Computed Tomography for Radiographers

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by

M. J. Brooker

Superintendant Radiographer
Poole General Hospital
Dorset

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Preface

The purpose of this book is to give the reader a basic introduction to computed tomography. It is also directed towards qualifying radiographers and to all qualified personnel who are associated with computed tomography.

Computed tomography is a complex technology, but it has been my aim to keep the text concise using diagrams and images wherever possible. A brief review of X-ray tubes and general physics has been given, although the mathematics of image reconstruction has been omitted. If required, dedicated works on the above subjects may be referred to by the reader. This book concentrates on the application of the above to computed tomography.

Computed Tomography for Radiographers is divided into two sections. Part 1 is dedicated to the technical aspects of computed tomography and Part 2 to the clinical aspects.

Computed tomography is also known as:

Computed axial tomography (CAT)
Computed aided tomography (CAT)
Computed transverse axial tomography (CTAT)
Computed transmission tomography (CTT)
Reconstruction tomography (RT)
Selective computed assisted tomography (SCAT)
Transmission axial tomography (TAT)

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Malcolm Brooker

Part 1
Technical Aspects of Computed Tomography

1
Introduction

Computed tomography (CT) has made great clinical and technical advances during the last decade.

Figure 1.1 shows the first 'prototype scanner', whereas Figure 1.2 is a photograph of a contemporary scanner. On initial comparison the photographs would seem to illustrate great differences between the two systems, however, the basic gantry in both systems is similar.

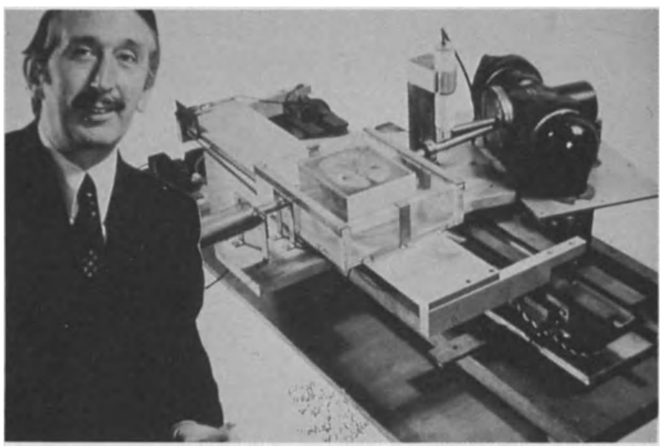


Figure 1.1 Godfrey Hounsfield alongside the original lathe bed prototype. Note the specimen contained within the perspex block

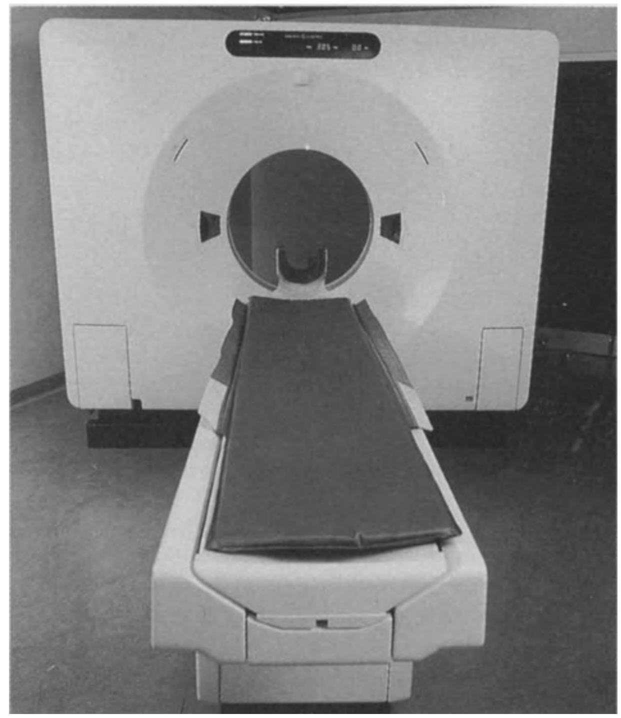


Figure 1.2 Modern day system

Originally an isotope source (Americium) was used, but this had the disadvantage of being of a low intensity when highly collimated, and in order to create an image an exposure time of 9 days was necessary. The prototype scanner used a lathe bed as the gantry frame which allowed precise movement of the specimen, the matrix size of 80×80 was dictated by the thread size on the lathe bed screw and the reconstruction of the images took, approximately, two and a half hours.

The exposure time of 9 days was the main limitation of this system, and the Americium source was later replaced with an X-ray tube, reducing the exposure time from 9 days to 9 hours.

The medical world was unaware of the major developments that were taking place in this field at this time. However, the next stage was to scan and produce images from clinical specimens. Two eminent radiologists then became involved with this work, one of whom supplied brain specimens and the other abdominal specimens; pigs' abdomens were readily available for research work.

The early work demonstrated clear differentiation between muscle, fat and other body tissue in the pig specimens and white matter could be easily distinguished from grey matter within the brain specimens.

The first prototype brain scanner was gradually developed amid great secrecy. However, in January 1970 a very important meeting took place at the Department of Health in London on *Transverse Axial Zonography* which subsequently changed the development of radiology.

Facilities were made available at the Atkinson Morley Hospital for the installation of the first brain scanner. This hospital was chosen because of its close proximity to the research laboratory. The scanning time was, approximately, 4 minutes, but the processing of the images was undertaken away from the site of the scanner.

Many experimental brain scans took place before the medical world was informed of this latest development. In April 1972 a paper was presented at the British Institute of Radiology by Godfrey Hounsfield and Dr J. Ambrose.

Technical advances combined with clinical advances enabled a machine to be developed with a 160 × 160 matrix. Within 2 years of this success the first body scanner was installed in a London hospital. Further technical advances were made, the two main features being a scan time of 20 seconds and the addition of an on-line computer.

Several companies then became interested in developing and manufacturing body and brain scanners resulting in a tremendous diversification in the theories of CT and scanner design.

At this point it is justifiable to stand back to admire and praise those who pioneered the project, Godfrey Hounsfield, the inventor, and his small team of experts, and the two radiologists, Dr J. Ambrose and Dr L. Kreel who carried out this hard, tedious work.

CT scanners are now manufactured worldwide by many different manufacturers, some systems being more popular than others. However, although all the different manufacturers have devised their own processes to obtain the CT image the basic principles of image production are, very often, the same.

CT scanning may be thought to be a difficult field of radiography to study and to understand because of the combination of two different fields of technology; one has to understand existing X-ray technology plus the sophisticated technology of computers. There is also the very distinctive terminology used in CT scanning.

However, by breaking down the main CT system into subsystems with a step by step analysis, CT scanning can be readily understood.

Basically, a CT scanner comprises five main units (Figure 1.3):

(1) gantry
(2) X-ray production
(3) computers
(4) consoles
(5) image storage and production.

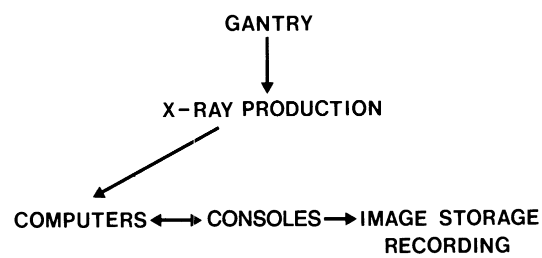


Figure 1.3 Diagram to represent the five major components of a CT system

These five main units can be further divided into subdivisions or peripherals (Figure 1.4).

The gantry houses the X-ray tube and detectors. The data, transmitted in the form of X-ray photons, pass into the detector where they are converted into a digital form, and

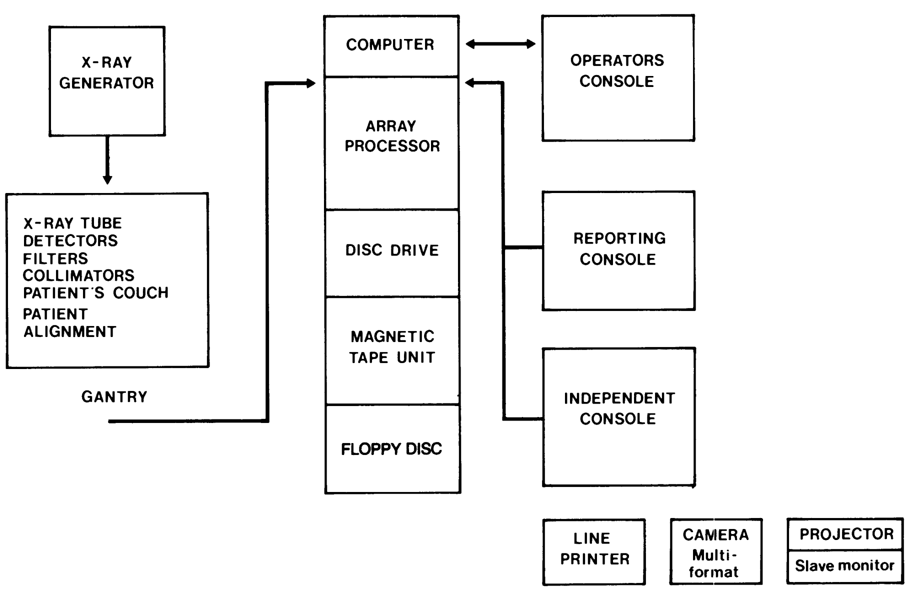


Figure 1.4 Diagrammatic representation of the CT scanning system

then passed into the computer where analysis and reconstruction takes place. The reconstructed images are viewed on the console and may be permanently archived on magnetic tape or imaged on to hardcopy film.

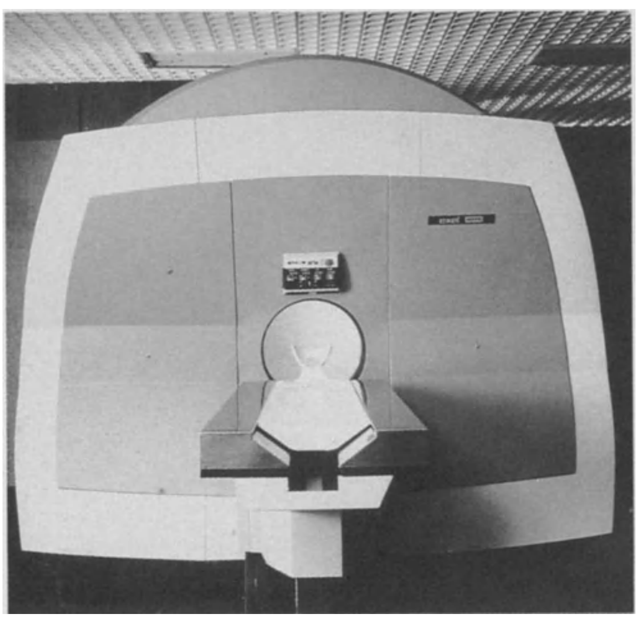
The image comprises a basic unit, the CT number, and is called the Hounsfield Unit, named after the pioneer of CT. The CT number is related to the attenuation coefficients of the body tissue, and will vary if one of the following criteria changes:

(1) As the X-ray tube ages permanent damage occurs on the anode (crazing), and the glass insert acquires a deposit of tungsten,
(2) kVp,
(3) Body temperature,
(4) If a mono-energetic beam rather than a poly energetic one is adopted,
(5) Presence of high atomic number material,
(6) Miscellaneous artefacts (see Chapter 6).

2
The gantry and X-ray production

THE GANTRY

The gantry is the moveable frame of the system housing the X-ray tube, detectors and associated electronics. It is through the aperture in the gantry that the patient passes backwards and forwards while lying on the table, and it is, therefore, the part of the entire system which is within the main view of the patient. Examples of various gantry designs are shown in Figures 2.1–2.3.



Figures 2.1, 2.2 and 2.3 Examples of gantry design

As the gantry acts as a support for the various elements described above, it has to be rigid, but at the same time lightweight. The size and design of the gantry varies between manufacturers and also depends on the technical features of the scanner complex, for example, the geometry adopted (Figures 2.4 and 2.5).

The gantry is usually 2–2.5 m high, 2–3 m wide and 0.5–1 m deep; there is a considerable variation in the weight of the gantry but, on average, it is usually about 2000 kg.

An important design feature must be the relatively easy access to the inside of the gantry for servicing. Generally, the front and back covers of the gantry swing upwards or extend outwards, these are usually made of fibre-glass which is chosen for its rigidity, strength and light weight; it is also easy to clean and is easily manufactured.

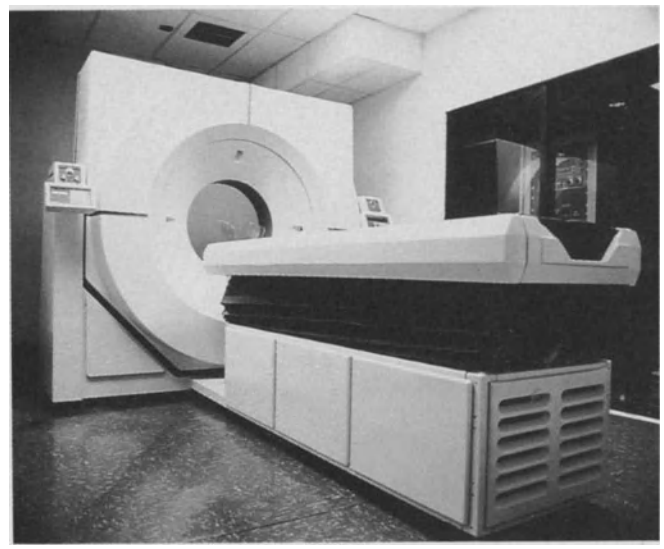


Figure 2.2

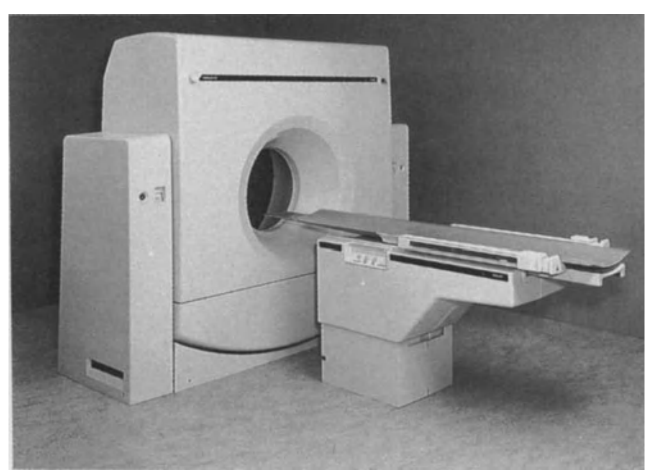


Figure 2.3

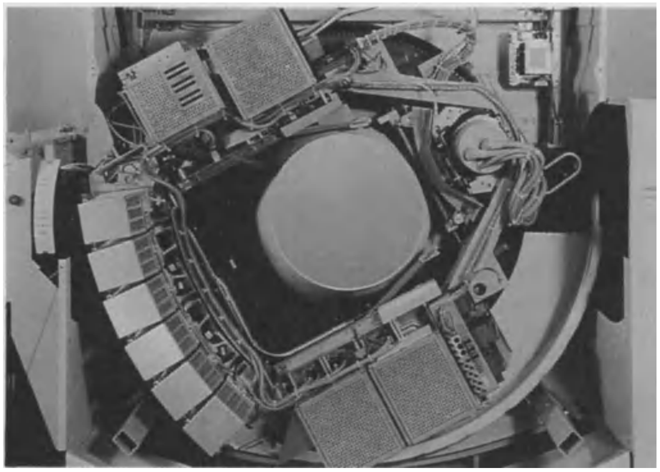


Figure 2.4 A typical third generation system

A recent clinical feature is the ability of the gantry to angle plus or minus 30° off the perpendicular (Figure 2.6). This angulation allows the operator to achieve coronal images more readily.

THE APERTURE

The aperture of the gantry is the part of the system in which the patient has to lie whilst being scanned. It may be thought that very little can be written about a hole, but the size of the aperture varies between the different systems manufactured.

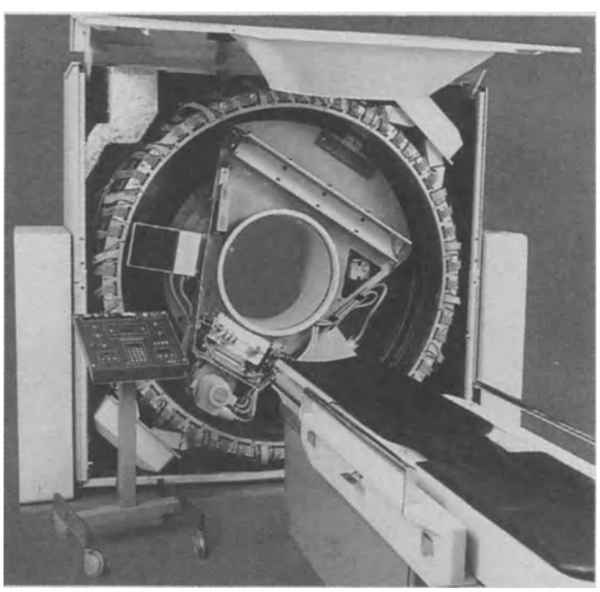


Figure 2.5 A typical fourth generation system

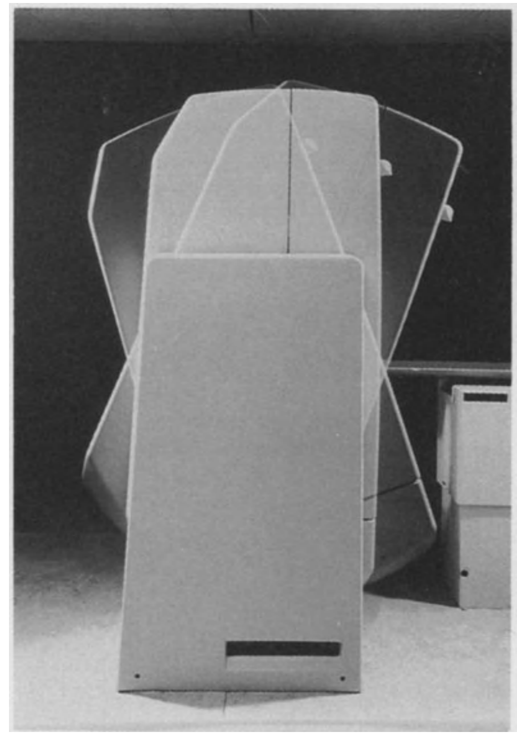


Figure 2.6 A gantry angled from the perpendicular plane

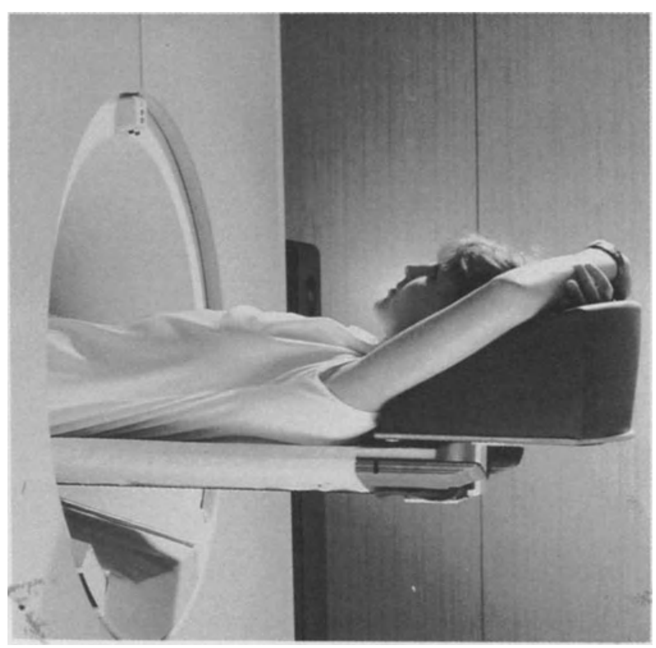


Figure 2.7 Careful design of the aperture creates a feeling of space for the patient

The aperture is usually 50–70 cm in diameter, the most modern systems usually having an aperture diameter of 70 cm. With older systems the small apertures caused many clinical problems.

In more recent systems the design of the gantry adjacent to the aperture is also of significance. If the edges surrounding the aperture are removed as much as possible the aperture appears to be larger (Figure 2.7).

Some systems allow scanning to be performed obliquely across the head or body by slewing the table which can only be achieved by scanners with large apertures. When scanning organs that are oblique to the slice axis, e.g. the pancreas, a slice demonstrating the organ in its entirety can be achieved, but in scanners without this feature a series of axial images are required to perform this study.

However, there are technical problems associated with large apertures. One is the fact that the X-ray source and detectors are further apart, which means that the X-ray tube must have a significantly higher output to ensure that there is a high enough X-ray photon flux at the detectors (inverse square law).

THE TABLE

As the patient lies upon the table during the scan, it is very important that it should be comfortable but firm in order to give the patient a feeling of security. It is very important for the patient to remain as still as possible during an examination and a patient who is feeling confident and comfortable will be more likely to achieve this effect.

The table, which is also known as the couch, platter or scoop, may be curved or flat. Flat tables are necessary for:

(1) *Radiotherapy planning* – so that the patient's position can be simulated as accurately as possible. There should be no distortion of the table under the weight of the patient.

(2) *Interventional techniques* – some radiologists/physicians prefer to perform biopsy techniques on a flat table.

(3) *Ease of patient transfer* – flat tables allow patients to be transferred from a bed or trolley onto the table and *vice versa* with the minimum discomfort. There is at least one system available where the patient's trolley can be detached from and re-attached to the gantry (Figure 2.8).

The table and the table's driving mechanism must have the ability to move smoothly and to be accurately positioned to an exact anatomical location. For example, in the scanning of the petrous bone 1.5 mm slice thicknesses are usually used.

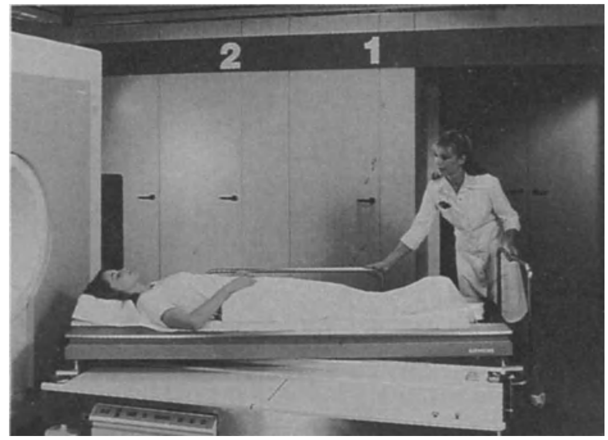


Figure 2.8 Detachable patient trolley

The constant re-positioning of the table to this accuracy should not present any problems providing the patient is co-operative.

The table should have the ability to be lowered to about 60 cm to allow non-ambulant patients easy access onto the table without the use of a footstool.

Rounded edges are essential on the table to prevent any abrasions or lacerations occurring to the patient while moving onto or off the table.

A rigid table top is essential so that accurate patient positioning can occur. Some patients are obese, and it is important that the table top can support their weight without any distortion. Contemporary table tops are usually made of carbon fibre, older systems usually consisted of marine plywood.

Radiolucent tops are necessary so that the beam intensity is not interrupted and to prevent artefacts occurring to degrade the image quality.

The table top should be easy to clean, as it must be cleaned after every patient.

PATIENT ALIGNMENT LIGHTS

As with conventional radiography patients being scanned must be accurately positioned. In body scanning, patients may be scanned in a decubitus, prone or supine position. As well as positioning the patient in the axial plane it is also important to accurately position the patient centrally in the isocentre of the field; should the patient be off centre there may be a degradation of the image quality (Figure 2.9).

There are two main types of patient centring devices:

(1) *White lights* – are usually high intensity halogen bulbs located in one or more planes, usually mounted on or

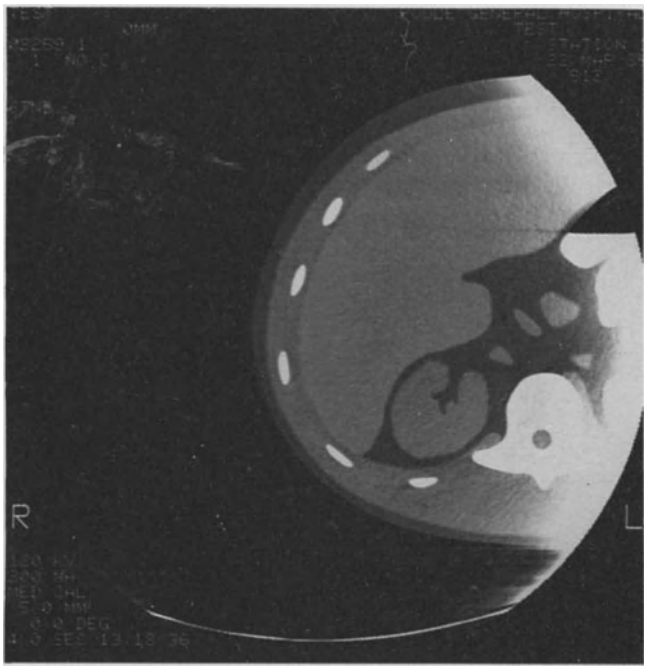


Figure 2.9 The anthropomorphic phantom demonstrating severe image degredation when off-centred in the reconstruction field

adjacent to the gantry. The beam is tightly collimated to shine a fine beam on the patient (Figure 2.10).

(2) *Low intensity lasers* – may be situated on the walls adjacent to the machine and are accurately set up to shine onto the patient; alternatively, they may be mounted onto the machine. It is usual for three lasers to be used to localize in three planes, midline, axial and depth of the body.

Figure 2.10 One or more light sources are used to obtain the isocentre of the scan field

Lasers have three main advantages over white lights.

(a) They have a longer life than halogen bulbs,
(b) They align more accurately, and
(c) They can be seen more clearly in a bright white light than can a halogen source.

With both sources the principle involved in aligning the patient into the correct position is the same, the distance from the X-ray beam to the light source is constant. When the patient has been positioned on the table the 'advance' button is depressed and the machine, under the control of the computer, moves the patient into the aperture to the first scan position (Figure 2.11).

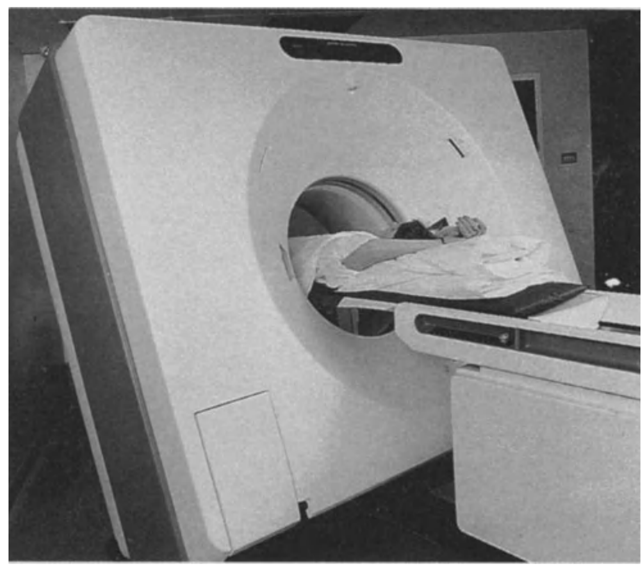


Figure 2.11 Patient lying on the table within the aperture in scan position. Note angulation of gantry

X-RAY PRODUCTION

The original lathe bed 'scanner' developed by Hounsfield used an isotope source (monochromatic beam), however, it soon became evident that its use was severely limited because of the low intensity of the beam. By using conventional stationary and rotating anode tubes this and other problems were overcome.

The method of rectification to produce the necessary wave form and type of voltage in CT, in the majority of cases, is similar to that used in high powered X-ray equipment in conventional radiography. However, the suitability of the two types of X-ray tubes in terms of advantages and disadvantages may differ when related to CT.

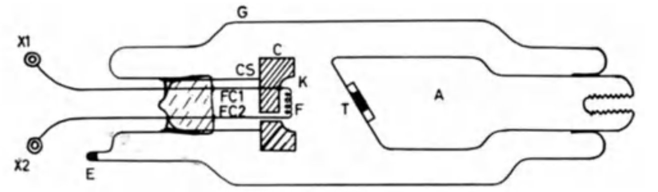


Figure 2.12 Stationary anode tube – A, anode; C, cathode; CS, nickel tubular support; G, heat resistant glass; E, vacuum pump attachment used during the manufacture of the tube to remove the air; F, filament; FC1, FC2, filament heating supply; K, focussing cup; T, target; X1, X2, terminals to filament supply

Stationary anode X-ray tubes (Figures 2.12 and 2.13)

The target of stationary anode tubes consists of a high atomic numbered material embedded into a copper block, e.g. tungsten (atomic number 74 which, characteristically permits greater X-ray production). The dimension of the tungsten is usually 1–1.5 cm^2, anode angle ranging between 15 and 20°, and a focal spot of 2 × 16 mm. Tungsten is also used because it has a high melting point, 3370°C, and a high thermal loading. The purpose of the copper surrounding the tungsten is to aid heat dissipation from the target.

As the stationary anode tube has a high heat loading capacity it allows long exposure times, being suitable, therefore, for CT scanning. The older systems had scan speeds ranging from 20 to 60 seconds, and therefore the use of the stationary anode tube was ideal; the maximum load capacity was in the region of 5 kW on continuous load.

The stationary anode tube has three main disadvantages:

(1) There is a higher radiation dose to the patient.
(2) There are higher 'noise' levels on the images as a result of the long continuous load.
(3) The cooling of the tube is achieved by using expensive, bulky equipment.

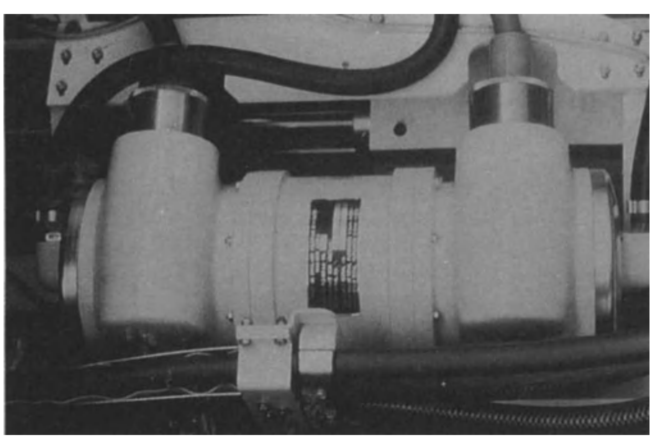


Figure 2.13 A stationary anode tube mounted on the gantry

Contemporary third and fourth generation systems have scanning times of under 10 seconds, thus, the stationary anode tubes do not have a sufficient output of X-rays over the shorter exposure time, and as a result many manufacturers are now using the rotating anode tube.

Rotating anode tube (Figures 2.14 and 2.15)

The rotating anode tube has either a continuous beam or a pulsed beam using the 'gating' principle over short exposure times. The focal spot size varies between manufacturers, but is usually 0.6–1.2 mm^2.

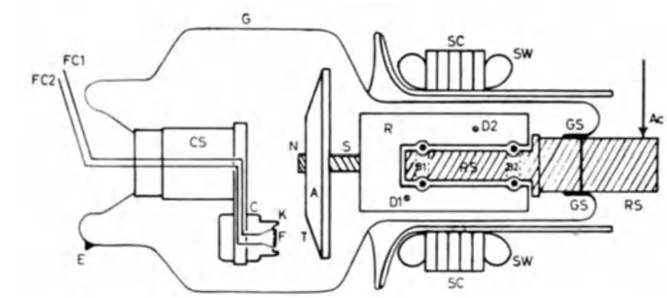


Figure 2.14 Rotating anode tube – G, heat resistant glass (borosilicate); E, evacuating stub; CS, nickel cylinder; C, cathode; K, focussing cup; F, filament; FC1, FC2, filament leads; T, target; A, anode; N, locking nut; RS, rotor support; S, anode stem; R, rotor; B1, B2, rotor bearing; SC, stator coil; SW, stator windings; GS, iron ceiling conducting ring; D1, D2, adjustment screws; AC, alternating current

The rotating anode tube has two main advantages:

(1) *Reduced focal spot* – geometrically this means that there is a reduced penumbra effect of the radiation beam, reducing the radiation dose to the patient during scanning in contiguous slices.
(2) *Efficient cooling capacity* – the cooling system is often built into the tube, and although this results in a slightly larger tube there is less wastage of overall space. The cooling takes place using oil cooled by an air fan.

Over a unit length of time, the rotating anode tube is more efficient in X-ray utilization than the stationary anode tube. The main disadvantage of the rotating anode tube is in loading. One manufacturer using 600 mA at 120 kVp produces a high heat load, hence the tube cooling time, which is computed into the system, is lengthy between slices.

Pulsed X-ray exposure is used by some manufacturers which allows the data acquisition system to re-calibrate between pulses, resulting in a more uniform image. The tube may be pulsed at 1, 2 or 3 ms.

CT scanners should be able to distinguish tissue density of 0.25–5% but an erratic tube output will not allow sufficient density resolution to be achieved.

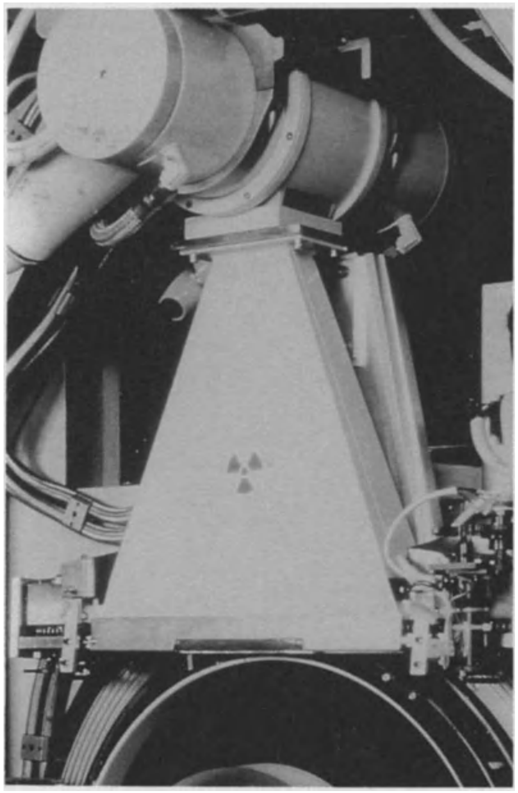


Figure 2.15 A rotating anode tube mounted on the gantry

A high workload is placed on CT scanners and hence on the X-ray tube. The stationary anode tube has a serviceable life of 500–3000 hours depending on the design of the system and on the skill of the operator. Rotating anode tubes vary between manufacturers, but the life of the tube is measured in exposure slices; at least 10 000 exposures should be guaranteed, and one tube manufacturer is now experimenting with a tube that should have a life of over 100 000 exposures.

Tube alignment

Tube alignment is critical in CT. Misalignment causes severe image degradation in the form of concentric rings within the scan on the rotating system, and in linear streaks in the translate/rotate systems.

Translate/rotate systems may have their tubes with the long axis parallel to the scanning plane. With rotation geometries their long axis is perpendicular to the scanning plane, thus avoiding the inertial force on the rotor bearings and the heel effect.

As with all good radiographic practice the better the equipment is looked after the longer is the life of the tube.

GEOMETRY AND GENERATION

In stating the geometry of a system one is stating a method of categorizing the technology and performance characteristics of a scanner. This system of categorizing was devised because of the rapidly changing technology.

The two main features considered in the geometry of a system are the type and number of detectors and the motion of the gantry frame.

There are four main divisions (in chronological order):

(1) First generation – pencil X-ray beam (translate/rotate)
(2) Second generation – multiple X-ray beam (translate/rotate)
(3) Third generation – fan beam (rotational movement only)
(4) Fourth generation – stationary detectors (rotating fan beam)

First generation system (Pencil beam)

In this system a 180° radius was scanned in 1° increments. The single beam (pencil beam) made a linear scan across the patient and then the frame on which the tube and detectors were mounted indexed 1°. The linear motion was then repeated and the system indexed another 1° (Figure 2.16).

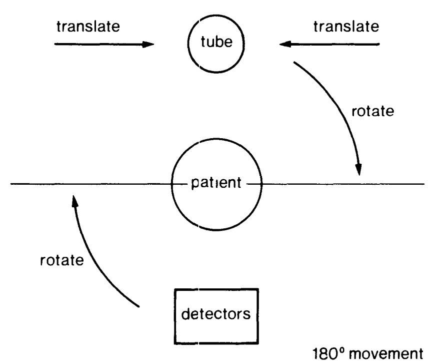


Figure 2.16 First generation scanner, pencil beam, rotate/translate

The linear movement is referred to as the translate and the 1° index as the rotate, hence the translate/rotate geometry.

To produce two CT image slices took approximately 5 minutes, and to produce a complete head study took approximately 35 minutes. Patient movement caused many problems, and these were exaggerated with patients who could not co-operate, e.g. patients with severe head injuries.

Using the above principle, body scanning was impractical because of respiratory movement. This movement caused severe image degradation because of the lengthy exposure times.

However, there were some scanners developed which were so designed that body images could be produced with the patient breathing gently although the image obtained was still of a poor quality.

A second type of scanner was produced using the same basic principle which adopted a multiple beam and multiple detectors. A 3° index was used instead of a 1° index, which reduced the overall scan time to 2.5 minutes. These scanners are no longer being manufactured although there are some still in use.

Second generation system (Multiple beam) (Figures 2.17 and 2.18)

The second generation system adopted a similar basic design to the above, but the number of detectors was increased to 30 or more. The rotate motion was increased to 18° in one system, which reduced the overall time per slice to approximately 20 seconds. The greater number of detectors also helped to reduce the time because the data collection was faster.

The reduction in scan time per slice was the breakthrough needed in CT body scanning. There are relatively few people who cannot hold their breath for 20 seconds, but even so with this system there are motion artefacts present on the image due to breathing or body movement.

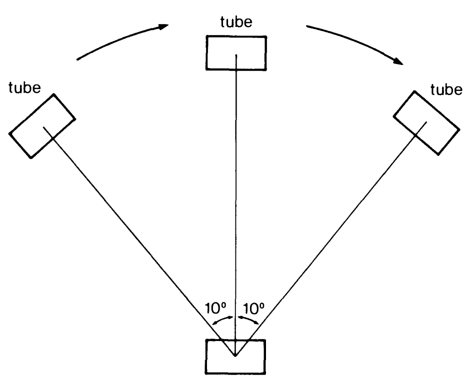


Figure 2.17 Translate/rotate diagram showing the rotate action (indexing)

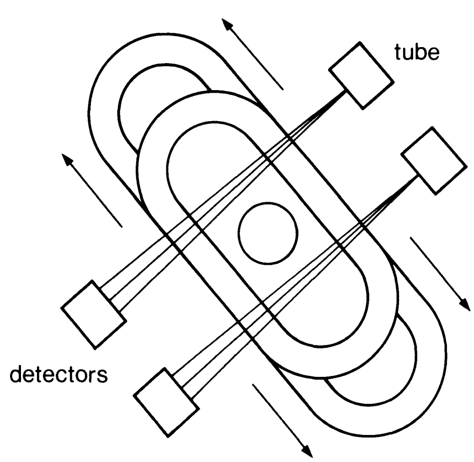


Figure 2.18 Second generation translate/rotate movement of the yoke. This diagram demonstrates the translate movement (linear)

Third generation system (Fan beam) (Figure 2.19)

Until mid 1976 the stationary anode tube was widely used as the X-ray source; however, with the reduction of scan speeds they became a restricting factor. In theory the scan time is decreased when the number of detectors are increased. It is difficult to increase the speed of the translate/rotate mechanism and this became a mechanical restricting factor, but this problem was overcome using rotational movement, resulting in only one motion of the gantry.

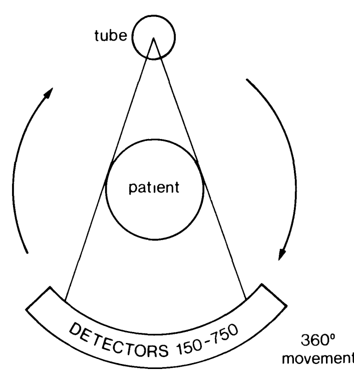


Figure 2.19 Third generation, multiple detector array, pulsed fan

This principle requires the detectors to be arranged in an arc opposite the X-ray tube; the tube now being the rotating anode tube results in a higher output over a shorter period of time. Rotation takes place through 360° not 180° as in the previous design.

The number of detectors were increased to between 250 and 500, and these were set up in an arc usually 40–60° opposite the X-ray source. Scan times of 2–12 seconds were now available.

Fourth generation system (Stationary detectors) (Figure 2.20)

This design comprises a ring of about 600–1200 stationary detectors encircling the patient. The X-ray tube, as before this is now the rotating anode tube, rotates within that ring around the patient. This new generation gives us scan times of 2–10 seconds.

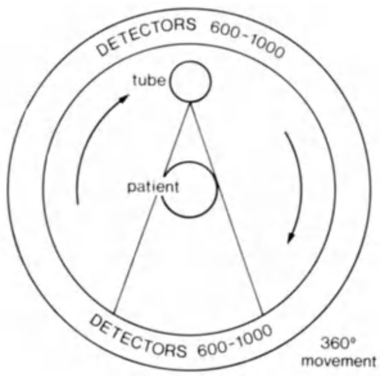


Figure 2.20 Fourth generation, multiple detector array, continuous fan

It is now the aim of every manufacturer to produce rapid images of a very high quality with the minimum radiation dose to the patient, but this is difficult. For example, to achieve rapid images in under 2 seconds would require a high radiation dose, but in achieving a low radiation dose this would introduce high noise levels on the image.

In the highly competitive CT market many manufacturers have attempted to produce a quality scanner and have failed. At one time there were over 20 CT manufacturers, but there are now only a third of this number who can claim any commercial success.

DETECTORS, COLLIMATION AND ASSOCIATED ELECTRONICS

The X-ray tube and detectors form the main part of the gantry, the detector system being situated opposite the X-ray tube.

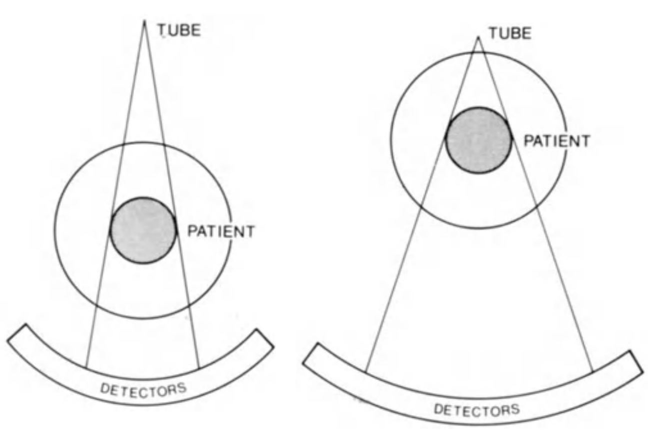


Figure 2.21 Diagram to demonstrate that reducing tube–patient distance allows greater detector exposure to X-ray beam

The purpose of the detectors is to collect the data transmitted from the patient and transfer this data to the computer. The data from the patient, in the form of an attenuated beam is very weak, so the detectors have to change this weak beam into a signal which can be understood by the computer. The detectors are the equivalent of the X-ray cassette and film in conventional radiography.

The distance between the X-ray source and detectors is usually 110 cm, although it is possible to reduce the tube–patient distance which has the effect of utilizing all the detectors. It is usual practice when scanning heads or small bodies not to use all the detectors (Figure 2.21).

The detectors must have certain essential qualities in order to reproduce similar responses to similar energies of X-ray photons from the patient. To assess these essential qualities the following properties must be considered:

(1) Stability,
(2) Physical size,
(3) Cost,
(4) Efficiency, and
(5) Response time.

The perfect detector will be stable, small in size, cheap to produce and very efficient with a short response time.

Stability

The detector must have a consistent response to the incoming beam data and a uniform conversion of these data into a signal that is understood by the computer. A detector is of no use if it responds to different levels of different X-ray photons and a detector which is unstable in this manner will require frequent re-calibration.

Physical size

The problem of space was less apparent with the older systems because of the smaller number of detectors. Modern systems can contain up to 1200 detectors, hence, each individual detector must be smaller in size. An increase in the size of the detector results in an increase in the overall weight and dimensions of the gantry. Whatever the size of the detectors they must always be easily accessible so that they can be replaced as quickly as possible by the service engineer.

Cost

Detectors are very expensive to replace when they break down, and multi-detector systems are very costly.

Efficiency

This is a statement of sensitivity of the detectors to the X-ray beam, the efficiency being measured as a percentage. The higher the efficiency the greater the conversion rate from the X-ray photon to the electronic data.

Response time

The response time, which should be as short as possible to reduce the afterglow effect, is the time taken by the detectors to respond to the X-ray photon and to return to their normal state.

Types of detectors

There are many different types of detectors currently used by manufacturers. Some of the types most frequently used are:

(1) *Sodium iodide* (NaI) - these detectors are usually found in second generation systems where they are coupled with photo-multipliers. This type of detector is highly efficient in its detection rate, and is also small in size and cheaper than other types of detectors.
(2) *Calcium fluoride* (CaF_2) - these detectors have no afterglow, but their efficiency rate is only 80%. They are small in size.
(3) *Bismuth germanate* (BGO) - these detectors have no afterglow and have an efficiency rate of 100%. They are small in size and are, therefore, satisfactory detectors.
(4) *Xenon* (Xe) - the xenon detectors are complex detectors, are small in size, have no afterglow effect and are efficient.

Detectors used in CT fall into one of two categories: (1) scintillation, e.g. sodium iodide and caesium iodide or (2) gas ionization. Xenon gas is used as the gas in ionization detectors, and this may be pressurized to between 20-30 atmospheres thus improving the sensitivity of the detector. Xenon gas type detectors may be used in third generation systems.

Scintillation detector principle (Figure 2.22)

The emitted X-ray beam from the patient strikes the detector's surface, and the crystal within the detector then absorbs these photons and produces flashes of lights (scintillations). This light is directed towards a photomultiplier and is proportional to the amount of X-rays hitting the surface of the detector; the greater the number of X-rays produced the greater the number of light photons produced.

The surface of the scintillation crystal is coated with a material that allows all the light photons to reach the photomultiplier tube, comparable to the effect of a funnel.

When discussing the detection of X-rays in the context of CT the photomultiplier is an integral part of the scintillation detector.

The photomultiplier is an evacuated glass tube comprising an anode, a cathode and dynodes.

When a scintillation from the crystal hits the cathode, the specially coated surface (photocathode) emits the electrons proportionally. The dynodes react with these electrons to form secondary electrons, each adjacent dynode having a slightly higher voltage thus producing a cascade effect. The final output of the photomultiplier is proportional to the scintillation input. This highly amplified signal is then converted, in an analog-digital converter, from an electrical signal into a signal which then passes into the computer.

Gas ionization detector principle

The gas ionization detector works on a simpler principle than the scintillation detector.

In the third generation systems the detectors are usually spaced closely together and, therefore, there are usually a large number of detectors, sometimes 1000 or more, accommodated within the gantry.

Xenon gas, under high pressure of 20 atmospheres or more, is used within the detector to increase the detector's sensitivity. Xenon gas is an inert, colourless gas which is reasonably inexpensive.

The chambers consist of many thin tungsten plates, usually 1.5 mm apart, accurately positioned in an arc to allow for the divergence of the X-ray beam. Alternate plates

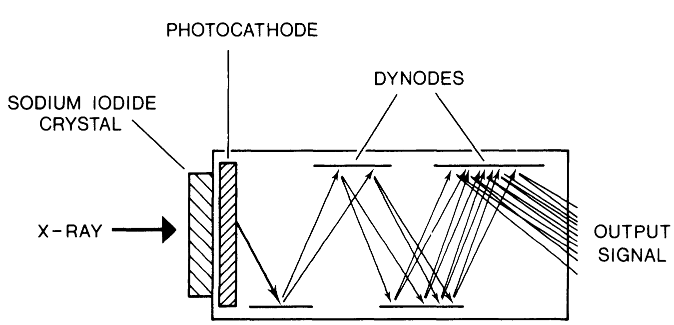


Figure 2.22 Diagram demonstrating the basic principle of a scintillation detector

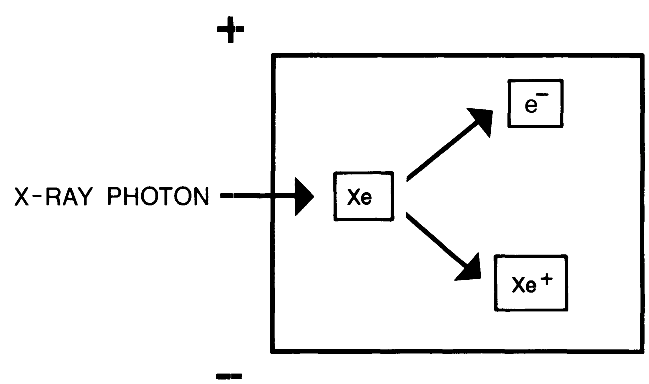


Figure 2.23 Diagram to demonstrate gas detector principle

have a large voltage placed on them while adjacent plates carry a zero voltage.

Ionization occurs as the X-ray beam passes into the chamber, resulting in the negative electrons moving towards the positive plates and *vice versa* (Figure 2.23). The current thus obtained from the plates is proportional to the intensity of the X-ray beam. The greater the number of X-ray photons hitting the detector system the greater the ionization and, therefore, the greater the electrical output signal.

Unless the pressure within all the cells remains constant a drift will occur and the detectors will not produce a stable response.

Gas detectors are 60% efficient, and the system works because xenon gas is used under great pressure thus increasing the size of the chamber and, therefore, allowing the maximum number of electrons to be captured.

Once the signal has been converted into an electrical form it passes into an analog–digital converter. A description and explanation of the analog–digital converter is a lengthy task and is beyond the scope of this book, but it is sufficient to explain that the analog–digital converter has the task of transforming the electrical signal into a digital signal that can be understood by the computer.

A logarithmic technique is used to compress the vast range of readings of converted signals to ensure a proportional value of X-ray photon to digital value.

In CT, collimation of the X-ray beam plays a significant role in the final image production and presentation. It has a particular effect on:

(1) *Patient dose*
Within CT, the diverging nature of the X-ray beam, the penumbra effect of the beam, the contiguous slices taken within the body and brain and the size of the focal spot would result in a very high radiation dose.

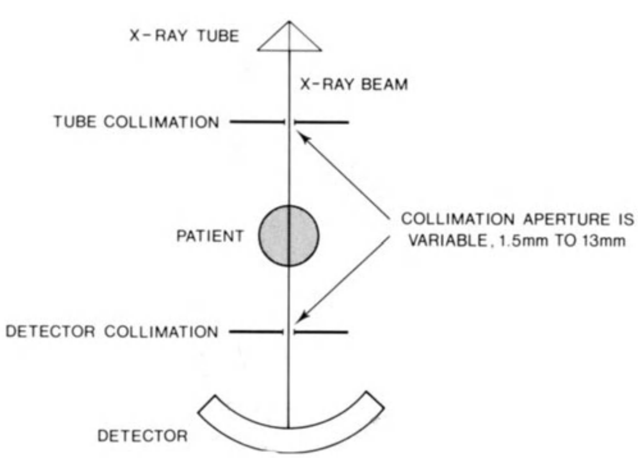


Figure 2.24 Diagram to demonstrate basic principle of CT collimation

The larger the focal spot the greater the penumbra effect and, therefore, the greater the radiation dose to the patient. Collimation is essential to reduce the penumbra effect (Figure 2.24).

(2) *Image quality*
In order to obtain clear visualization of small structures within, for example, the head, thin slice facilities are required. High resolution scanning of the middle ear, orbits and base of the skull is often taken with a slice thickness of 0.5–2 mm. Decreasing the slice thickness increases the resolution of the image, but this also means that the exposure factors have to be increased thus raising the patient radiation dose. The above is an example of a 'trade off' situation regularly occurring within CT.

There are two types of collimators to be found in CT.

(1) *Tube collimator* – this increases or decreases the slice thickness of a cut, and usually ranges from 1.5–13 mm.
In some of the older CT systems these collimators would have had to be fitted manually to the system when required, but with modern day systems collimation is automatically controlled by the operator using a software program integrated into the scanning program.
(2) *Detector collimator* – is located between the patient and the detector. In deciding the collimation a 'trade off' situation occurs, and the following criteria have to be considered:

  (a) scan noise,
  (b) patient dose, and
  (c) slice thickness.

Scan noise is increased if the radiation dose to the patient is reduced. Conversely, if an image has a high resolution the image demonstrates fine structure and can be clearly seen, but this requires high exposure techniques and thin slice thicknesses.

Radiation dose

There are two major factors involved when considering the radiation dose to the patient:

(1) inherent design features,
(2) radiographic techniques.

Inherent design features

Within the design of the CT system there are many features involved in determining the radiation dose to the patient.

(1) Type of X-ray tube – stationary/rotating anode,
(2) Geometry of the system,
(3) Collimation,