spirals, as shown in **Figure 4**.

Figure 4: uncoiled cochlea
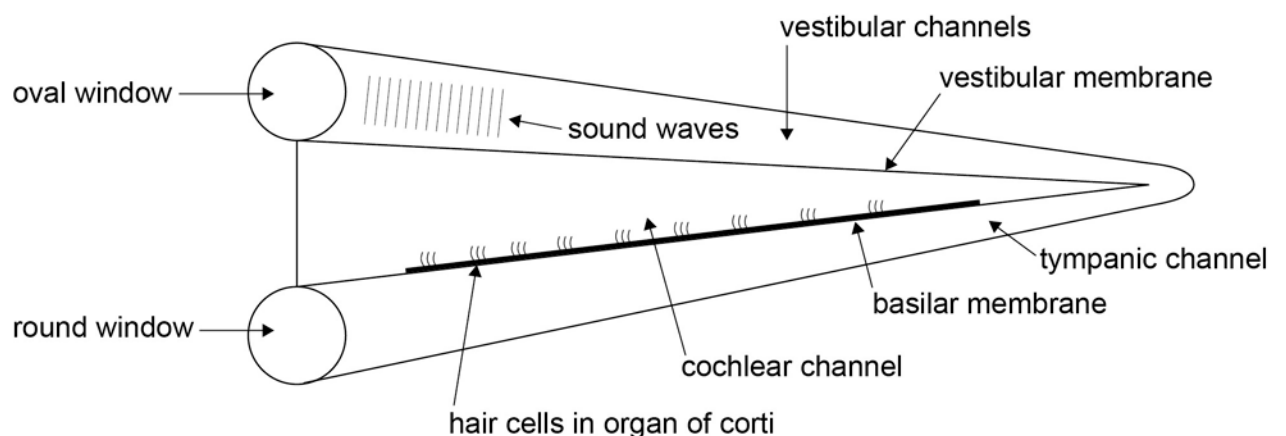
The upper vestibular channel is attached at its base end to the oval window. At its other end, at the apex of the cochlea, the vestibular channel has a small opening which communicates with the lower tympanic channel. The base end of the tympanic channel terminates at the round window. Both the tympanic and vestibular channels are filled

with a clear fluid called perilymph which is derived from cerebrospinal fluid. The third and smallest channel, which rests between the other two, is filled with a clear fluid called endolymph and is separated from the overlying vestibular channel by the vestibular membrane and from the underlying tympanic channel by bone and the basilar membrane. The receptors for hearing are several rows of hair cells in the middle channel which contain numerous cilia projecting from the free end of each cell into the endolymph. The rows of hair cells, together with the supporting cells, constitute the organ of Corti which rests on the basilar membrane in the cochlear channel; there are about 25 000 hair cells.

d) Mechanism of hearing

Sound pressure waves in the air, impinging on the eardrum cause it to vibrate. This causes the ossicles to vibrate, vibrating the oval window; pressure waves are set up and transmitted through the fluid in the coclea, up the vestibular channel in the perilymph. These vibrations cause displacement of the basilar membrane, producing a shearing action of the cilia which results in nervous activity. The physical properties of the cochlea are such that different frequencies tend to act at different points along the basilar membrane. High-frequency waves act at the base of the cochlea and low-frequency waves act at the apex of the cochlea, ie high frequency waves do not penetrate as far down the cochlea as those of lower frequencies. The distribution of nervous activity thus depends on the frequency content of the sound and also on the amount of sound energy received, since the response of each cell varies with the size of the disturbance suffered by the basilar membrane.

e) Sensitivity and logarithmic response

The sensory response to sound signals is not one of simple proportionality: the perceived loudness is not linearly dependent on the signal intensity, but is approximately logarithmic. Intensity is the quantity of energy arriving at the ear per unit area per unit time, or the power per unit area arriving at the ear. The unit of intensity is $W\ m^{-2}$. The average human ear can just detect a sound of intensity about $1\ pW\ m^{-2}$ at 1 kHz; this is known as the threshold intensity, I_0. The maximum intensity that the ear can withstand is about $1\ W\ m^{-2}$. Since this is a range of 10^{12} and since the ear is logarithmic in its response, it is convenient to use a logarithmic scale, for sound intensity. The unit used is the bel.

For an incident intensity, I_1.

Relative intensity in $\text{bels} = \log_{10} \dfrac{I_1}{I_0}$

One bel is defined as the logarithm to base 10 of the ratio of two sound intensities, with relative intensities in the ratio 10:1.

In practice $I_0 = 1\ pW\ m^{-2}$ is taken as the reference level and relative intensities referred to this level may be called 'the sound intensity level referred to $1\ pW\ m^{-2}$.

For convenience, a smaller unit, the decibel (dB), is used, therefore:

relative intensity level = $10 \log_{10} \dfrac{I_1}{I_0}$ dB.

Hence an increase in intensity level of 1 dB corresponds to a multiplication of the sound intensity by about 1.26. A doubling of the sound intensity corresponds to the relative intensity level increasing by 3.01 dB. That is,

relative intensity level = $10 \log_{10}^{-2} = 3.01$ dB

If the sound intensity increases by 100 times, the relative intensity level accordingly increases by 20 dB.

The approximate relative intensity levels from various sources are given in **Table 1**.

Table 1

| description of sound	relative intensity level / dB	sound intensity / W m^{-2}
threshold of hearing	0	10^{-12}
whispering	20	10^{-10}
background music	40	10^{-8}
conversation	60	10^{-6}
busy street	70	10^{-5}
jet plane overhead	100	10^{-2}
thunder overhead	110	10^{-1}
threshold of pain	120	1

f) Sensitivity

The ear is not uniformly sensitive over the entire hearing range, as is shown in **Figure 5**. The ear is very sensitive in the region from 2 kHz to 5 kHz, being most sensitive at 3 kHz. The solid line in **Figure 5** shows the threshold of hearing for a young person with good hearing; the dotted line shows the average threshold for all ages. It can also be seen that average hearing requires a relative intensity level of 30 dB or more to detect a sound of 100 Hz than to detect one at 1 kHz. The sensitivity of the ear changes with age: as a person gets older, the highest frequency which can be heard decreases and the relative intensity level also needs to be higher for a person to detect the higher frequency levels. For example, a person aged 45 years very often cannot hear frequencies above 12 kHz and, in relation to a 20-year-old, the 45-year-old person also needs about 10 dB increase in relative intensity level in order to hear a

4 kHz sound. Usually, by the age of 65, a 25 dB loss in sensitivity in the frequencies above 2 kHz has occurred. Hearing may deteriorate more rapidly if the ears are subjected to continuous loud sounds; people who are continually exposed to loud music or who work in noisy factories can suffer serious hearing losses. Such hearing loss is a maximum at 4 kHz.

If the ear were as sensitive at low frequencies as it is in the 3 kHz region, we would be aware of many physiological noises, such as blood flow and movement of joints.
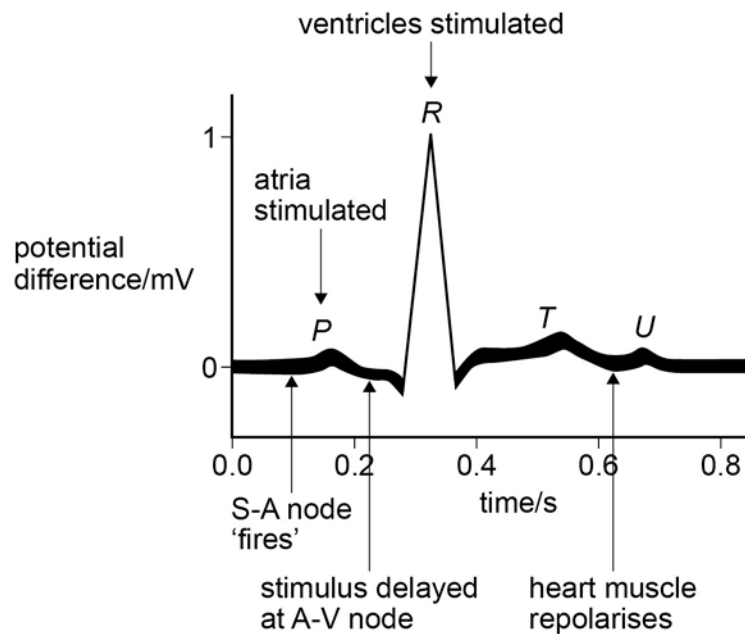
Figure 5
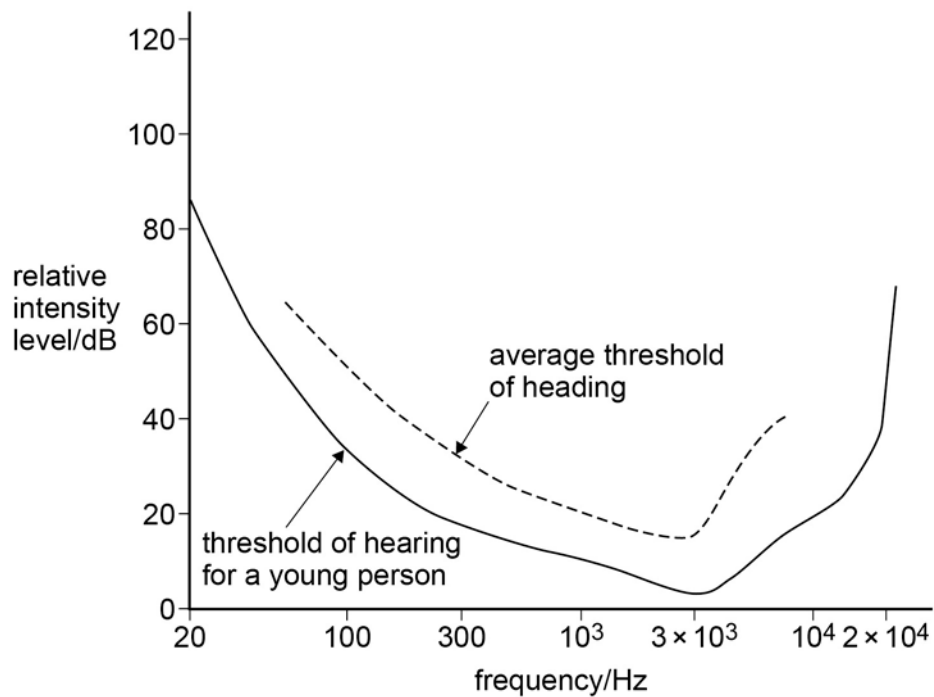
The curves shown above, equal loudness curves, can be produced for a person and are known as audiograms. The hearing of the person is tested through ear phones in a soundproofed room. Students will be expected to have an understanding of this procedure.

Section 3: Biomedical measurement

a) Simple ECG machines and the normal ECG waveform

Figure 6: typical ECG

The study of this cardiac waveform is known as electrocardiography (ECG); the machines used are known as electrocardiographs which produce electrocardiograms. In addition to the arms, electrocardiographers use other standard electrode sites which allow observation of the heart's electrical activity to be made. These standard sites are connected in pairs to the two terminals of an ECG amplifier to give 12 alternative connection systems (known briefly as 'leads').

right arm	and left arm	lead I
right arm	and left leg	lead II
left arm	and left leg	lead III
right arm	and left arm and leg	aVR
left arm	and left leg and right arm	aVL
left leg	and light and left arm	aVF
one of six chest sites	and both arms and left leg	V_1 V_2 to V_6

Bio-electrical signals are detected at the body's surface by metal plate electrodes held in position by adhesive, rubber straps or suction. To reduce the impedance of the electrode-skin interface, which with bare metal and dry skin can be as high as $1\ \text{M}\Omega$, the skin is first rubbed with abrasive paper to remove unwanted hairs and some dead

skin, and a conducting electrode gel is used between the metal electrode and the skin. This contains a strong electrolyte in a non-irritant base gel, eg glycerol, and can lower the impedance to around $1 \text{ k}\Omega$. The electrodes should not react with the chemicals produced by the skin.

In order to ensure good transfer of the signal voltage to the amplifier, the amplifier should have a high input impedance. Unwanted signals, ie noise, which can be of biological or electrical origin, may have the same order of magnitude as the signal itself. If the patient is tense or uncomfortable, for example, muscle tremors are detected in an ECG, so the patient should be as relaxed and as comfortable as possible. Electrical noise has many origins, the most common being nearby power cables and electrical equipment which induces charge on the patient. Noise caused in this manner will be of the same magnitude in each connection; thus, if a differential amplifier is used, the output will be proportional to the difference of the potentials at each input so that the noise will be much reduced if not eliminated completely. For safety reasons the amplifier power connections should be electrically isolated from the patient.

Section 4: Non-ionising imaging

a) Optical fibres and endoscopes

If light enters a glass rod so that the refracted light is incident on the side of the rod at an angle greater than the critical angle, then total internal reflection occurs. If light continues to be internally reflected, the light emerges at the other end with very little loss of intensity.

If the rod is drawn out into a long fibre, light is transmitted along its length with little loss of intensity. Such a fibre, or bundle of fibres, provides a convenient method of conveying light from one place to another. Each fibre, consisting of a long, thin cylinder of transparent material, transmits light even when the fibres are curved. When two or more fibres have sidewalls in contact over some distance, the light can leak from individual fibres into neighbouring ones. Leakage also occurs if the walls of the fibres are not clean. In order to avoid these losses, it is usual to clad the glass fibre in a glass skin which has a lower refractive index than that of the fibre. The cladding protects the fibre surface and isolates it from its neighbours.

Figure 7: clad fibre
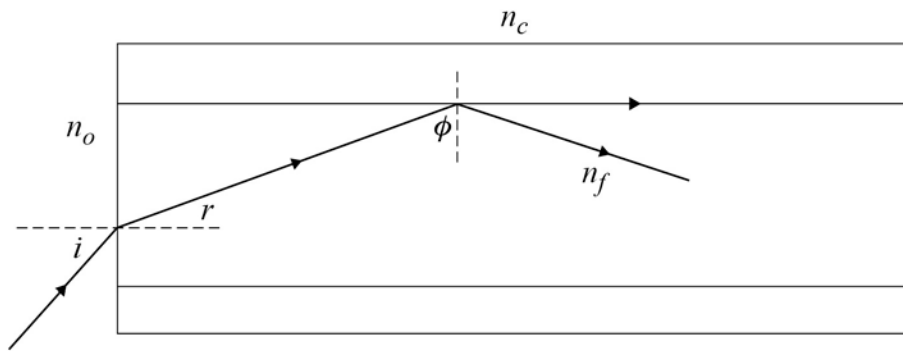
$$\phi = \left(\frac{\pi}{2} - r\right)$$

n_o = refractive index of cladding
n_f = refractive index of fibre
n_o = refractive index of material outside the fibre

If the fibre is curved, the angles of incidence are changed as the light travels down the fibre and losses occur if the angles fall below the critical angle. More curvature can be obtained at the expense of the numerical aperture. A radius of curvature down to about twenty times the fibre diameter can be tolerated in practice.

One bundle of fibres is used to illuminate the area under observation whilst another bundle of fibres, together with lenses, is used to convey the image back to the eyepiece. Tools for holding and cutting tissue, wires for performing diathermy and an irrigation port may be included in the instrument.

If a bundle of fibres is placed together in an orderly manner along its length with, the ends cut square and polished smooth; it is possible to transmit images along the bundle as shown in **Figure 8**.

Figure 8: coherent bundle of fibres
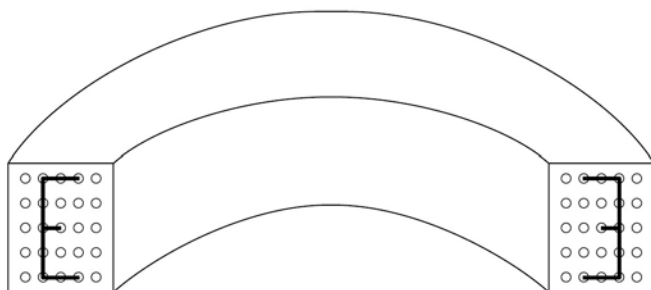
The image falls on a certain number of fibres and each fibre receives light forming a small part of the image. The fibres each transmit a small part of the image and it is

possible to see the transmitted image at the opposite end of such a 'coherent bundle' of fibres.

In an 'incoherent bundle', no attention has been paid to the positioning of the fibres and such bundles are suitable only for the transport of light into the body to illuminate the area under investigation.

The definition in the transmitted image depends on the quality of the fibre bundle. If tightly packed fibres with very small diameters are used, the spaces between the fibres are small; hence the grain of the final image is small. The diameter of the individual fibre in good-quality coherent bundles can be of the order of 0.012 mm and the light lost per internal reflection as low as $4 \times 10^{-5}\%$.

Fibre bundles are often used in medicine where it is required to make observations at points within the body. Any instrument for direct viewing inside the body is called an endoscope, and a fibre-optic, flexible version is illustrated in **Figure 9**.

Figure 9: the endoscope together with details of distal tip and some tools which are used
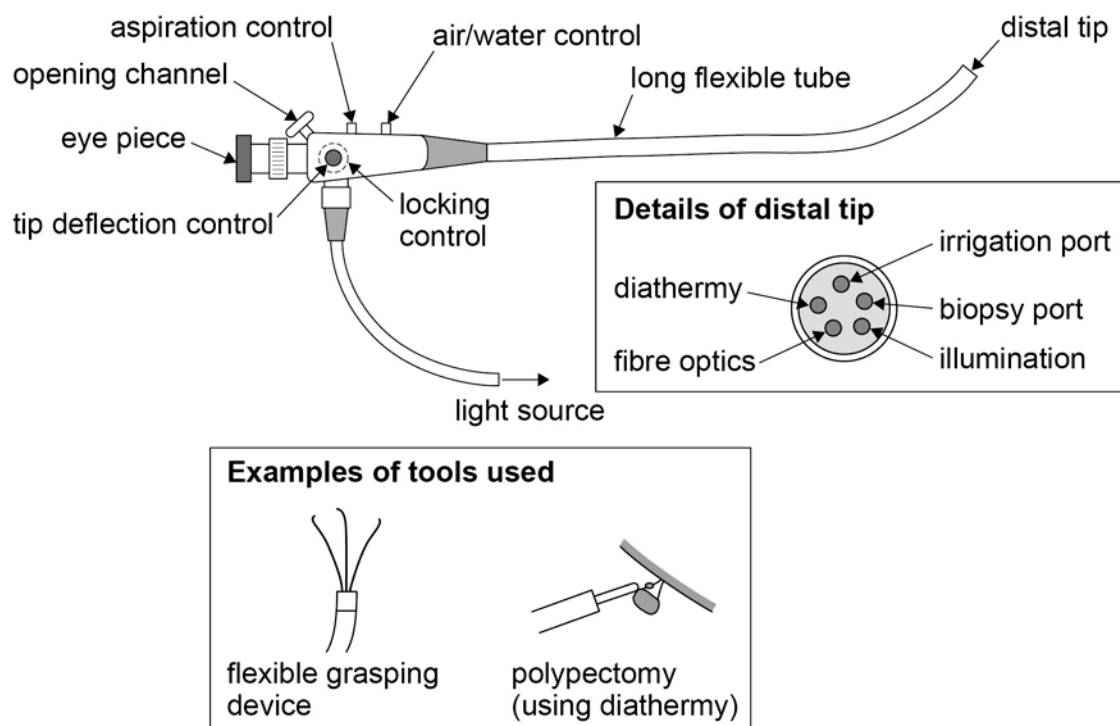
b) Ultrasonics

Normal sound waves consist of longitudinal pressure waves in a medium. The human ear is sensitive to waves of this type in the frequency range from about 20 Hz up to $15 / 20 \text{ kHz}$. Ultrasound waves have frequencies above the audible range. Ultrasonic scanning is the use of high-frequency sound waves in diagnosis; frequencies between 1 MHz and 5 MHz, for example, are used in obstetrics. These waves have a wavelength

of the order of $1\ \mathrm{mm}$ which limits the resolution. The ultrasound technique is non-invasive and has not yet been shown to produce any side effects.

c) Production of ultrasound waves

One technique used for the generation of ultrasound waves involves the piezoelectric effect, shown by certain crystals such as quartz. These crystals produce mechanical deformation (ie changes in dimensions) when a potential difference is applied across them. Conversely, when the crystal is mechanically caused to expand or contract, a potential difference is produced across it. The device used to produce ultrasound is called an ultrasonic transducer and is illustrated in **Figure 10**.

Figure 10: transducer
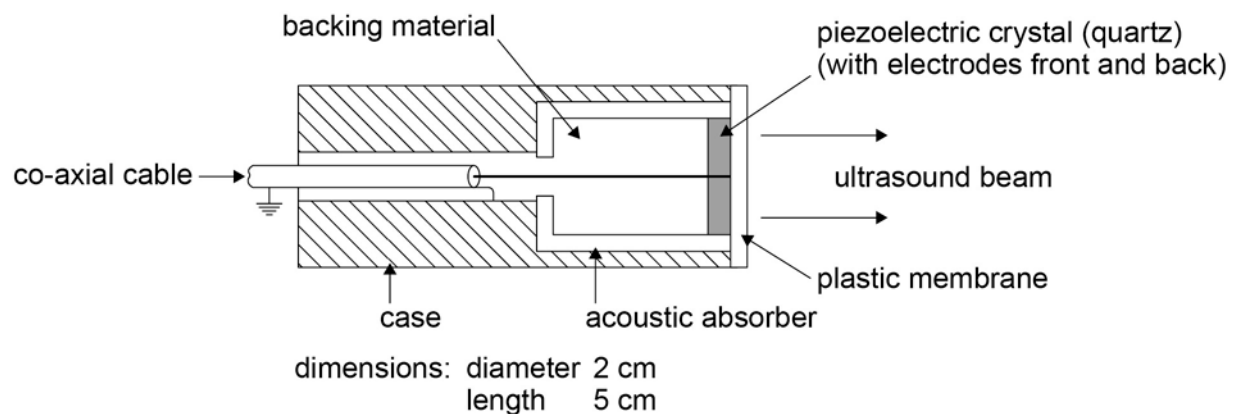
When an alternating pd is applied to the transducer, periodic changes in stress occur and forced vibrations are excited which have the same frequency as the applied pd. The vibrations have maximum amplitude when this frequency is equal to one of the natural frequencies of vibration of the crystal. A short-voltage pulse $(600\ \mathrm{V})$ causes the crystal to vibrate (ie to expand and contract about its mean position). The crystal then sends out a pulse of ultrasound waves into the medium with which it is coupled. Similarly, if ultrasound waves hit the crystal face, the crystal becomes deformed, some of the sound energy will be converted into electrical energy and a pd will appear across the crystal. The pd is amplified and displayed on the cathode ray tube. Clear distinct echoes are obtained by using short pulses. To generate short pulses, the vibrations must be damped out as quickly as possible by bonding a damping material (epoxy resin) to the back face of the crystal. The function of the transducer is two-fold: to generate the pulse of ultrasound and then to act as a detector of the returning echoes.

d) Reflection of ultrasound

Reflection of ultrasound takes place when an ultrasonic pulse crosses an interface between two media and the size of the echo produced at such an interface depends on the difference of the characteristic acoustic impedances of the two materials. Acoustic

impedance is defined as the product of density and the velocity of the sound in that material. The difference in acoustic impedances of two soft tissues is small and only about one per cent of the energy of the ultrasonic beam is reflected. However, this is sufficient to produce a detectable echo. More intense echoes are produced at 'bone–soft tissue' boundaries since the impedance difference is large. Similarly, an 'air–soft tissue' boundary reflects almost all of the incident pulse because of the large impedance differences. For this reason it is essential to use a coupling medium to exclude air at the point where the transducer makes contact with the skin. It should be noted that the presence of air within the body (ie in the lungs) prevents structures behind it from being examined. Various characteristic impedances are given in Table 2.

Table 2

material	velocity / $\mathrm{m\ s^{-1}}$	density / $\mathrm{kg\ m^{-3}}$	acoustic impedance / $\mathrm{kg\ m^{-2}\ s^{-1}} \times 10^6$
air	330	1.3	4.29×10^{-4}
oil	1500	950	1.42
perspex	2680	1200	3.22
water	1500	1000	1.50
bone	4080	1908	7.78
brain	1540	1025	1.58
muscle	1580	1075	1.70
liver	1585	1042	1.65
fat	1450	952	1.38

It can be shown that the intensity reflection coefficient is

$$\frac{I_r}{I_i} = \left(\frac{Z_2 - Z_1}{Z_2 + Z_1} \right)^2$$

where

I_i = incident intensity in medium 1

I_r = reflected intensity

Z_1 = acoustic impedance of medium 1

Z_2 = acoustic impedance of medium 2

Note that if $Z_1 = Z_2$ there is no reflection and if $Z_1 \ll Z_2$ most of the incident energy is reflected, hence the lack of penetration beyond the lung.

e) Methods of scanning tissues

A-scan

The transducer, which emits short, sharp pulses of ultrasound of a few microseconds duration, is placed on the surface of the body under examination. At the same time as a pulse of ultrasound is emitted by the transducer, the spot on a cathode ray oscilloscope screen is triggered and starts moving across the screen. The ultrasound pulse travels into the body and, when it reaches an interface, some of the sound energy is reflected back towards the transducer where it causes an electrical signal to be generated. The signal, after suitable processing, is applied to the Y-plates of the oscilloscope where it causes the spot on the screen to be deflected vertically for the duration of the received echo pulse. The spot then returns to its original level and continues to move across the screen. The returning pulse is displayed as a deflection of the trace on the screen of the oscilloscope, as shown in **Figure 11**. Since the pulse was emitted at the same time as the spot started to move across the screen, the displacement of the echo from the starting point is a measure of the time taken for the transmitted pulse to reach the first surface plus the time for the echo to return to the transducer. If the velocity of sound in tissue is known, the depth of the reflecting surface can then be calculated and read directly on suitably calibrated instruments. Further echoes from other surfaces in the body produce similar deflections on the oscilloscope screen.

When a pulse of ultrasound passes through the body, the amplitude of the pulse is attenuated, so that echoes from surfaces further away from the transducer require to be amplified more than those from surfaces nearer the transducer. This amplification is achieved by a swept-gain amplifier linked to the time base of the oscilloscope.

Figure 11: production of A-scan
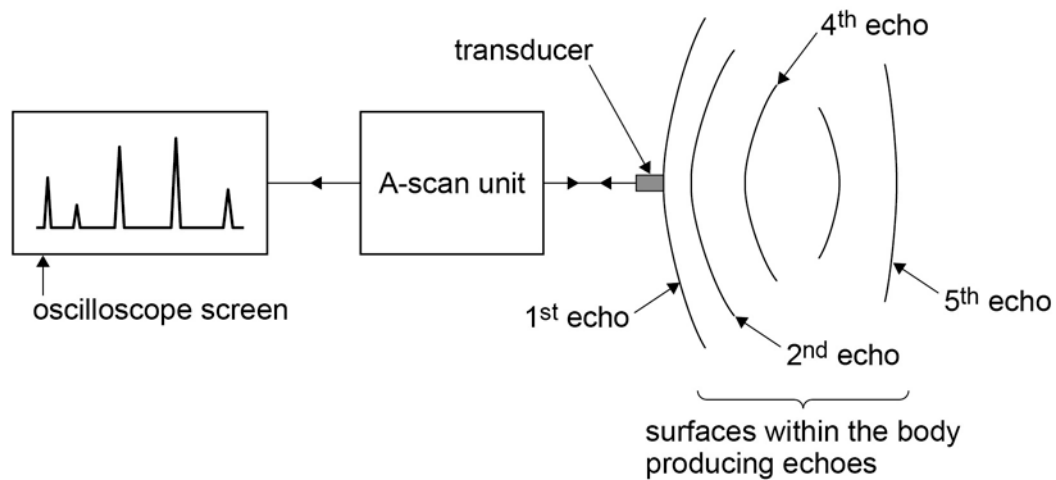
The A-scan method is used to measure the bi-parietal diameter (size of the foetal head) (see **Figure 12**) which may be used in order to determine the length of gestation and also to monitor the growth of the foetus.
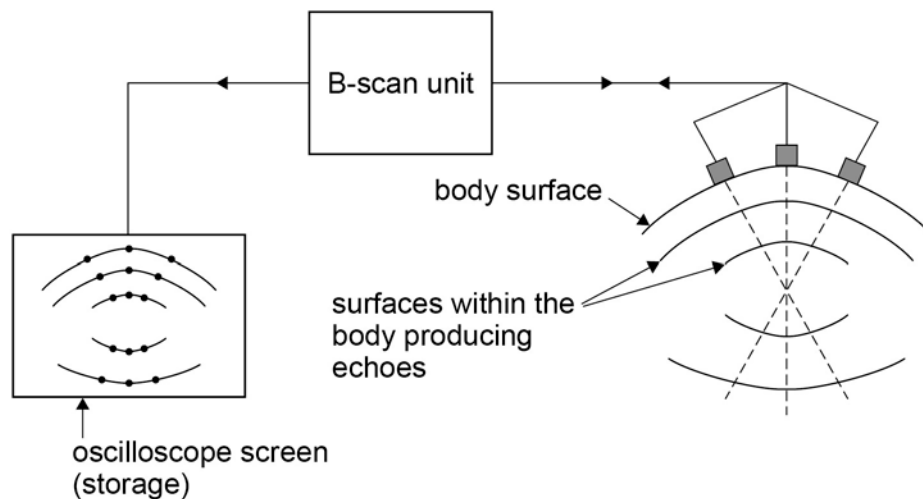
Figure 12: A-scan of foetal head
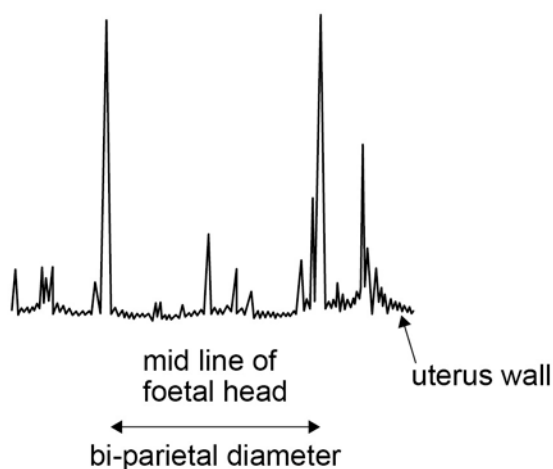
B-scan

The A-scan technique is used where the surfaces of interest within the body lie along a particular line. For more complex structures, however, it becomes very difficult to identify those surfaces which give rise to the echoes for an A-scan.

Figure 13: production of B-scan

Ultrasound techniques (B-scan) have been developed which outline structures by producing images of sections through the body. In the B-scan the echo signals are not applied to the Y-plates of the oscilloscope but are used to control the brightness of the spot on the oscilloscope screen. Voltages from potentiometers which record the position and orientations of the transducer: are fed to the X- and Y-plates and determine the position of the spot on the screen. The co-ordinates must be accurate, and in a B-scan the error in the position of the spot on the screen is within 0.15 mm. With the B-scan a storage oscilloscope is used (**Figure 13**).

It should be noted that echoes are received only if the interface is normal to the beam of ultrasound. The interfaces inside the body have various contours; hence, in order to obtain the best information about an interface, the probe may need to be rocked while it is moved on the surface of the body. This rocking is a skilful operation since the probe must be kept in contact with the surface; otherwise any air space will reflect the ultrasound before it can enter the body. In order to ease the movement of the probe and to provide the required coupling between the transducer and the surface, oil is smeared on the patient's skin.

The largest diameter of the foetal head perpendicular to the midline is the bi-parietal diameter and is measured on the A-scan. In order to select the correct place for the A-scan, a B-scan is used to search for an echo from the midline structure of the brain. Such an echo produces a sharp line bisecting the image of the foetal skull.

An important use of the B-scan in obstetrics is to determine the position of the placenta in relation to the entrance to the cervix. The condition called placenta praevia, ie when the placenta covers the entrance to the cervix, can then be diagnosed and the baby delivered by Caesarean section. A further use of the B-scan is prior to and during amniocentesis, which involves inserting a tube into the uterus to remove some fluid and cells surrounding the foetus in order to carry out a genetic test on the chromosomes of the cells of the foetus. The positions of the foetus and placenta have to be located accurately, which is done by the B-scan.

Section 5: X-ray imaging

a) Production of X-rays

X-rays for diagnostic use are produced in a highly evacuated tube by accelerating a stream of electrons, produced by thermionic emission from a tungsten filament, through potential differences between about $20\,\text{kV}$ and $120\,\text{kV}$ and then allowing the electrons to strike a metal target (see **Figure 15**). The resultant X-rays, caused by the release of energy as the incident electrons slow down in the target, have a continuous spectrum of energies (see **Figure 14**) with a maximum energy depending upon the pd between the anode and cathode.

Figure 14: a typical X-ray spectrum using a tungsten target
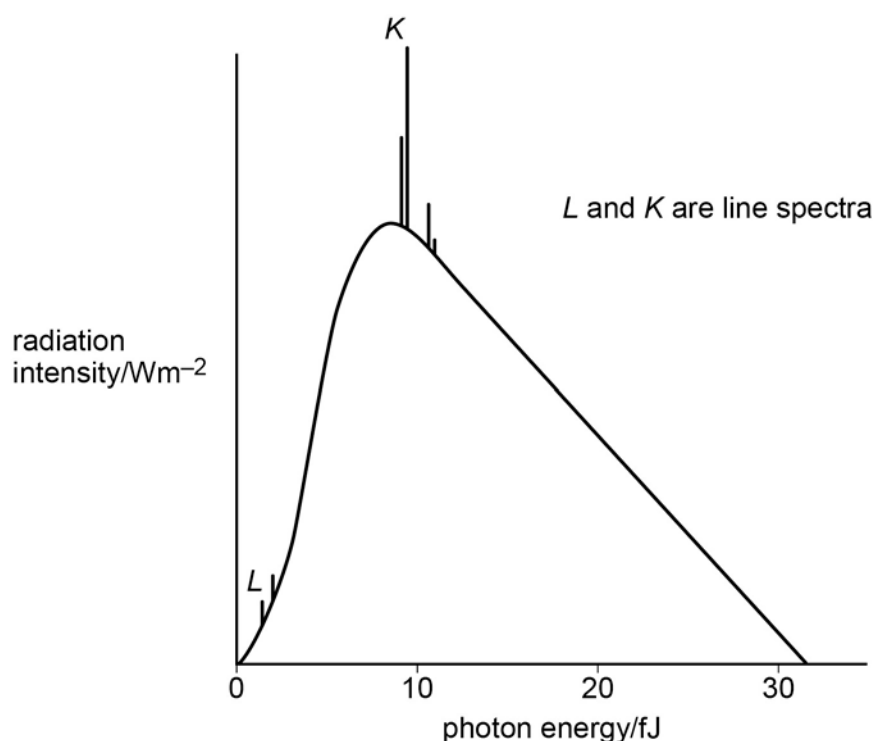
Superimposed on this continuous spectrum is a line spectrum characteristic of the target material. The line spectrum arises from the interaction of incident electrons with individual electrons in the atoms of the target material, where the incident electrons have sufficient energy to remove some inner orbital electrons from the target atoms. The resultant vacancy in the electron shell of the target atom is then filled by an electron from an outer shell with the emission of a photon of radiation. The contribution of the line spectrum to the total radiation output of an X-ray tube depends upon the pd applied, the material and thickness of the tube wall and any filters through which the X-rays pass. Only K radiation is sufficiently energetic to escape from a normal tube; because tungsten is the almost universal target material, there is no

contribution to the useful beam from 'characteristic' radiation for X-ray energies below about $70\,\text{keV}$ ($11\,\text{fJ}$).

The potential difference applied across an X-ray tube not only governs the maximum photon energy but also alters the whole spectral pattern by increasing the peak value of the radiation intensity and moving the peak in the direction of larger photon energy. The positions of the characteristic spectral lines are unaffected by changes in the applied potential difference. The intensity of radiation emitted is approximately proportional to the square of the potential difference applied across the X-ray tube and also increases with increasing tube current, ie as the number of electrons striking the target rises.

b) Rotating anode X-ray tube

A potential difference, between about $20\,\text{kV}$ and $120\,\text{kV}$, is maintained between the anode and cathode (see **Figure 15**). Electrons emitted from the heated filament are attracted towards the positive anode which is bombarded by high-velocity electrons with the result that X-rays are obtained. The anode consists of a disc of tungsten about $70\,\text{mm}$ in diameter and about $6\,\text{mm}$ thick which is rotated at high speed (3000 revolutions per minute).

Figure 15: a rotating anode X-ray tube
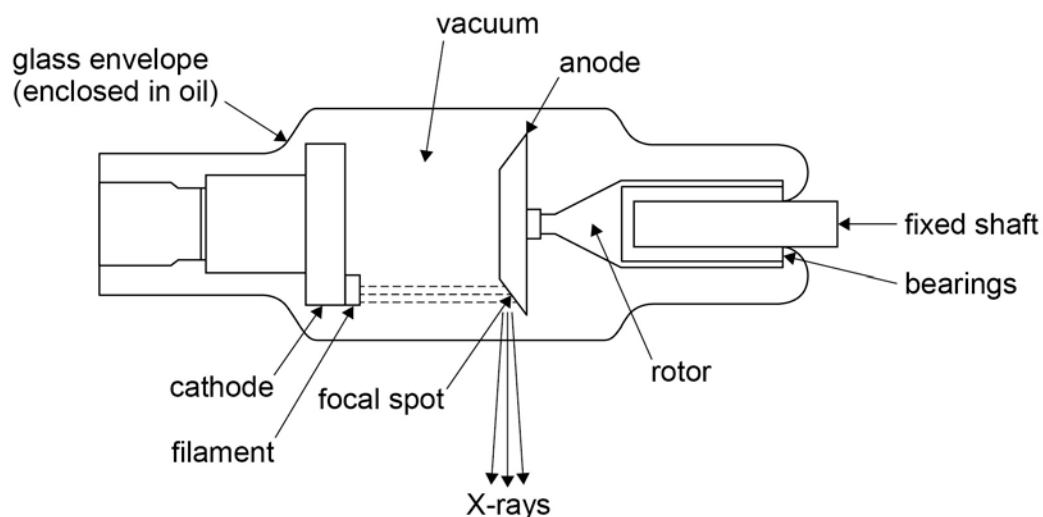
The edge of the tungsten disc is bevelled. This bevelled edge forms the target surface on to which the electron beam is focused. Since the area of tungsten under bombardment is continuously changed because of the rotation, the rise in temperature of the surface of the target at any point is much smaller than would be the case with a stationary target, thus reducing the chance of the surface actually melting. Furthermore, the large mass of the anode allows it to absorb large quantities of energy without excessive rises in temperature. It is therefore possible to use far greater values

of tube current and accelerating voltage without causing an excessive rise in anode surface temperature; hence, much larger X-ray outputs can be achieved using rotating anode tubes instead of stationary targets. The major route of heat loss from the rotating anode is by radiation.

c) Image contrast enhancement

To obtain a clear X-ray image, the difference between the proton number or density of the part of the body being examined and those surrounding it should be large. When this is not the case, a contrast medium might be introduced to the area to improve the clarity of the image. This medium would normally be a compound of a material of high proton number such as barium. The so called 'barium meal' is a thick suspension of barium sulfate which the patient swallows. This passes into the gastrointestinal tract which allows the tract to show up clearly against the surrounding tissues.

d) Absorption of radiation

X-rays carry no charge and there is an element of chance as to whether or not a particular photon interacts with the material. If no interaction takes place, the photon passes through unaffected. When a beam of monoenergetic X-rays passes through matter, its intensity is reduced or attenuated, the amount of absorption depending on the thickness, atomic number and density of the medium. The number of photons absorbed from a beam containing N photons depends directly on N and it is found that:

$$\frac{\text{fractional reduction of intensity in a thin layer}}{\text{thickness of thin layer}} = \text{a constant } (\mu)$$

ie $\mu = \dfrac{\frac{dN}{N}}{dx}$ where x is the thickness of the layer.

(compare this with:

ie $\lambda = \dfrac{\frac{dN}{N}}{dx}$

for radioactive decay)

This constant μ is called the linear attenuation coefficient, usually abbreviated to **linear coefficient**. The mathematical equation which represents the above situation is of the same form as that for radioactive decay. The equation is

$N = N_0 e^{-\mu x}$ where N is the number of photons transmitted by, and N_0 is the number of photons incident on, a thickness x of material. In terms of radiation intensity the equation is

$I = I_0 e^{-\mu x}$ where I is the transmitted intensity and I_0 is the incident intensity.

The density of a given attenuating medium is one of the factors which determines the value of μ, for photons of a given energy, because attenuation is primarily due to interactions between photons and the electrons of the medium. The more atoms per unit volume, the more interactions occur; hence, the attenuation is greater. Therefore, high-density materials have greater values of μ than low-density materials. The dependence of attenuation on density is particularly important in radiography because it allows tissues to be differentiated, even if the atomic numbers of their constituents are similar. There is little difference between the atomic numbers of the elements constituting renal and fatty tissue but, because the density of the kidney is greater, the kidney can be differentiated from the superficial layer of fat surrounding it.

Since the value of y is proportional to the density p of the material for a given photon energy, the quotient μ/ρ is a constant known as the mass attenuation coefficient, μ_m.

For aluminium, the value of μ is 33 m^{-1} for photons of energy 200 keV (32 fJ) and the density is 2700 kg m^{-3}, hence, the value of the mass attenuation coefficient, μ_m, is given by

$$\mu_m = \frac{\mu}{\rho} = \frac{33 \text{ m}^{-1}}{2700 \text{ kg m}^{-3}} = 0.012 \text{ m}^2 \text{ kg}^{-1}$$

One of the consequences of exponential attenuation is that it is theoretically impossible to reduce the intensity of an X-ray beam to zero, no matter how thick the material. The penetrating power or quality of a monoenergetic beam of X-rays can be defined in terms of a thickness of material called the half-value thickness, $x\frac{1}{2}$, ie the thickness of a given material which will reduce the intensity of a narrow monoenergetic beam of X-rays to half its original value.

$\mu, = (\ln 2) / x\frac{1}{2}$ and if μ for the material is known, $x\frac{1}{2}$ can be found.

e) X-ray filters

The radiation emitted by an X-ray tube is made up of photons of many different energies. The maximum energy in the spectrum depends on the pd applied across the tube, whereas the minimum energy present depends on the nature and thickness of the wall of the X-ray tube. A typical spectrum is shown in **Figure 14**.

The photons in a beam of this kind have different penetrating powers because of their different energies. Photons of energy 20 keV (3.2 fJ) and 150 keV (24 fJ) are present in approximately equal numbers. The linear attenuation coefficient in soft tissue is 79 m^{-1} for photons of energy 20 keV and 15 m^{-1} for photons of energy 150 keV, so that about 0.04% of the low-energy photons will penetrate through 0.1 m of material compared with 22 per cent of the high-energy photons (these values are calculated by using the equation $I = I_0 e^{-\mu x}$). It should also be noted that as many low-energy photons are

removed by 0.02 m of tissue as would be lost from the same number of high-energy photons incident on 0.1 m of tissue.

The above points are very important when considering the effect of a beam of X-rays incident upon a patient. Large numbers of the low-energy photons are absorbed in the superficial layers of tissue and very few of them reach the deeper parts of the body, or emerge from the other side of it. Since it is often those parts deep inside the body which are treated by radiotherapy, whereas it is the rays which emerge from the body which produce the radiograph, these low-energy radiations serve no useful purpose at all, but, worse, produce harmful effects in the tissue where they are absorbed. These protons are therefore reduced in intensity by choice of appropriate filters before they reach the patient. The resultant beam is therefore composed of a large fraction of high-energy photons and the decrease in intensity as it passes through tissue is not as great as before. The beam is then said to be 'harder', ie it has greater penetrating power.

An ideal filter would remove the unwanted radiation and leave the required radiation as shown in **Figure 16**. In practice, however, it is found that when a filter is introduced it affects photons of all energies, as shown in **Figure 17**.
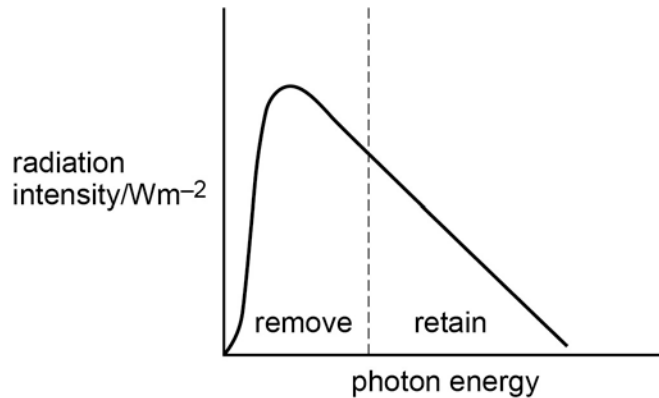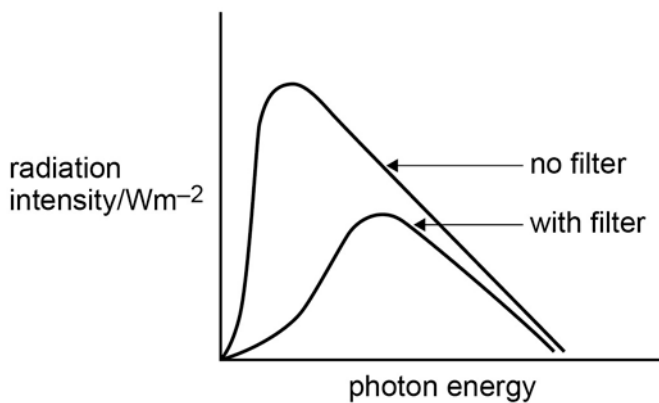
Figure 16: an ideal filter

Figure 17: a practical filter

When a filter is introduced into an X-ray beam it has two effects: first, it removes unwanted radiation and so hardens the beam or increases its penetrating power; second, it reduces the beam intensity. The filter is used to remove the unwanted low-energy radiation as efficiently as possible whilst having the smallest possible effect on the required higher energy photons. The filter must discriminate against the low-energy photons. This is precisely the effect of attenuation by the photoelectric process, the probability of a photon interacting by this effect being inversely proportional to the cube of the photon energy, whereas the scattering probability varies inversely with the photon energy. Hence, a material in which the photoelectric effect predominates is required, ie a material of high atomic number (Z) since the photoelectric effect varies with Z^3. Copper is commonly used as a filter in the therapy range, ie above $250\,\mathrm{keV}$ $(40\,\mathrm{fJ})$, whilst aluminium is used in the radiography range, ie less than $150\,\mathrm{keV}$ $(24\,\mathrm{fJ})$.

f) Factors affecting the sharpness and contrast of a radiographic image

X-rays originate from all points within a small region of the target, known as the focal spot, and the cone of rays from each point forms its own shadow of object **AB** as shown in **Figure 18**.

Over the region **QR** very few X-rays reach the film and this is the region of full shadow (umbra). To the left of **P** and to the right of S-rays reach the film from all parts of the focal spot so the film is blackened.
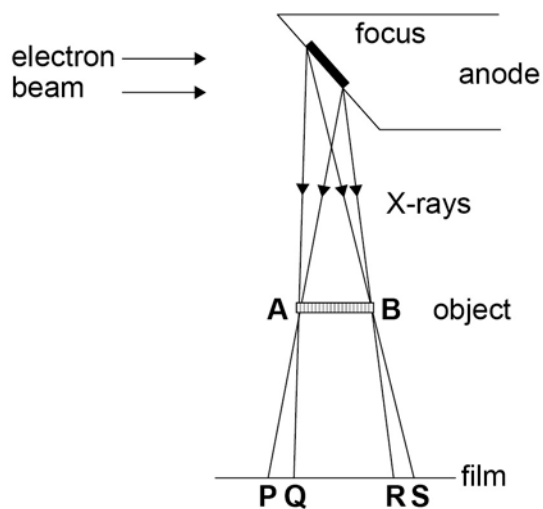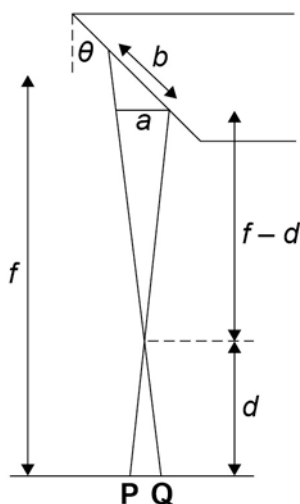
Figure 18: geometric unsharpness

Figure 19: diagram for determination of geometric unsharpness

f = distance of film from centre of focal spot
d = distance of film from object

From **P** to **Q** and from **R** to **S**, X-rays reach the film only from some points of the focal spot so that the film, being in partial shadow (penumbra), is blackened to a lesser extent. The image of **AB** on the film consists of a transparent central region **QR** surrounded by a fuzzy edge **PQ** and **RS**, which is called the geometric unsharpness (U_g). To minimise geometric unsharpness, the values of α and d should be small and f should be large. Angle θ is typically about $17°$ for diagnostic tubes and the effective focal spot size, α, has a diameter of about 1 mm; f is about 1 m for most radiography work, and the distance d of the film from the object depends upon the position of the structure undergoing an X-ray examination. The X-ray film is placed in a cassette immediately below the patient.

There is a limit to the reduction of focal-spot size; too small a focal spot causes an excessive concentration of heat in the target. The size of the focal spot is determined by the value of the pd between filament and target, the tube current and the exposure time, all of which govern the production of heat in the target.

The pd across the tube controls the maximum energy of the X-ray beam and, therefore, the difference in optical density on the radiograph of objects within the body having different physical densities or atomic numbers. The greater the difference in optical density, or 'contrast', the more easily an observer can visualise on the radiograph body organs of different composition. The exposure time governs the density or blackening of the film, but the exposure time may itself be limited by the amount of movement of the structure being examined. The hand, for example, can be completely immobilised and the exposure time can be quite long, of the order of a few seconds, but for the stomach, where there are involuntary movements, the exposure time is short, about 0.5 s. Movement produces further blurring of the image which is known as movement unsharpness (U_m). In an ideal situation the voltage and exposure would be selected and then the tube current adjusted in order to give the required density on the film. Very often, however, the required current would exceed the tube rating, so that the size of the focal spot would have to be increased to allow for the required current. This increases the geometric unsharpness; hence, in order to obtain the best radiograph, a compromise has to be reached.

g) Beam size incident on patient

In order to limit the radiation received by the patient, the X-ray beam is limited by a diaphragm which consists of two movable pairs of metal sheets, one pair at right angles to the other, so that a rectangular beam is produced. The area of the beam at the film depends upon the aperture at the diaphragm and the distances of the diaphragm and film from the target. The knobs which control the diaphragm are usually calibrated so as to show the size of the beam at the film for several values of distance of the film from the target. Since the diaphragm is close to the target, the edge of the beam is not very sharp. In diagnostic work, this is of no importance, but in therapeutic work a sharp edge is desirable and special collimators are used. The diaphragm system usually incorporates a lamp and mirror positioned so that the light beam which shines on the patient is in the same direction and of the same size as the X-ray beam.

h) Scattered radiation

Not all the photons removed from the X-ray beam are absorbed by the patient: many photons are scattered, so that some of the incident photons are replaced by others travelling in different directions and with reduced energy. The film is therefore irradiated by two groups of photons. The first group is the primary photons which have passed through the patient and carries the pattern on to the film; the second group is the scattered photons which have been produced in the patient, mainly as a result of Compton scattering. The second group is generally patternless and irradiates the whole area of the film more or less uniformly. The effect of this scattered radiation on the film is to reduce the contrast.

A grid is used in the majority of radiographic techniques in order to limit the scattered radiation reaching the film (see **Figure 20**), especially when a large volume of tissue is being examined. A small amount of primary radiation is absorbed by the grid; hence, in order to compensate for this loss and the loss by scattering, an increase in exposure is used to maintain a suitable film density. The grid is composed of a large number of long parallel strips of lead which are held apart by an interspace material transparent to X-rays and surrounded by a metal outer casing. The grids normally have about two or three strips of lead per mm and the strips are about $0.05 \, \mathrm{mm}$ thick and $5 \, \mathrm{mm}$ deep.

Figure 20: use of grid to reduce scattered radiation (schematic diagram)
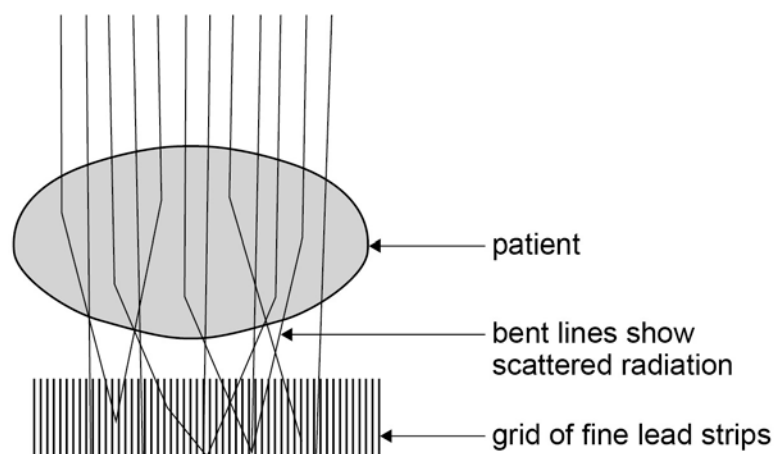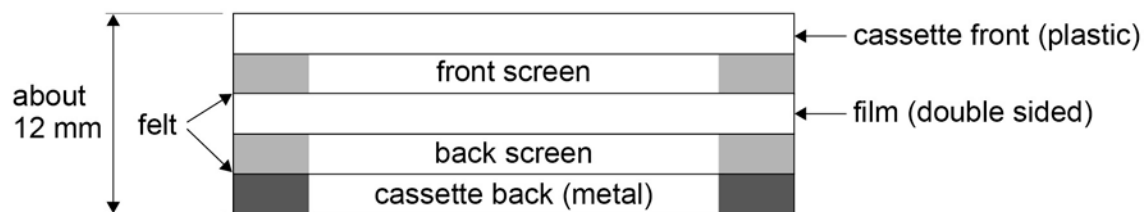
i) The intensifying screen

The use of intensifying screens increases the resulting exposure of the film, an important consequence of which is to allow a smaller X-ray exposure to be used than would be necessary in order to give the same density without the screen.

Some materials are able to absorb X-rays and re-emit some of the energy in the form of visible light photons, the amount of light being proportional to the quantity of X-ray energy absorbed. Hence any pattern in the X-ray beam will be converted into an identical visible pattern.

The intensifying screen consists of a stiff sheet of cardboard (about $1\,\mathrm{mm}$ thick) over which is spread a uniform layer of calcium tungstate or other fluorescent material in the form of small crystals, attached to the cardboard by means of an inert binding material. Between the cardboard and the fluorescent layer, there is usually a thin layer of a very white material (eg magnesium oxide or titanium(IV) oxide) which serves to redirect to the film a large proportion of the emitted visible light which would otherwise be lost. The screen is protected from scratching by means of a very thin but tough, waterproof material.

**Figure 21: a double coated film and intensifying screens in a cassette**

Intensifying screens are used in pairs, being held in close contact with the emulsion on each side of the double-coated X-ray film; the sandwich of screen-film-screen is contained in a light-tight envelope called a cassette. A sectional diagram of a loaded cassette is shown in **Figure 21**, the felt pad ensures that the screen and film emulsion are in close contact over their whole surface. The metal backing to the cassette reduces the amount of radiation scattered back to the film sandwich, an undesirable effect resulting in reduced contrast. A pair of calcium tungstate screens have an intensification factor of about 30 times, varying with the screen type.

j) The image intensifier

A radiologist often needs to observe movement inside the body, eg in passing a catheter, by viewing the image produced by X-rays passing through the body. In order to observe such movements, the patient has to be examined over a period of a few minutes; therefore, the exposure to X-rays must be kept to a minimum in order to reduce the dose to the patient and to protect the radiologist. An earlier method involved the X-rays striking a fluorescent screen in order to produce a visual image

which could be observed by the radiologist, but the light output of the screen was low and long periods of 'dark adaptation' of the eyes were necessary. In order to increase the light output and keep the X-ray dose to the patient to a minimum, image intensifiers were developed.

Figure 22: image intensifier
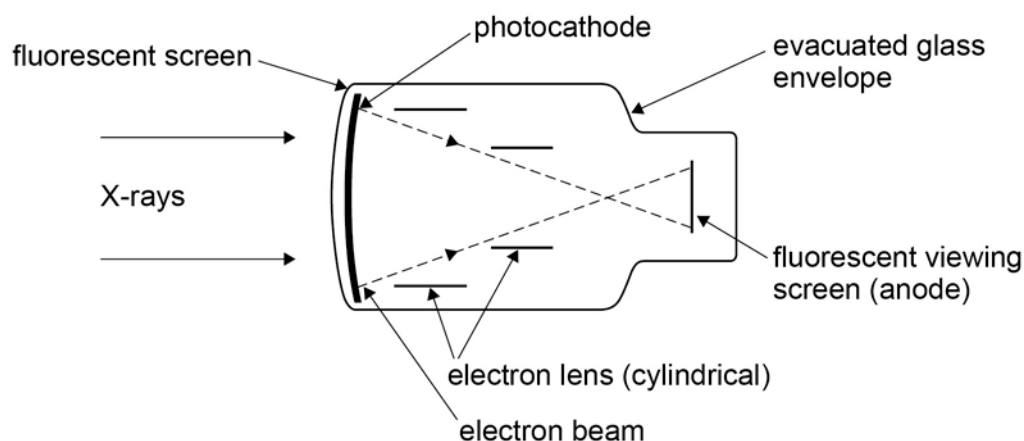
The construction of a typical image intensifier is shown in **Figure 22**. It consists of a fluorescent screen on to which the X-ray pattern is directed. The corresponding pattern of light emitted from this screen is incident on the photocathode. The intensity pattern of the electrons emitted from the photocathode corresponds exactly with the pattern of light incident on it and hence to the pattern of the X-ray beam. The electrons are accelerated by the potential difference (approximately $25 \ \text{kV}$) applied between the cathode and the fluorescent viewing screen. The light produced by the absorption of the electrons, now of higher energy, in this screen is very much brighter than the light which is emitted from the first fluorescent screen. The side of the fluorescent viewing screen nearest to the photocathode is covered by a thin layer of aluminium, which allows electrons to pass through but stops any light produced at the viewing screen from passing back to the photocathode. The pattern is preserved in the electron beam by the electron lens, which also decreases the size of the pattern to about one-fifth of the original on the initial screen. The overall increase in brightness, brought about by the accelerating potential and the reduction in size, is of the order of 5000 times. The viewing screen is observed directly through a lead glass window and can be photographed. Alternatively, the image intensifier can be incorporated in a closed-circuit television system, thus making it possible for more than one person to view the screen at any one time.

k) Indirect flat panel detector

A scintillator, such as cesium iodide, converts the X-ray photons into light photons and channels the light photons towards a photo-diode array. The low-noise photo-diodes in the array absorb the light photons and convert them to stored electronic charge. Each photo-diode represents a pixel or picture element. The charge on each pixel is then read out by electronic scanning and sent out as digital signals to an image processor to form a digital image.

Flat-panel detectors are more sensitive and faster than film. Their sensitivity allows a lower dose of radiation for a given picture quality than film. The image is also more transportable, being available to view on a tablet computer.

l) CT scans

This method of imaging produces images of a cross-section of the body. A highly collimated, monochromatic X-ray beam is rotated about the object being scanned, and the transmitted intensity is registered by detectors directly opposite to the source. The X-ray source and the detectors are housed in the scanner gantry and the patent lies on a couch in the void at the centre of the gantry. The detected intensities are fed into a computer which analyses the results and constructs a picture of the cross-section. The whole scan needs to be repeated for each cross-section. In general, several cross-sections are scanned and this allows the area under investigation to be seen as you move through it.

Advantages

* Good for investigating critical bone fractures and calcification of organs, will also give good image of the brain and abdominal organs.

* Better resolution compared to ultrasound.

* Gives full cross-sectional image.

* It is non-invasive.

Disadvantages

* Highly ionizing.

* Limited contrast between tissues of similar density: images of the brain can be distorted by nearby bone.

* More expensive than a conventional X-ray picture.

* Often requires patients to hold their breath which some may find hard to do.

Section 6: Magnetic resonance imaging (MRI)

The patient is made to lie inside a large, very strong, cylindrical magnet. The spinning hydrogen nuclei align themselves with the magnetic field. A pulse of specific radiofrequency electromagnetic radiation is applied which causes the hydrogen nuclei to align in a different direction. When the pulse is removed, the nuclei return to their equilibrium positions and release RF radiation. This occurs over a period of time called the relaxation time, different types of tissue having different relaxation times. A ring of detectors pick up the emitted radiation pulses, and this information is fed into a computer which translates the information into a sectional image.

Advantages

- MRI scans do not involve ionising radiation. As far as is known there are no harmful effects.

- The ability to distinguish different types of soft tissue is better than that of a CT scan.

- The resolution for soft tissue is better than a CT or ultrasound scan.

- It can show both three-dimensional and cross-sectional images.

- It is non-invasive,

Disadvantages

- MRI scans cannot be used if the patient has any metallic implants, such as a pacemaker.

- The patient might have to be still for up to an hour.

- Bone and calcium do not show up on this type of scan.

- The cost is greater than any other scan.

Section 7: Radionuclide imaging and therapy

a) Imaging techniques

Radionuclides may be used to target specific regions of the body so that the amount absorbed will give an indication of problems in that region. Ideally, the radionuclides used should emit only gamma photons as all alpha and most of any beta particles emitted would be absorbed by the body and would thus serve no useful purpose.

The most widely used radionuclide for diagnostic purposes is technetium 99m, a metastable isotope of technetium which decays to technetium 99 by emitting two gamma photons. It has a physical half-life of six hours, short enough to prevent it exposing the patient to excessive radiation, but long enough to allow the relevant investigation to be carried out fully. The energy of the emitted gamma photon is suitable to be detected by a gamma camera. It can be manufactured in a hospital laboratory using a molybdenum-technetium generator relatively cheaply. It can be incorporated easily into a wide range of labelled radiopharmaceuticals, thus allowing the compound formed to target the organ under investigation.

Iodine 131 is used for both diagnosis and therapy of the thyroid gland. It is both a beta and gamma emitter, but the extra radiation hazard caused by it being a beta emitter is a problem which is outweighed by the iodine being absorbed by the thyroid. It has a half-life of 8 days.

Indium 111 is a gamma only emitter with a half-life of 2.8 days and is used to label antibodies and blood cells. It is much more expensive, but far more effective for this particular labelling.

b) Positron Emission Tomography (PET) scans

This method of imaging produces both 3-D images and cross-section images of the body. A radionuclide is injected into the patient and the patient relaxes for about 1 hour. The patient then lies down on an examination table, which is then moved into the large hole in the scanning machine so images of inside the body can be taken. During scanning the patient should stay as still as possible. It normally takes around 30–60 minutes to take a scan but it depends on which part of the body needs to be scanned.

The injected radionuclide has been labelled to be absorbed by the body part to be examined. Once absorbed it is broken down and emits positrons which are quickly annihilated producing gamma photons. The gamma photons are detected and the signals from the detectors are sent to a computer to analyse the signals and build up an image.

c) Gamma camera

One of the basic components of a gamma camera is a photomultiplier tube. Light falls on a photocathode which releases a small number of electrons. The electrons are accelerated towards the first of a series of electrodes called dynodes, each dynode being more positive than the previous one. When the electrons collide with a dynode, each incident electron releases about 4 electrons. This happens at each dynode before the final electrons reach the anode. If 10 dynodes are used, this produces an overall amplification in the number of electrons reaching the anode compared to the number of electrons released from the photocathode of 4^{10} or 1×10^6. This pulse of electrons reaching the anode creates the electrical output pulse.

In the camera, gamma photons from a patient first travel through a lead collimator before reaching a large diameter crystal of sodium iodide, typically about $40\,\mathrm{cm}$ in diameter and $1\,\mathrm{cm}$ thick. The collimator ensures that the gamma photon reaching the crystal has come from a position in the body directly below that point in the crystal. The gamma photon causes scintillation in the crystal and light photons are produced. Just above the crystal is an array of photomultiplier tubes. The light photons pass into the photomultiplier tubes which produce electrical pulses. The output pulses from the tubes are then analysed by a computer to build up an image.

A single gamma came can be used to produce a two dimensional image, but if more than one camera is used as in a PET scan, a three dimensional image can be produced.

d) Physical, biological and effective half-lives

When used in radiotherapy, the radiation emitted in the body from the radioactive sample placed within the body will reduce with time due to two processes, the normal physical decay process and the secretion of the material from the body by biological processes. The combination of these give rise to an effective half-life for the decay.

The physical half-life, TP, is defined as the time taken for half the initial nuclei in a sample to decay.

The biological half-life, TB, is defined as the time taken for half the initial sample of material to be excreted from the body by biological means.

The effective half-life, TE is defined as the time taken for the rate of decay within the body due to the initial sample to decay to half.

These are related using the equation $1/T_E = 1/T_P + 1/T_B$

e) Use of high energy X-rays in therapy

Radiotherapy is used to try to kill or contain malignant tumours. Tumours deep within the body must be treated using high-energy X-rays. These X-rays will penetrate the body whilst hardly any will be absorbed by the skin and surface tissue. However, the ionising radiation will kill cells indiscriminately, so not only will tumour cells be destroyed, but healthy cells in the path of the X-ray beam may also be affected. In order to limit the effect on heathy cells in the region of the tumour several steps will be taken. These steps include:

- use of accurate scans to locate the tumour to be treated

- correct energy X-rays to be selected

- use of multiple-beam or rotating-beam therapy controlled by computer to focus energy at the tumour and limit exposure of surrounding tissue.

- use of fine collimated X-ray beam.

f) Radioactive implants in therapy

Some tumours are best treated using radioactive implants which emit beta radiation. This radiation will lose most of its energy in the region very close to the implant, killing cells at the implant site, but doing little to damage to cells or tissue further from this site. These implants need to be active for longer and so Iridium 192 which has a half-life of 74 days is often used.

As mentioned in section A, Iodine 131 is absorbed by the thyroid gland and is used for killing tumours as well as in the initial diagnosis.