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# X-ray computed tomography

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**Abstract**

X-ray computed tomography (CT) is a medical imaging technique that produces images of transaxial planes through the human body. When compared with a conventional radiograph, which is an image of many planes superimposed on each other, a CT image exhibits significantly improved contrast although this is at the expense of reduced spatial resolution.

A CT image is reconstructed mathematically from a large number of one-dimensional projections of the chosen plane. These projections are acquired electronically using a linear array of solid-state detectors and an x-ray source that rotates around the patient.

X-ray computed tomography is used routinely in radiological examinations. It has also been found to be useful in special applications such as radiotherapy treatment planning and three-dimensional imaging for surgical planning.

## Introduction

In an x-ray radiograph, e.g. a chest x-ray as shown in figure 1, the three-dimensional (3D) structure of the body is represented by a two-dimensional (2D) image. During the imaging process one of the dimensions is lost. All of the planes in the patient that are parallel to the x-ray film are superimposed on top of each other. In figure 1 one can see ribs, soft tissues and lung all overlying each other. The image is a projection of a 3D volume onto a 2D surface.

Because of the superimposition of structures, radiographs do not exhibit high contrast. Bone and air cavities are easily seen but very little contrast is obtained in soft tissue regions, e.g. a blood vessel surrounded by muscle will not be seen. In some situations this lack of contrast can be overcome by using contrast media containing elements such as iodine and barium which highly attenuate the x-rays. For example, a liquid contrast medium injected into the bloodstream is used in angiography to make the blood vessels visible.

X-ray computed tomography (CT) is an imaging technique which overcomes the problem of poor contrast and the inability of radiographs to provide depth information. It does this by producing a 2D image of a 2D plane (slice) through the patient as shown in figure 2. Because there is no longer superposition of structures on top of each other, contrast in the image is greatly improved. Tomography refers to any technique that produces an image of a single plane in the body. A description of positron emission tomography can be found elsewhere in this issue (Badawi 2001).

## The production and detection of x-rays

In an x-ray tube, electrons, emitted by thermionic emission from a cathode consisting of a heated filament and a focusing cup, are accelerated by a high voltage and focused into a beam so that they impinge on a target which forms part of an anode (figure 3). These electrons are referred to as projectile electrons. X-rays are produced by the sudden deceleration of the projectile electrons by the nuclei of the target material or by collision of projectile electrons with electrons in the target atoms. These are referred to as bremsstrahlung x-rays and characteristic x-rays respectively.

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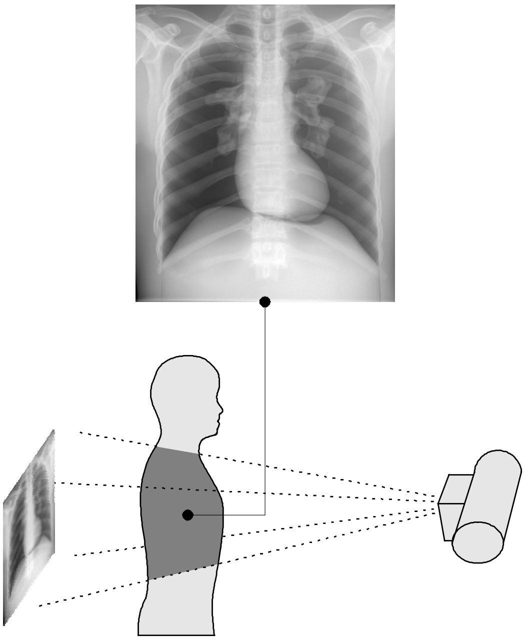


**Figure 1.** A chest x-ray.

Bremsstrahlung x-rays are emitted when an electron slows down by coulombic interaction with the electrostatic field of a target nucleus. When the electron slows down it loses kinetic energy, which is converted to electromagnetic radiation, i.e. an accelerated (or decelerated) charge will radiate electromagnetic radiation.

The amount of energy radiated depends on the closeness of approach of the projectile electron to the target nucleus. Therefore bremsstrahlung x-rays can have a range of energies from 0 up to a maximum equal to the energy of the projectile electrons. Thus the maximum possible x-ray energy (in eV) is numerically equal to the accelerating electric potential, e.g. if the x-ray tube is operated at 50 kV the maximum x-ray energy is 50 keV. Bremsstrahlung radiation is also called *white radiation* because of the broad continuous spectrum of x-ray energies produced.

Characteristic x-rays arise from vacancies that are created in the electron shells when the target atoms are ionized by the projectile electrons. Electrons from outer shells drop down to fill the vacancy and release their excess energy in the form of x-rays. Characteristic x-rays are monoenergetic. Their energies depend on the binding energies of the electrons in the electron shells, which in turn depend on the type of the atoms in the anode.

Both bremsstrahlung and characteristic x-rays are produced simultaneously. An example spectrum for a tungsten target is shown in figure 4. The x-ray tubes used in CT are typically operated at potentials in the range 100 kV to 150 kV.

Usually a projectile electron does not lose all its energy in one interaction. It undergoes multiple collisions with the target atoms. Most of these collisions occur with outer electrons of the target atoms and produce mainly heat. Only

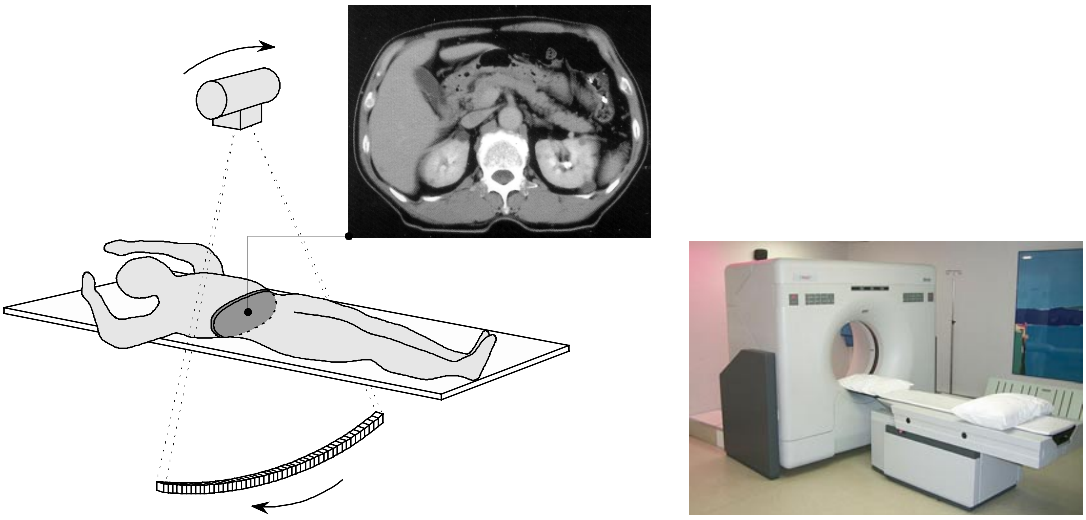


**Figure 2.** X-ray computed tomography.

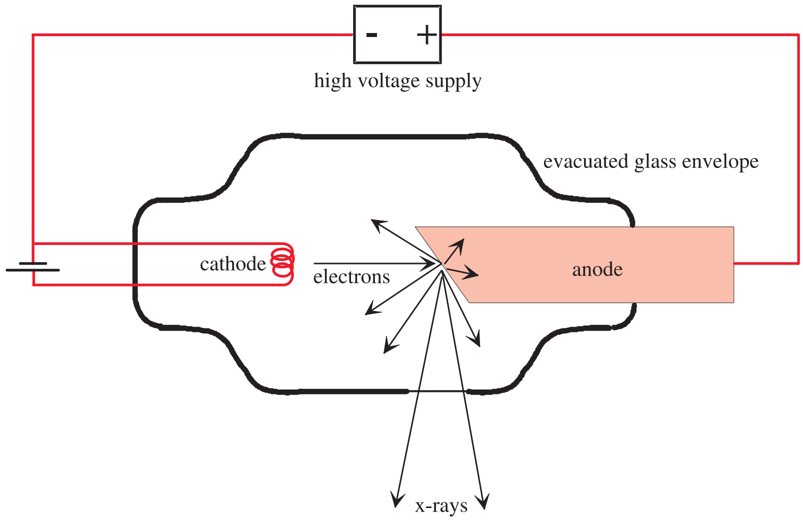



**Figure 3.** Schematic diagram of a stationary anode x-ray tube. X-rays are emitted in all directions and must be collimated by lead shielding (not shown) to produce a useful beam.

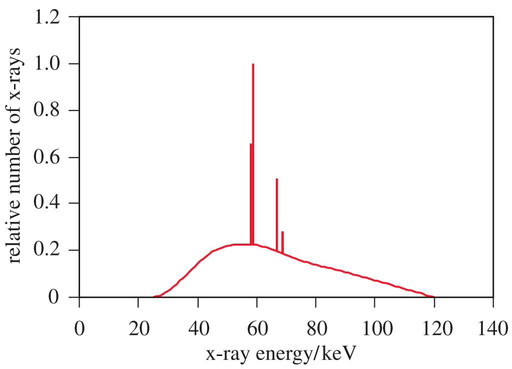



**Figure 4.** A typical x-ray spectrum for a CT x-ray tube operated at 120 kV.

a small proportion of projectile electrons undergo interactions that produce x-rays.

The efficiency of x-ray production is given by (Krestel 1990)

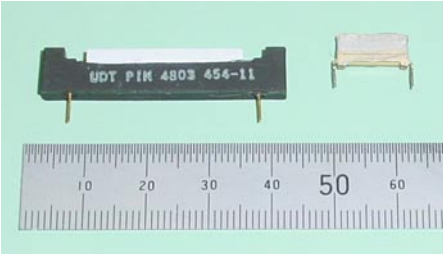
$$\eta = aUZ$$

where $U$ is the accelerating potential (V), $Z$ is the atomic number of the target material, and $a$ is a constant equal to $1.1 \times 10^{-9}$ ($V^{-1}$).

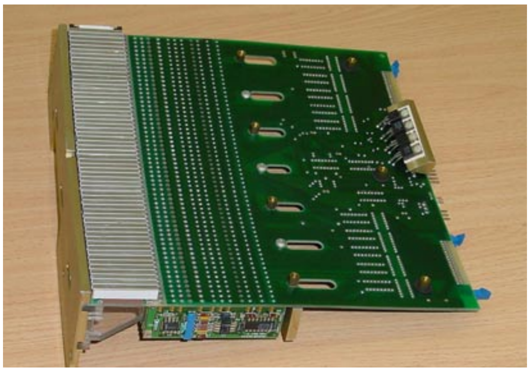
For a tungsten target ($Z = 74$) and an accelerating potential of 100 kV the efficiency is 0.0081, i.e. $< 0.1\%$. Most of the electron energy is converted to heat. Therefore many of the design features of x-ray tubes are for the dissipation of large quantities of heat. These design features include a line focus and rotating anode to distribute the heat over a large target area. A typical rotating anode medical x-ray tube is shown in figure 5. X-ray tubes used in CT scanners must withstand heavy loads and incorporate additional features such as large diameter compound anodes, metal envelopes and circulated cooling fluid.

**Figure 5.** A typical rotating anode x-ray tube.

The most common detectors used in CT scanners are scintillator–photodiode solid state detectors. Some of the materials used as scintillators are caesium iodide (CsI), cadmium tungstate ($CdWO_4$) or ceramic materials based on yttrium–gadolinium oxides. The x-rays interact with the scintillator and produce visible light. This light is then converted to an electric current by the photodiode. Some features of these detectors are high detection efficiency, good stability, small size and high packing density. Examples are shown in figure 6.

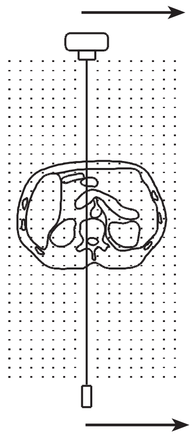



(*a*)



(*b*)

**Figure 6.** (*a*) Two solid state CT detectors. The one on the left is a $CdWO_4$ detector from a fourth generation CT scanner. The other is a CsI detector from a third generation scanner. The scale units on the ruler are mm. (*b*) A board from a fourth generation CT scanner containing 60 of the larger detectors from (*a*). Here they are shown, on the left-hand side of the board, with a protective aluminium covering. This particular scanner has 20 such boards containing a total of 1200 detectors in a full circle.



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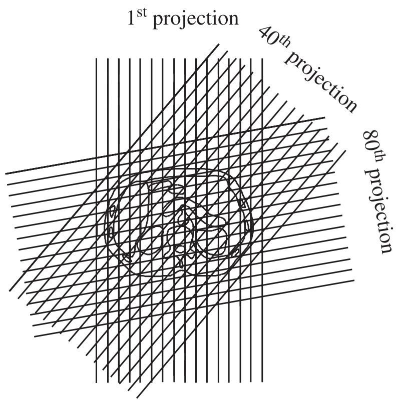


(*b*)

**Figure 7.** (*a*) The pencil-beam of x-rays is scanned across the patient to acquire a projection that is made up of a large number of raysums. (*b*) A large number of projections are acquired, each at a different angle.

## The measurement process

Unlike the x-ray detector used in radiography (film) the x-ray detectors in a CT scanner do not directly produce an image. They make measurements of the transmission of a narrow (1 to 10 mm) fan-beam of x-rays through the chosen slice (figure 2) from many different directions and an image is reconstructed mathematically from these measurements. The depth information along the direction of the x-ray beam that is lost in radiography is recovered by viewing the slice from many different directions. This is analogous to using triangulation to determine the distance to a faraway object: the angle between a baseline and the object must be measured from the two ends of the baseline, i.e. views from two different directions are required.

The measurements made by a CT scanner are most easily explained using a pencil-beam of x-rays (a fan-beam can be considered to be made up of many adjacent pencil-beams). The beam of x-rays is translated across and rotated about the patient. Figure 7 shows the acquisition being made up of a large number of projections (views) taken at different angles around the patient. Each projection is made up of a large number of pencil-beam attenuation measurements. Each pencil-beam measurement is referred to as a raysum.

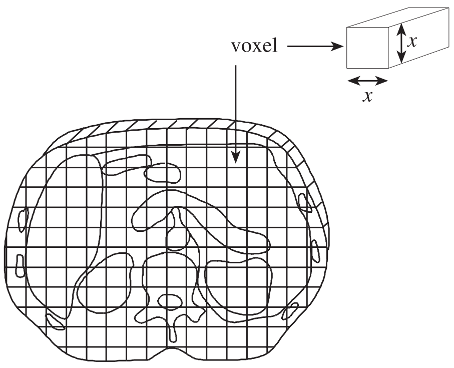


**Figure 8.** Each pixel in the image represents a voxel of tissue in the patient.

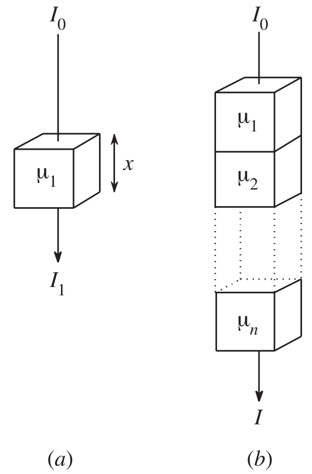


**Figure 9.** (*a*) The transmission of an x-ray beam through a single voxel. (*b*) A raysum is the sum of attenuation coefficients along the path of a single ray through the patient.

Mathematically, a measurement made by a CT detector is proportional to the sum of the attenuation coefficients that lie along the ray defined by the pencil-beam of x-rays, hence the term raysum.

The CT scanner, being an inherently electronic imaging device, produces a digital image that consists of a square matrix of picture elements, i.e. pixels. Each of the pixels in the image represents a voxel (volume element) of tissue in the patient. This is depicted in figure 8.

Assuming that the x-rays are monoenergetic then the intensity $I_1$ of an x-ray beam of incident intensity $I_0$ transmitted beam through a small volume of tissue having thickness $x$ and attenuation coefficient $\mu_1$ is

$$I_1 = I_0 \exp(-\mu_1 x) \tag{1}$$

This is depicted in figure 9(*a*).

In traversing from one side of the patient to the other, the x-ray beam will be attenuated by all of the voxels through which it passes (refer to figure 9(*b*)). The emerging x-ray beam will have an intensity $I$ given by

$$I = I_0 \exp\left(-x \sum_{i=1}^{n} \mu_i\right). \tag{2}$$

This can be rearranged to give

$$\ln \frac{I_0}{I} = x \sum_{i=1}^{n} \mu_i . \tag{3}$$

Thus, the natural logarithm of the ratio of incident to transmitted x-ray intensities is proportional to the sum of the attenuation coefficients of the voxels

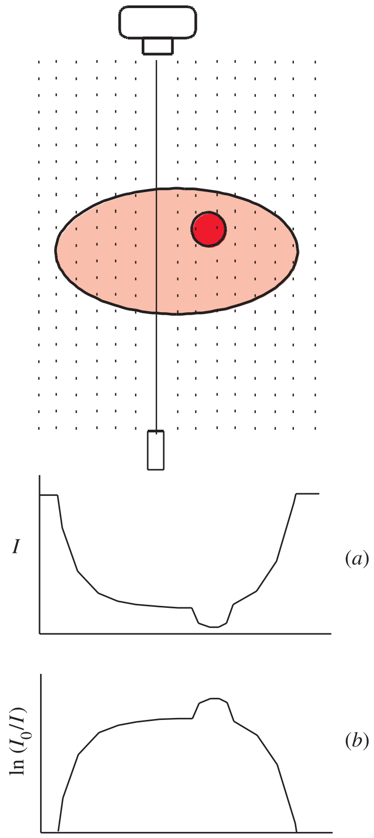


**Figure 10.** A single projection shown as (*a*) an intensity profile and (*b*) an attenuation profile.

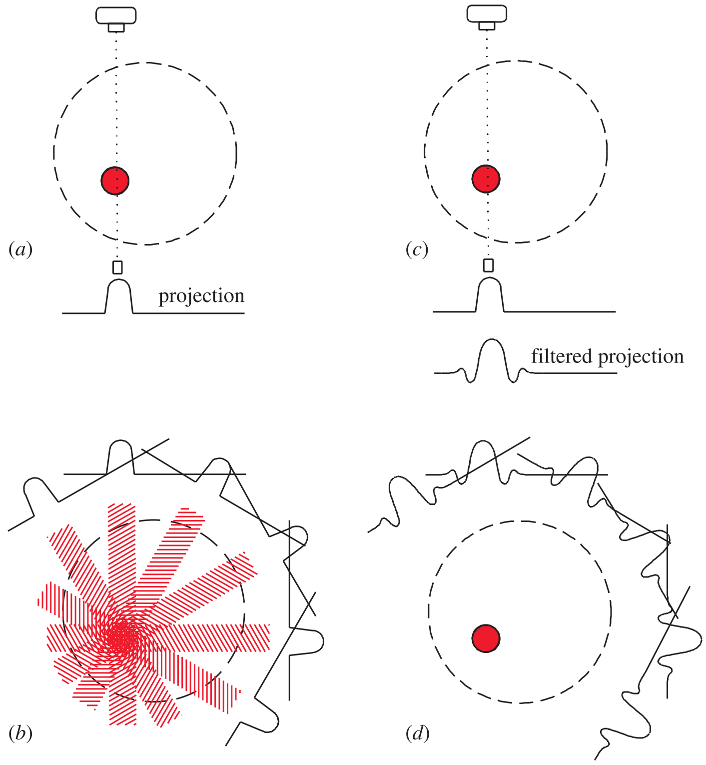


**Figure 11.** The methods of simple back-projection and filtered back-projection. In simple back-projection the measured attenuation profiles (*a*) are simply spread out, or projected, across the image plane (*b*). In filtered back-projection the attenuation profiles are convolved with a filter function (*c*) before being back-projected (*d*).

in the path of the beam. The measured x-ray intensities and the corresponding logarithms of the intensity ratios are shown for one projection in figure 10. The graph of the logarithms of the intensity ratios is often referred to as an attenuation profile.

Since the x-rays produced by an x-ray tube are polyenergetic, equation (3) is only an approximation. As the beam passes through the patient the lower energy x-rays are preferentially absorbed and the average energy of the beam increases. This effect is referred to as *beam hardening* and it leads to underestimation of the pixel values in the image. A number of algorithms (for example, McDavid *et al* 1977) have been developed to correct for beam hardening. They achieve this by modifying the raysums prior to image reconstruction.

## Image reconstruction

Consider a CT image that consists of 512 rows each containing 512 pixels, i.e. a square matrix with a total of 262 144 pixels. This is a common image format for current scanners. The image reconstruction process must calculate a value of the attenuation coefficient, $\mu_i$, for each of the 262 144 voxels corresponding to these pixels.

One possible method is to measure 262 144 raysums (i.e. 512 projections each containing 512 raysums) so that we have 262 144 equations in the form of equation (3). These 262 144 simultaneous equations can then be solved for the 262 144 unknown values of $\mu_i$. Solving this many simultaneous equations is not an easy task and is complicated by the fact that the x-rays rarely follow paths that correspond to rows or columns of pixels as depicted in figure 9(*b*). Fortunately there is a better way to reconstruct the image.

The method used on all current CT scanners is the method of *filtered back-projection*. The mathematics of filtered back-projection dates back to 1917 with the work of the mathematician, J Radon. Webb (1990) gives an excellent account of the history of radiological tomography.

Filtered back-projection is best explained by first describing the process of *simple back-projection*.

Simple back-projection is depicted in figures 11(*a*) and 11(*b*). A single attenuation profile (projection) lacks any depth information and so the value of each raysum is shared out equally among all of the voxels through which the ray passed, i.e. the raysums are back-projected. As shown in figure 11(*b*), as each attenuation profile is back-projected an image, albeit a poor one, is formed. Each point in the image is surrounded by a star-burst pattern which degrades contrast and blurs the edges of objects.

The degrading star-burst patterns of simple back-projection can be removed by convolving the attenuation profile with a filter function prior to back-projection (see figure 11(*c*)). Thus the term filtered back-projection. The action of the filter function is such that the negative values created in the filtered projection will, when back-projected, exactly cancel out the star-burst patterns and produce an image that is an accurate representation of the original object (figure 11(*d*).)

An additional advantage of filtered back-projection is that image reconstruction does not need to wait until all of the projections have been acquired. As soon as the first set of measurements are made by the x-ray detectors they can be pre-processed (i.e. calculate the logarithm of the intensity ratios), filtered and then back-projected. Image reconstruction can proceed simultaneously with scanning.

The mathematics of filtered back-projection and other methods of image reconstruction are described in a review article by Brooks and Di Chiro (1976) and in detail by Kak and Slaney (1988). The book by Kak and Slaney is available free of charge (for personal use only) in electronic form on the internet at www.slaney.org/pct/.

## Scanning geometries

The x-ray tube and detector geometry depicted in figure 7 is referred to as *first generation*. Only the very first CT scanners (early 1970s) were of this configuration. Because of the single detector and the combination of both translate and rotate motions, scan times were very long: of the order of a few minutes.

Second generation CT scanners (figure 12) were developed to decrease scan times. They had a narrow fan-beam of x-rays and a small number of detectors. Multiple projections were acquired simultaneously (one per detector) during each translation and therefore the number of translations required was reduced accordingly.

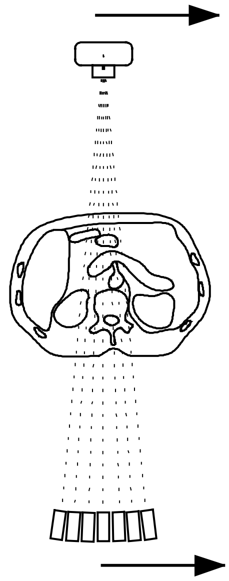

**Figure 12.** Second generation CT scanner.

Nearly all current CT scanners use either the third or fourth generation geometries. Third generation is that shown in figure 2, i.e. the fan-beam of x-rays is wide enough to cover the full width of the patient and therefore no translation motion is required. The x-ray tube and detector arc rotate only. The arc of detectors may contain up to 1000 individual detectors.

In fourth generation scanners the detectors form a complete ring around the patient and only the x-ray tube has to rotate. These scanners may use up to nearly 5000 individual detectors. A fourth generation CT scanner is depicted in figure 13.

The fifth generation of CT scanners has eliminated all mechanical motion by using an x-ray tube with an anode that forms a circular arc of about 210° around the patient. The scanning motion is achieved by using magnetic fields to sweep the electron beam along the anode. This scanner can acquire a complete set of projection data in as little as 50 ms and is designed specifically for cardiac imaging.

For the reader who wishes to investigate these geometries and methods of image reconstruction further, an excellent software simulation of a CT

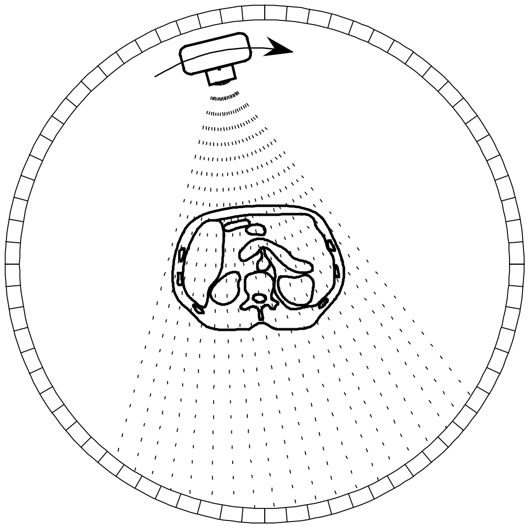

**Figure 13.** The fourth generation geometry.

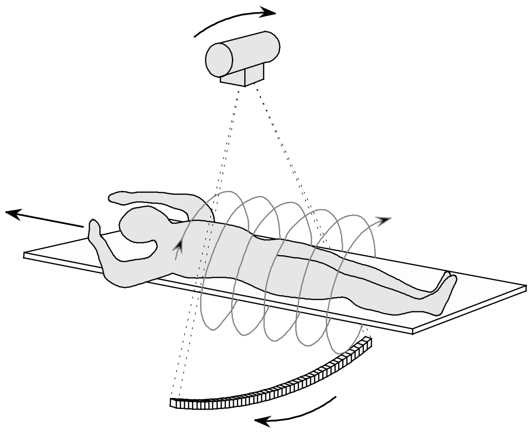

**Figure 14.** Helical computed tomography.

scanner has been written by Kevin M Rosenberg and can be downloaded free of charge from the internet at www.ctsim.org.

## Helical CT

The conventional way in which CT is used is to keep the patient stationary during the scanning process. Scanning is then halted while the patient is positioned, using a motorized table-top, for the next scan. During chest and abdominal scans it is necessary for the patient to hold his/her breath during scanning. Therefore, although one image may be acquired in as little as 0.5 s, a procedure requiring 20 to 30 images may take many minutes to complete.

In helical CT (also referred to as spiral CT and volume scanning) the x-ray tube and detectors use slip-rings, rather than cables, for their electrical connections so that they may rotate continuously without stopping. In acquiring multiple images, the x-ray tube and detectors rotate continuously while the patient is translated perpendicular to the plane of rotation (figure 14). The relative motion of the x-ray tube and the patient is that of a helix. Helical CT scanners can scan a large volume of the patient during a single breath-hold. All the acquired projection data are stored on the computer's hard disk and the images are reconstructed retrospectively.

The most recent advancements in CT scanning are multi-slice helical scanners. These scanners have up to four parallel banks of detectors so that they acquire multiple slices simultaneously. With a complete set of projection data being acquired in 180° rotation, two complete rotations per second and four banks of detectors, the latest multi-slice helical CT scanners can acquire up to 16 images per second.

Further information on current CT scanner technology can be obtained from manufacturers' internet sites, e.g. www.gemedicalsystems.com, www.Picker.com, www.SiemensMedical.com and www.Toshiba.com. The GE Medical Systems site also has a good collection of CT images.

## Image display

Before the image is displayed, the CT scanner converts the measured attenuation coefficients of the voxels to CT numbers. For a voxel, $i$, the CT number, $N_i$, of the corresponding pixel is related to the attenuation coefficient, $\mu_i$, of the tissue in the voxel by

$$N_i = 1000 \times \frac{\mu_i - \mu_w}{\mu_w} \qquad (4)$$

where $\mu_w$ is the attenuation coefficient of water. Expressing the attenuation coefficient relative to that of water dates back to the first clinical CT scanner, the EMI Mark 1 head scanner of 1972. The detectors of this scanner had a low dynamic range. Therefore the patient's head was surrounded by a water bag to reduce the very high x-ray intensity of the unattenuated x-ray beam that would otherwise reach the detectors. The scaling factor of 1000 in equation (4) is

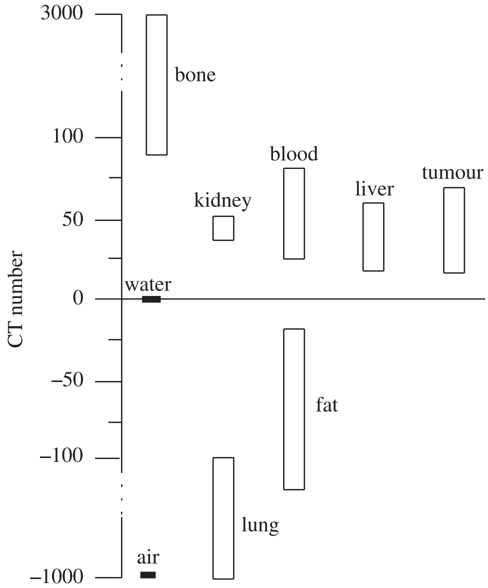


**Figure 15.** Range of CT numbers for some tissue types.

**Figure 16.** CT scan of the abdomen with a display window optimized for the liver.

used so that the CT numbers can be expressed as integers. This reduces computer memory and storage requirements.

From equation (4) it can be seen that the CT number of water will be zero, the CT number of air ($\mu \approx 0$ cm$^{-1}$) will be $-1000$, and a voxel with tissue having an attenuation coefficient twice that of water (e.g. bone) will have a CT number of +1000. Most soft tissues of the human body lie in the range $\pm100$. Figure 15 shows the typical range of CT numbers for a number of tissues.

CT images are usually viewed as greyscale images on a computer monitor. The contrast in the image can be further enhanced by displaying only a window of CT numbers. Rather than allocating the full range of CT numbers to the range of available grey levels (from black to white) only a limited range of CT numbers is displayed. For example, for the brain, only the range of CT numbers from 0 to 70 might be displayed, whereas for the lung, a window from $-1000$ to $-200$ might be used. Figure 16 shows an image of the abdomen which uses a narrow window centred on the range of CT numbers appropriate for the liver. Pixels with CT numbers outside this range, such as bone or fat, appear as either pure white or pure black.

## Applications

Computed tomography is used routinely in radiological examinations because it provides cross-sectional images and improved contrast compared with other x-ray imaging techniques. Its main limitation compared with x-ray film is poor spatial resolution. The minimum object size visible with CT is about 0.5 mm compared with about 0.05 mm for x-ray film.

Special applications of CT include bone mineral content measurement, cerebral blood flow using diffusion of inhaled xenon, the use of stereotactic frames to assist in accurate location for needle biopsies, and 3D imaging. Pseudo-3D displays are created from a stack of 2D images of a large number of parallel planes. An example is shown in figure 17. Clinical applications of 3D imaging include surgical planning, prosthesis design, craniofacial reconstructions, stereotactic biopsy planning and radiotherapy treatment planning.

CT is useful for radiotherapy treatment planning for a number of reasons, i.e. high soft tissue contrast provides good visualization of tumours, the images provide accurate spatial information for treatment planning, and the pixel values provide radiological properties of the tissues that can be used for inhomogeneity corrections in dose calculations. More information on the application of CT in radiotherapy is provided by Oldham (2001).

## Conclusion

Despite the rapid growth of magnetic resonance imaging (MRI) in the past decade, MRI has not superseded CT. CT is still used routinely in radiological examinations and is, itself,

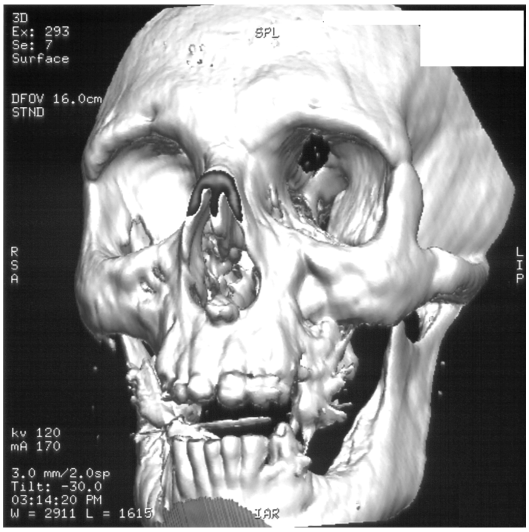


**Figure 17.** A pseudo-3D image of a skull, computer generated from a stack of 2D images of a large number of parallel slices.

undergoing rapid growth and development in the form of multi-slice helical x-ray computed tomography.

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