4 Principles of CT

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Abstract

This chapter summarises the basic technical principles which underpin computed tomography (CT). The key advantage of CT over conventional radiography is its ability to obtain 2D sections and 3D volume representations of the human body, with greatly improved contrast discrimination between tissues. This is enabled by a rotating X-ray tube and detector array which obtain multiple image projections during scanning. Much CT development occurred via a series of scanner generations, especially spiral (helical) scanning and multi-detector row designs. Imaging is based on the conversion of X-ray linear attenuation values to Hounsfield units which can be transformed to an extended greyscale of signal intensities. Windowing is a means of improving the visualisation of image contrast. Image resolution is determined by factors such as the slice width and pixel matrix. The effect of exposure factors such as kilovoltage peak (kVp) and milliampere seconds (mAs) is considered. Modern methods for CT image formation from raw data include back projection and iterative reconstruction.

4.1 Introduction

From our twenty-first century perspective, it is hard to imagine a diagnostic imaging world without computed tomography (CT). The technique has truly revolutionised the two-dimensional (sectional) and three-dimensional (volume) depiction of internal human anatomy, to the extent that there is no hidden corner of the living body which cannot now be explored by it. This includes the colon and adjacent structures. Prior to the introduction of CT in the early 1970s, internal anatomy could only be explored fully by the surgeon's knife or partly portrayed by ultrasound and radionuclide imaging. Conventional X-ray imaging had long been in existence, but could only display structures and organs in a superimposed state, without much discrimination between different soft tissues. Although magnetic resonance imaging (MRI) would later surpass the soft tissue discrimination capabilities of CT it still

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J.H. Bortz et al. (eds.), *CT Colonography for Radiographers: A Guide to Performance and Image Interpretation*, DOI 10.1007/978-3-319-29379-0_4

suffers to this day from longer scan times, poorer spatial resolution and a weaker ability to depict air-filled structures or tissue calcifications.

4.2 CT Principles

Computed tomography uses mathematical computation to obtain imaging sections or 'slices' of the human body. Indeed its name is derived from the Greek word *tomos* meaning a cut or section. It allows the content of discrete body sections to be seen with clarity and detail, avoiding the superimposition of structures which is such a disadvantage in conventional or 'plain' radiography. Sections are obtained by allowing the X-ray tube to rotate around the body within the axial plane during a procedure, thereby obtaining projections from many different positions rather than from just a single perspective. By mathematical computation, the huge amounts of projection data are transformed into an image. Often CT is described as a CAT scan, which stands for computed axial tomography. Although the mathematical principles of obtaining a 2D slice image from a large number of projections were described by Radon in 1917, the practical realisation of the technique was delayed until the arrival of improved computers in the 1970s. Only then could the large amounts of projection data be transformed into a viable image.

Both CT and plain radiography use X-rays emitted by X-ray tubes and received by X-ray detectors. Both imaging modalities are affected by the properties of X-rays. The useful properties of X-rays include the ability to penetrate through the body in straight lines and cast a shadow projection of internal structures, recordable by detectors. This can provide a 'true' representation of these structures unless the X-rays become scattered from their straight line path or if there is 'noise' in the imaging system. Noise refers to random fluctuations in received signal which degrade an image. Another useful property of X-rays is their 'differential absorption' in body tissues. This means that some tissues, namely, those that are of high density and atomic number, will absorb more X-rays than those that are not. When a lot of X-ray absorption takes place, such as in metal, bone or contrast media, there is a bright white appearance in the CT image. When very little absorption takes place, such as in bowel gas, there is a resultant black appearance in the CT image. Soft tissues tend to show up as intermediate shades of grey. It should be remembered that CT gives us images based on a single physical property – the absorption of X-rays. Different tissues will appear the same on CT imaging if they have the same X-ray absorption characteristics. This can be regarded as a relative disadvantage of CT compared to MRI, since the latter has more ways of depicting tissue properties and also has better functional imaging capabilities. However, CT provides shorter scan times and finer image resolution than MRI, properties which are of particular benefit when imaging the moving structures of the colon.

4.2.1 CT Fundamentals

The key features of a modern CT scanner are as follows and are depicted in Fig. 4.1.

- An X-ray tube which acts as the source of an X-ray beam, rotating in a continuous 360° arc in the axial plane around a patient's body
- Filters which modify and improve the X-ray beam
- Collimators which reduce the size of the X-ray beam, thereby reducing patient dose and improving image quality
- X-ray detectors, arranged in rows, rotating in a continuous arc and located directly opposite the X-ray tube
- A moving X-ray couch, on which a patient lies

As the X-ray tube and detectors rotate in a circular fashion around a patient's body, a large number of X-ray projections are obtained. These can be considered as consisting of many 'ray' traces, with each ray encountering different tissues during its linear course through the body. It is the presence of these multiple traces that makes CT fundamentally different from conventional radiography and enables a complete two-dimensional section to be obtained through the patient. The X-rays within each trace will experience attenuation, according to

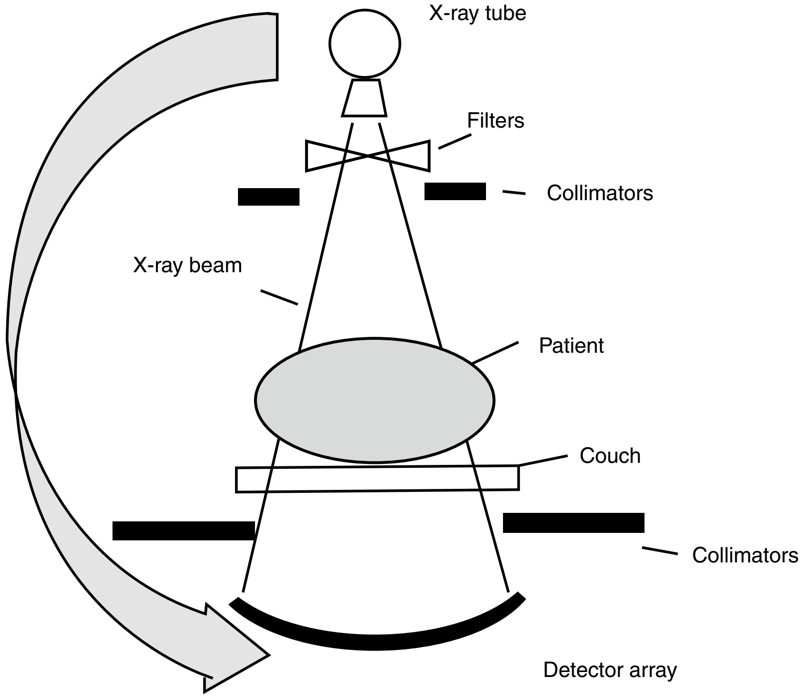

Fig. 4.1 CT scanner components

the total amount of tissue encountered along their path. The amount of X-rays received in the detector array after the X-rays have passed through the patient will depend on the body thickness in that particular direction as well as the combined density and atomic number of the tissues present. Thus, a ray tracing through the body in a particular direction will encounter a superimposed stack or column of tissues. In CT and other forms of digital imaging, the body tissues are considered and depicted as two-dimensional squares (pixels or 'picture elements') or three-dimensional cubes (voxels or 'volume elements'). The total X-ray attenuation in a particular direction will depend on the combined attenuations of the individual pixels or voxels in the column, as shown in Fig. 4.2. The diagram shows a two-dimensional slice through the patient, consisting of a number of pixels (picture elements). Each pixel has an X-ray attenuation value related to its atomic number and density. The total attenuation experienced by each ray will depend upon the combined attenuation of the pixels it encounters within its trace. In the example shown here, ray 2 will be the most attenuated (since it passes through a dense pixel) and ray 3 the least. Each ray here corresponds to an X-ray projection. The X-ray beam experiences attenuation (a reduction in intensity) as it passes through a patient's body, in a way that is

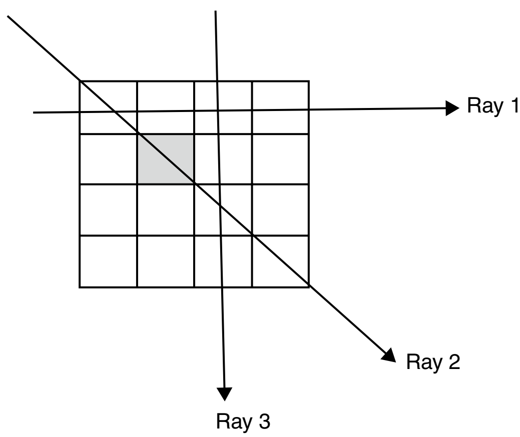

Fig. 4.2 Rays in CT. The diagram shows a two-dimensional slice through the patient, consisting of a number of pixels (picture elements)

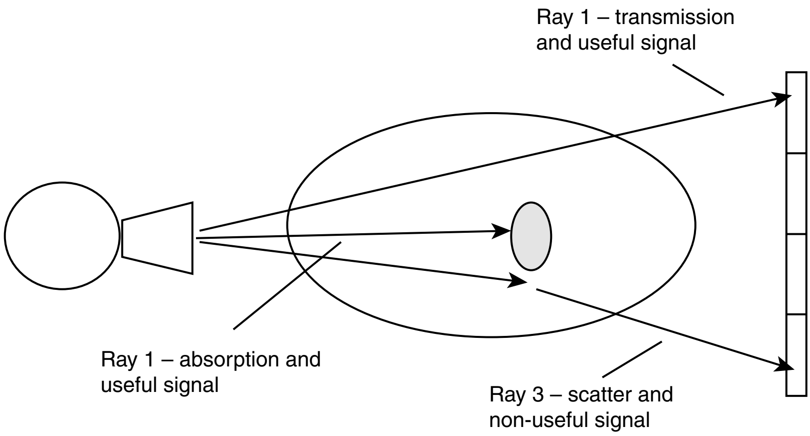

Fig. 4.3 The possible paths of an X-ray photon in CT

determined by the tissues encountered along its path. This process is essential to us in CT, since it produces a signal pattern in the X-ray detectors. There are three things that can happen to an X-ray in CT, as shown in Fig. 4.3.

1. The X-ray may pass right through the body in a straight line (linearly), producing a useful signal in the X-ray detector array.
2. The X-ray may be absorbed in the body. This is also useful, since it produces an 'absence of signal' in the detector array, enabling an absorption pattern to be obtained. This 'shadow projection' results in an image. The X-ray absorption is directly proportional to body tissue atomic number and density as well as the body thickness.
3. The X-ray may be scattered in the body and possibly received by a detector some distance away from the X-ray's original straight line path. This is definitely not useful, since it produces a signal in the detector array that does not correspond to the body anatomy. The X-ray scatter is directly proportional to body tissue electron density (which roughly relates to tissue density) and body thickness.

In CT, we want to obtain both process 1 (signal) and process 2 (absorption) to some extent, whilst minimising process 3 (scatter). The relative amounts of these processes are very much affected by the X-ray exposure factors used.

4.2.2 CT Exposure Factors and the CT Image

There are two principal exposure factors that can be adjusted by the operator during CT scanning.

- Milliampere seconds (mAs)
  This is the amount of electrical current passing through the X-ray tube during an exposure. It has a simple direct effect on the number of X-rays produced, so that a doubling in mAs results in a doubling of X-rays. We should note that mAs does not affect the penetration or energy of X-ray photons, only their number. An increase in the number of X-rays will increase the signal received in a detector array and improve the amount of signal relative to noise. Noise manifests itself as a random fluctuation in image signal, giving what is often termed a 'salt-and-pepper' appearance on the image. X-ray detectors tend to work best when they are receiving sufficient amounts of X-rays, giving an image of high contrast and resolution, without appreciable noise. But of course this comes with an unwelcome increase in patient radiation dose, since dose is directly proportional to mAs. A balance needs to be struck, wherein there is both acceptable image quality and dose.
- Kilovoltage peak (kVp)
  This is the peak voltage in kilovolts applied across the X-ray tube during an exposure. Values used in CT may range from about 80 to 140 kVp. An increase in kVp has a dual effect,

increasing both the energy of the X-ray beam and the number of X-rays produced. In fact the number of X-rays produced is proportional to the square of the kVp. So a simple increase in kVp has the effect of increasing patient radiation dose if other exposure factors remain unchanged. It also has the effect of improving signal in the detectors and reducing noise and also has an impact on image contrast. At this point we should note that a high contrast image is one in which there are large differences in signal between different tissues or structures – in other words it is an image which provides good tissue discrimination. In general, high image contrast is provided by low kVp. This is because X-ray absorption (photoelectric absorption is the physics term) occurs more at low kVp and this absorption process emphasises atomic number differences between tissues. As a result, the signal difference between iodine-containing contrast media (high atomic number) and soft tissue in the bowel (low atomic number) will be maximised at low kVp. At high kVp values, it is X-ray scatter (Compton scatter) which predominates, and thus, the contrast between structures of different atomic number will be reduced. Importantly however, we should note that the X-ray beam must always be of sufficient energy to penetrate through a patient, or else no image will result. Also noise in the X-ray detectors will increase if insufficient rays are able to penetrate through the body and reach those detectors.

There are some other technical factors which affect the CT image; however, not all of them are within the direct control of the operator during a scan: focal spot, geometry, beam filtration, slice thickness, image matrix, detector dimensions, pitch and scan time.

- Focal spot. This is the source of X-rays within the X-ray tube. A small ('fine') focal spot improves the spatial resolution (sharpness) of the image, but also reduces the heat capacity of the X-ray tube. This is an example of the geometric unsharpness (penumbra) effect which is also seen in conventional radiography.
- Geometry. The spatial resolution of the image will be maximised by a small distance between the patient and the detector array, together with a large distance between the X-ray tube and the patient. This is dependent on the scanner design and is another aspect of the penumbra effect.
- Beam filtration. Metal filters, placed between the X-ray tube and the patient, are designed to remove low-energy X-rays ('soft' X-rays) and improve the penetrating capability of the beam. They may also be used to even out the intensity of the X-ray beam across the patient anatomy.
- Slice thickness. A thicker slice contains more signal and thus suffers less from image noise. But a thinner slice provides improved spatial resolution and better ability to depict small objects.
- Image matrix. An axial section in CT consists of a two-dimensional grid of square-shaped pixels (picture elements). A typical value for the matrix is 512 × 512 (i.e. 512 pixels in each of the two dimensions). For a given scan field of view, the size of individual pixels is inversely proportional to the matrix. A matrix of 1024 × 1024 will provide pixels that are half as large in each of the two dimensions and thus four times smaller in area. This 1024 × 1024 matrix will allow better spatial resolution than the 512 × 512 matrix, but each pixel will contain less signal and thus will be more liable to image noise.
- Detector dimensions. The spatial resolution of the image will be improved when using small detector elements but may suffer from reduced signal and thus worse noise [1].
- Pitch. This adjustable technical factor describes the relative speed of the CT couch movement through the scanner during a single X-ray tube rotation, divided by the total width of any simultaneously acquired slices. A large pitch factor (faster couch movement) results in reduced signal and image quality as well as reduced patient radiation dose.
- Scan time. A reduced scan time provides improved temporal resolution, thereby reducing the adverse effects of patient motion on image quality and also permitting dynamic studies of the body in real time. Scan time is affected by a number of factors, including pitch, X-ray tube rotation speed, scan volume and the number of image slices.

4.2.3 CT Image Contrast

Compared to conventional radiography, CT is able to amplify the image contrast that can be seen between different tissues [1]. Image contrast means the amount of signal difference that exists between tissues – so an image containing white and black shades is regarded as higher contrast than one consisting of intermediate shades of grey. How can CT amplify image contrast? There are four ways in which it achieves this.

1. Removal of overlying structures. Conventional radiographic images are compromised by the fact that all anatomy is seen superimposed. This reduces image contrast. In CT, only tissues within a thin section or 'slice' are visible, and this tends to increase the available contrast between them by removing overlying image 'clutter'.
2. Reduction of X-ray scatter. In CT the X-ray beam is tightly collimated ('coned down'), not only before it reaches the patient but also before it reaches the detector array. The overall effect is to lower the amount of scatter reaching the detectors. Scattered X-rays reduce image contrast by raising the amount of background signal and consequently give an unwelcome image 'greyness' which reduces tissue discrimination.
3. X-ray attenuation calculation. CT is able to detect very subtle differences between the X-ray attenuation values of different tissues. This is because it obtains large amounts of X-ray attenuation data, using multiple projection angles. The attenuation values are converted to a range of signal intensities on the CT image, using a scale known as the Hounsfield scale [2], as shown in Table 4.1. This is named after Sir Godfrey Hounsfield, whose pioneering work resulted in the first clinical CT scanner at the Atkinson Morley Hospital in London in 1971. The Hounsfield scale is derived from the relative sizes of the X-ray attenuation coefficients of tissues compared to water. Water is assigned a value of zero Hounsfield units (HU) on the scale, with air having a value of −1000 HU and dense bone a value of +1000 HU. It is interesting to note that fatty tissue has a value of about −100 HU and other soft tissues are in the range of +20 to +80 HU. Hounsfield values are converted to a greyscale of image intensities in CT. In practice a scale of CT numbers is often used. This is based on the Hounsfield scale but extended to about +3000 to allow for the high X-ray attenuation values of metal implants which may be present in the patient.
The Hounsfield equation below indicates that tissue Hounsfield unit values are based on the relative linear X-ray attenuation coefficients μ of the tissue and water, multiplied by 1000. A Hounsfield unit difference of 5 between two tissues corresponds to a linear attenuation difference of 0.5 %:

$$\text{HU value} = \left(\mu_{\text{tissue}} - \mu_{\text{water}}\right) / \mu_{\text{water}} \times 1000$$

Table 4.1 The Hounsfield scale

Tissue or substance type	Typical Hounsfield unit values
Air	−1000
Lung	−1000 to –500
Adipose (fatty) tissue	−100 to –50
Water	0
Soft tissues	+20 to +50
Liver	+40 to +70
Acute haemorrhage	+50 to +100
Cancellous (marrow) bone	+50 to +200
Contrast media enhancement	+100 to +300
Cortical (hard) bone	+250 to +1000

4. Windowing. In CT, tissues are displayed using pixels or voxels, each having a given signal intensity value. The tissues are depicted using a greyscale, whose extremes are white (high X-ray attenuation tissues such as bone or

haemorrhage) and black (low X-ray attenuation tissues such as lung or bowel gas). Soft tissues present as intermediate shades of grey. There are too many shades (too many different signal intensity values) contained within the extensive data of a CT image to be visible to the human eye 'all at once'. In fact the human eye can only visualise about 30–40 distinct shades of grey within a single image. Thus, the data is post-processed using 'windowing'. A window enables only a particular range of tissue attenuation values to be seen on the image, thereby increasing the contrast between them. For example, it is possible to use a bone window, a lung window or a soft tissue window. The user selects a window level corresponding to the midpoint of the range of tissue X-ray attenuation values and a window width which prescribes the range of X-ray attenuation values to be visualised in the image. A narrow window width results in a high contrast image. The process of windowing is shown in Fig. 4.4.

A CT image contains a wide range of X-ray attenuation values, converted into Hounsfield units. But the eye cannot visualise so many grades of signal within the associated greyscale. Thus, the range of tissue attenuations to be visualised on a CT image is narrowed down to a window width of values, centred around an attenuation value which is called the window level. Typical window levels used in practice might be about +50 for soft tissues, −500 for the lungs and +250 for the bone. Windowing is especially important for soft tissue imaging as it amplifies the image contrast between tissues which have small attenuation differences between them and would otherwise not be visible as distinct entities. Tissue attenuation values above and below the window width cannot be visualised as greyscale intensities and show as very low contrast structures – bright white or pitch black, respectively.

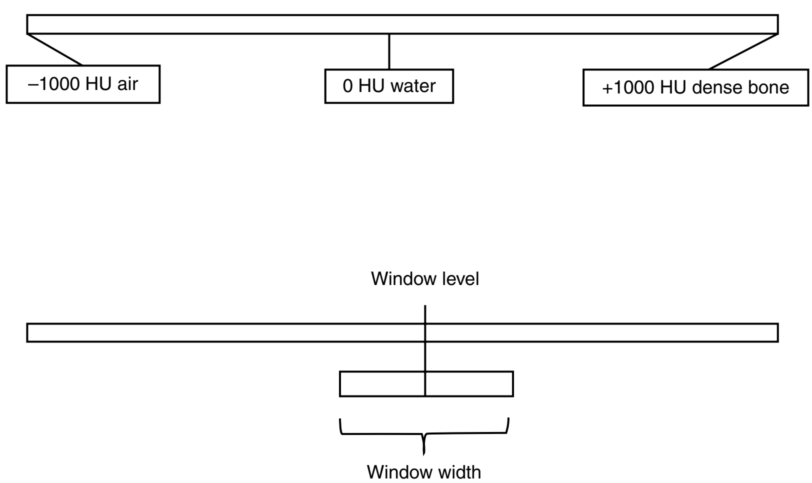

Fig. 4.4 Windowing in CT image processing

4.3 CT Scanner Development

There have been tremendous developments in CT scanner technology and capabilities since the advent of the first system in 1971. Initially CT was restricted to the study of small and relatively motion-free body areas such as the head, but can now image all body contents, including the colon. The benefits of advances in CT include:

- Greatly reduced scan times
- Improved spatial and temporal resolution
- Volume (3D) acquisition
- Slice reconstruction in the sagittal and coronal planes from original axial scan data
- Enlarged scan volume coverage
- Real-time (dynamic) imaging
- Improved signal-to-noise ratios
- More accurate quantification of tissue X-ray attenuation
- Ability to provide some functional information in addition to anatomical depiction
- Advanced image reconstruction techniques
- Image artefact reduction
- Radiation dose optimisation

From its inception, CT provided improved tissue discrimination relative to conventional radiography, due to its improved image contrast and sectional imaging capability, providing anatomical slices free from overlying information. However, even modern CT cannot compete with the spatial resolution capabilities of conventional radiography. A typical CT image matrix of 512 × 512 pixels compares poorly with the 4096 × 4096 pixel matrix of a digital chest radiograph. The real strength of CT lies in its ability to display structures which can be separated on the basis of their X-ray attenuation characteristics, such as bone, calcification, fresh haemorrhage, fat, air and tissue enhanced by contrast media.

CT scan times have shortened from about 5 min for a single slice in 1971 to less than a second for multiple slices in 2015. Volume imaging has been permitted by the introduction of continually rotating X-ray tubes and moving patient couches. X-ray detectors have progressed from single elements and single image slices to arrays of over 300 rows of detectors, permitting the simultaneous acquisition of data from multiple slices. New techniques for CT data acquisition and computation have improved image signal and reduced patient radiation dose. The real milestones in CT have been the introduction of spiral (helical) scanners in 1989 and multi-slice scanners in circa 1998.

4.3.1 CT Scanner Generations

There have been several generations or phases in CT scanner development since 1971 [2]. All modern CT units have many parallel rows of X-ray detectors, enabling them to acquire data from multiple imaging slices simultaneously. This technology is termed multi-detector CT (MDCT). However, the key elements of CT are still based on earlier third-generation designs, consisting of a rotating X-ray tube and a rotating array of X-ray detectors.

4.3.1.1 First-Generation (Translate-Rotate) CT

Hounsfield's original CT scanner involved a thin 'pencil' X-ray beam of parallel rays, which rotated around a patient into 180 projection positions and then translated sideways across the patient within each position. Hence, the scanner was termed a 'translate-rotate' design. Scan time was exceptionally slow by modern standards, at 5 min per slice, and the pixel matrix was coarse, consisting of 128 × 128 pixels in each two-dimensional axial slice.

4.3.1.2 Second-Generation CT

This design was based on a translate-rotate X-ray beam movement but introduced a diverging 'fan-shaped' X-ray beam. This enabled about 30 detectors to receive X-rays simultaneously and helped to shorten the scan time.

4.3.1.3 Third-Generation (Rotate-Rotate) CT

This development was introduced in circa 1977 and laid the foundations for modern CT scanners, which still use its rotate-rotate configuration.

This consisted of a rotating X-ray tube, linked to a rotating arc-shaped detector array which was located on the opposite side of the patient from the X-ray tube. The detector array contained up to 960 elements. The X-ray tube rotated 360° clockwise or anticlockwise and then had to reverse its motion, to avoid twisting the attached high voltage X-ray cables. This was still a single slice CT (SSCT) technology. The CT couch and patient were moved in increments through the X-ray beam as each slice was acquired in turn. Scan time was about 5 sec per slice.

4.3.1.4 Fourth-Generation CT

This design from circa 1980 proved to be a dead end in terms of development and did not lead to subsequent derivatives. Here the rotating X-ray tube was enclosed in a fixed circular array of detector elements, about 4800 in total. The intention was to avoid some circular image artefacts which could be associated with a rotating detector array. The X-ray tube still had to rotate in one direction then reverse its motion, and the design was still single-slice CT. Disadvantages inherent in the design included high cost, increased radiation dose and increased geometric unsharpness.

4.3.1.5 Fifth-Generation CT

In 1983 a specialist and non-mainstream CT scanner was introduced, designed to assess coronary artery calcification. This was a design based on electron beam CT (EBCT). It had a very original configuration in which an electron beam was rapidly swept by electromagnetic fields around the patient in a circular motion. X-rays were produced when the rotating electron beam struck a circular target track which surrounded the patient. The commercial name for the system was the Imatron, and its main advantage was very short scan times, permitting the freezing of coronary artery motion.

4.3.1.6 Sixth-Generation (Spiral) CT

A revolutionary improvement in CT technology was pioneered in 1989 by Willi Kalender and his team within Siemens. This was termed spiral or helical CT, so-called because the X-ray beam now prescribed a helical (corkscrew) pattern during the scan whilst the patient couch moved continuously through the scanner gantry. The speed of couch movement was referred to as the pitch factor. A major advantage was the possibility of volume scanning, enabling high-quality 3D images and superior scan reconstructions in other imaging planes from the axial scan data. This was enabled by the production of isotropic voxels (cubic voxels with sides of equal lengths in all three dimensions). Spiral CT required that the X-ray tube should be capable of continuous rotation in the same direction during scanning. Slip ring technology was the key development which allowed this, replacing high voltage X-ray tube cables by rotating rings connected electrically by conductive brushes. The X-ray tube and detector array were still rotate-rotate in design, with a rotating single row of detectors aligned opposite the X-ray tube. The single detector row meant that this was still a single slice design, as in previous generations of scanners. Scan times were about 3 sec per slice.

4.3.1.7 Seventh-Generation (Multi-slice) CT

Scan times could be further reduced if the spiral CT technology was coupled with multiple parallel rows of detectors, enabling data for several image slices to be acquired simultaneously. A four-slice design was launched in 1998 and was termed multi-slice or multi-detector CT (MDCT). Scanners with 64 slices were available by 2004, and the latest units may have over 300 slices. It is possible to combine detector elements in various ways to achieve different effective slice widths [3]. The broadening of volume coverage means that X-ray beams in such scanners are now cone shaped rather than fan shaped. It should be noted that the cone beam should not be confused with cone beam CT, which uses a rotating flat panel detector and is a different type of scanner. Minimum scan time per slice using MDCT is now about 0.1 s. There has also been a reduction in X-ray tube rotation time, from about 2 sec per 360° rotation in sixth-generation CT to less than 0.3 s per rotation in modern designs.

It should be noted that modern CT scanners are based on a rotate-rotate X-ray tube and detector array geometry, spiral scanning and multi-slice detector arrays [4]. The latest units utilise the incremental improvements that have occurred in previous scanner generations.

4.3.1.8 Dual-Energy CT

Many of the latest CT scanners are able to produce scans based on two distinct X-ray beam energies [5]. The advantage of this is that very precise information can be obtained about pixel and voxel X-ray attenuation (and hence Hounsfield unit values). This can enable more effective subtraction of unwanted tissues as well as functional studies based on the perfusion rates of contrast media into tissues. The subtraction technique can be useful for removing unwanted tissue such as faeces and image artefacts from CT colonography studies. There are a number of solutions for achieving dual-energy CT.

- Dual source. This technology uses two X-rays tubes, each operating at a different kVp, as well as two separate detector arrays. The X-ray tube kVp determines the peak and mean X-ray beam energy.
- KVp switching. Here a single X-ray tube is used, with rapid switching between two different kVs applied across it.
- Dual detector arrays. This solution uses a single X-ray tube and two superimposed layers of detectors, each absorbing X-rays of a different energy range.

4.4 Image Construction

In CT, a two-dimensional image matrix is obtained in the axial plane, typically consisting of 512 × 512 (which totals 262,144) pixels (picture elements). The X-ray attenuation value of each individual pixel must be calculated and then converted to a Hounsfield unit in order that a signal intensity can be assigned to it within the slice image. The mathematical calculations involved are huge and complex, because each pixel is contained in a column of other pixels. Individual pixel image intensities must be extrapolated from total X-ray linear attenuation values obtained from X-ray tracings [6]. This is only possible with a large amount of attenuation data obtained from multiple directions, each direction being an individual X-ray beam projection as the tube rotates circularly around a patient. A simple representation of this problem is shown in Fig. 4.5.

Let's look at a very simple situation based on only four pixels, which highlights the mathematical problems faced in producing a CT image. Let's say that we are trying to find the attenuation value for pixel A. Ray 1 provides a total attenuation value of four for pixels A and B combined. There could be many solutions to this – for example, $A=3$, $B=1$, $A=2$, $B=2$, $A=1$, $B=3$ and so on. We cannot assign a unique correct attenuation value to A on the basis of this one ray. Now let's add ray 2 data, based on another projection angle, which gives at total attenuation value of ten. Once again there are many possible solutions based on this data – for example, $A=1$, $C=9$, $A=2$, $C=8$, $A=3$, $C=7$ and so on. Let's add a third ray which gives a total attenuation value of six. Even with data from three rays, the attenuation value of pixel A is still unknown – it could be either 0, 1, 2, 3 or 4. Try inserting possible values yourself. Adding a further three rays, passing through pixels C-D, B-D and B-C would give enough data to solve the value of pixel A. This example is only illustrative, but it gives an idea of the difficulties involved in calculating pixel values using algebra. Imagine how much data would be needed for a 512 × 512 matrix.

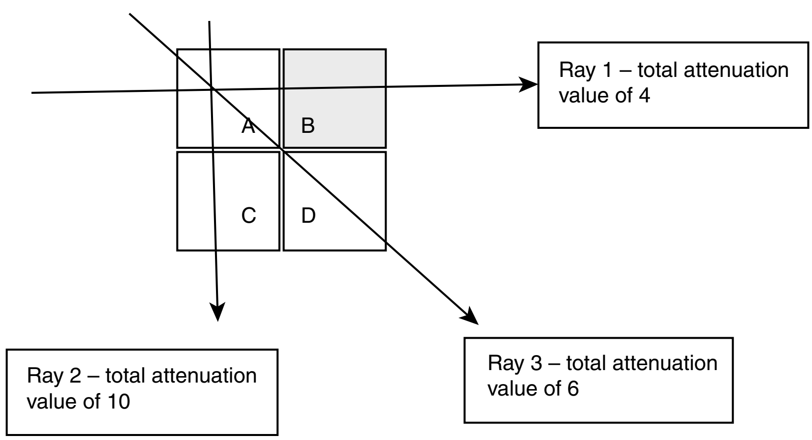

Fig. 4.5 The pixel attenuation calculation problem

A single X-ray transmission measurement through a patient's body made by an individual detector at a given point in time is called a ray or line. A series of rays that pass through a patient from the same X-ray perspective or orientation is called a projection. Modern CT scanners might use 600–1200 rays taken at 800–1500 projection angles. This is likely to give about 1,000,000 transmission measurements. This is a type of forward projection technique for image construction, since real X-ray attenuation data is linearly projected onto the detector array. In practice the algebraic calculation of pixel attenuation values is not feasible for large matrices, especially when random noise is present due to X-ray scatter, low X-ray intensity and electrical fluctuations in detectors.

4.4.1 Back Projection Methods

Back projection is a faster means of constructing the CT image from X-ray attenuation data. Attenuation values obtained in detectors are projected back along the ray paths into the image as shown schematically in Fig. 4.6. Each back-projected ray tracing uses the average value of the X-ray attenuation encountered along its linear path. The values are combined to produce summated signal where there is anatomical detail in the projection data. A disadvantage of the approach is image blurring. Attenuation information from the detectors is projected back into the patient to produce image signal. Note that this signal is summated where the projections overlap, thus indicating an X-ray attenuating object in the patient.

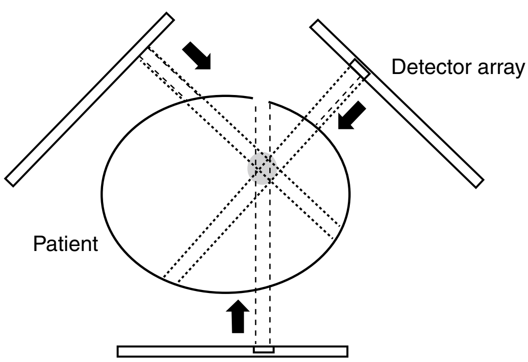

Fig. 4.6 Back projection in image construction. Attenuation information from the detectors is projected back into the patient to produce image signal

A process of filtration is applied to the back-projected image data in order to reduce blurring effects. This process is not the same as X-ray beam filtration. It involves a mathematical process known as convolution which tidies up the image and enhances the edges of objects. The mathematical filter applied is called a kernel. Soft and hard filters may be applied to CT data – soft filtration tends to be applied when soft tissue information is important and hard filtration when edge information is required. A hard filter improves image sharpness but may result in an increase in image noise.

4.4.2 Iterative Image Reconstruction

Iterative techniques are mathematical calculations that first make an assumption that all pixels have the same linear attenuation value and then repeatedly compare the real data situation with that assumption, making corrections. Mathematical recycling refines the image solution during the process. Back projection data is compared with a correct data model based on a forward projection through the patient. The mathematics is constantly refined through iteration until the two models agree, producing a less noisy image. The techniques have a lot of advantages, since iteration not only reduces image noise but also therefore permits lower exposure factors to be used whilst maintaining an acceptable noise level [7]. CT manufacturers have adopted slightly different techniques to achieve this, such as Adaptive Statistical Iterative Reconstruction (ASIR) by GE, iDose by Philips, Sinogram Affirmed Iterative Reconstruction (SAFIRE) by Siemens and Adaptive Iterative Dose Reconstruction (AIDR) by Toshiba.

4.4.3 Volume Rendering Techniques

Spiral CT brought about true volume imaging, since it enabled the production of isotropic voxels which are equal in size in all three dimensions. This permitted 3D datasets which achieve image reconstructions in any plane, since image resolution is the same in any plane. There are a number of techniques for rendering volume images from CT data, all of which can be rotated for viewing from multiple angles as follows.

- *Maximum intensity projection (MIP)*
  Anatomical structures are projected using ray paths from all directions. Image projection is based on the highest X-ray attenuation value that the ray encounters within its linear path. This is converted to an image signal intensity. The technique may be used when a contrast medium is introduced, since this will generally be the most attenuating material. Data are generally excluded from soft tissues which lie below a set cutoff value measured in Hounsfield units. However, high-density material such as tissue calcification may still show up and may obscure detail. Overlying high attenuation structures such as other vessels may interfere with vessel visualisation. The MIP approach produces a series of 2D images from multiple projection angles, and the 3D relationship between structures may be difficult to visualise. A variant of MIP is termed minimum intensity projection (MinIP) and can be useful for the visualisation of air or gas.
- *Surface shaded display*
  In this technique, the surface contours of structures such as vessels are displayed as if illuminated by an external simulated light source. Upper and lower Hounsfield unit threshold values are chosen, so that only tissues within these threshold limits are visualised. Each tissue type is allocated to a colour or transparency within the image, and surfaces are defined which represent boundaries between different tissues. The technique is limited by the fact that voxels with a mixed tissue composition cannot be properly represented since their Hounsfield value is averaged out. This can cause faulty representation of tissue boundaries. The technique is also vulnerable to image noise and variable contrast media enhancement.
- *Volume rendering*
  This approach provides a more accurate anatomical depiction since it can allow voxels to be displayed in terms of their true percentage composition of different tissue types rather than a false ‘all or nothing’ composition. Each voxel is given a colour or transparency within the 3D image, so that intermediate tissue compositions can be assigned values of hue or opacity. Simulated light rays are passed through the model so that it can be viewed. It is possible to view multiple tissue types such as bone, soft tissue and contrast-filled vessels within a single rendering. Also unwanted tissue types can be ‘peeled away’ if required.
- *Perspective volume rendering (pVR)*
  This is a novel way of displaying volume data and enables a virtual fly-through to be obtained within the hollow cavities such as the bowel and vessels. Internal structures and walls are assigned colours or transparencies for better visualisation, using perspective images which simulate the presence of the observer within the structure. There are applications for virtual colonoscopy (also called CT colonography).

4.5 CT Image Resolution

Resolution can be thought of as the ability to resolve small objects or detail. It is affected by a number of factors. It should be noted that many of the factors which lead to increased resolution may also result in increased image noise. Here are some of the factors.

- *Slice thickness*
  In single-detector row CT, the slice thickness was principally determined by the X-ray beam collimator aperture. But in multi-detector row CT, the slice thickness is affected by combining detector elements in the z direction (along the long axis of a supine patient). Effective slice thickness can be measured using the slice sensitivity profile, which is the measured slice width taken at 50 % of the maximum signal value across the slice using a quality assurance phantom. Pitch factor affects the slice sensitivity profile (and thus the slice width) by widening it in the z-direction during table movement. It should be noted that a perfect slice sensitivity profile would be rectangular, but factors such as pitch and X-ray beam geometry tend to convert it to a bell-shaped curve as shown in Fig. 4.7. Small slice thicknesses provide a high spatial resolution and suffer less from 'partial voluming'. This means that they are more able to accurately depict small objects which might otherwise become averaged out within thicker slices.
- *Detector size*
  This is often a limiting factor with regard to resolution. It is affected by the detector element aperture as well as the spacing between the detector elements. In multi-detector row CT, there tends to be increased X-ray scatter due to the width of the cone-shaped X-ray beam. Thus, there needs to be thicker septa (dividers) between the detector elements in order to prevent stray scatter from entering. The smallest detector elements tend to be about 0.5 mm in aperture. Detector rows may be combined in multi-detector row CT in order to provide a range of possible slice thicknesses.
- *Geometric effects (penumbra)*
  These are due to the diverging nature of the X-ray beam. Resolution is increased by a small X-ray tube focal spot size, large X-ray tube to patient distance and a small patient to detector distance. In addition penumbra effects can cause a drop off in the received X-ray intensity at the beam edges. This is less of a problem in multi-detector row CT that uses more than 16 detector rows.
- *Scan field of view and pixel matrix*
  Pixels of smaller physical dimensions will provide increased spatial resolution. This can be achieved via a 'finer' pixel matrix, such as a 512 × 512 array rather than 256 × 256, as well as a small scan field of view.

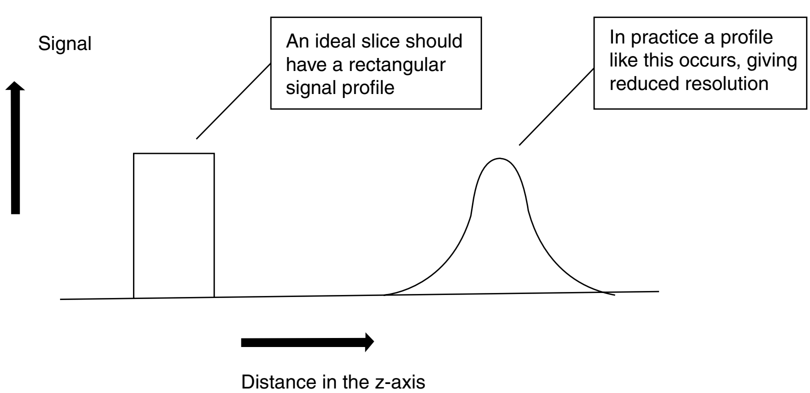

Fig. 4.7 Slice sensitivity profiles

4.6 Key Messages

- Computed tomography (CT) utilises the attenuation of X-ray photons to produce sectional information about human body structures. X-ray absorption results in a useful pattern of information in the X-ray detector array. X-ray scatter results in non-useful information and increases image noise.
- A sectional image in CT is based on many X-ray attenuation measurements based on multiple projections and many ray tracings obtained during a rotation of the X-ray tube around the patient's body. This presents a complex mathematical problem which is solved via back projection and iterative image reconstruction.
- Signal intensity (brightness) values within individual voxel elements are obtained via the conversion of X-ray attenuation data to Hounsfield units or CT numbers. Zero on the Hounsfield scale corresponds to the X-ray attenuation value for water.
- The main advantage of CT relative to conventional X-ray imaging is its ability to amplify image contrast between tissues. This is achieved via removal of overlying anatomy, X-ray scatter reduction and the use of an extended scale in which the small X-ray attenuation differences between anatomical features are amplified.
- Relative to MRI, CT provides improved spatial resolution, faster scan times, poorer soft tissue discrimination and reduced functional imaging capability. Other advantages of CT include its ability to depict cortical bone, calcification, fresh bleed and air.
- Progressive technological developments in CT have resulted in shorter scan times, improved spatial resolution, volume imaging, multiplanar reconstructions, improved soft tissue characterisation and some functional imaging capability. This has enabled CT colonography and many other examinations which would not have been possible in the early days of CT.
- Modern CT scanners use a helical (spiral) rotation of the X-ray tube relative to a moving X-ray couch, together with multi-slice imaging based on the presence of many parallel rows of X-ray detectors.
- Increase in the kVp (kilovoltage peak) setting results in increased X-ray photon energy and an increased number of photons, increased signal-to-noise ratio, reduced soft tissue contrast and increased radiation dose.
- Increase in the mAs (milliampere seconds) setting results in an increased number of X-ray photons, increased signal-to-noise ratio and increased radiation dose.

4.7 Summary

Computed tomography (CT) is a powerful technique for providing sectional 'slices' of the human body, free from the obscuring presence of overlying structures. The physical principles of CT enable improved image contrast, by accentuating the differential X-ray attenuation that occurs within different body tissues and translating this into the Hounsfield scale of signal intensities. Progressive technical developments since the early 1970s have resulted in sub-second scan times, 'freezing' of involuntary patient motion, improved spatial resolution, volume acquisition and 3D display. These assets have greatly advanced the non-invasive radiological examination of the colon.

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