

Computed Tomography

Scientific/Technological fact sheet

Introduction

In conventional X-ray imaging, the entire thickness of the body is projected on a film: structures overlap and are difficult to distinguish (see: Scientific/Technological fact sheet – X-ray Photography). One of the problems is the loss of information about depth. Suppose a small lung carcinoma can be seen at a front-to-back chest photograph (Figure 1). But where is it? The radiologist cannot determine the exact location of this carcinoma in the forward-backward direction. One could make a lateral photograph (a side view), but the carcinoma might disappear behind a rib. What is needed in such a case is a cross-sectional image (Figure 2). This became possible when Geoffrey N. Hounsfield presented the first CT scanner in 1972.



Figure 1 – Typical AP (anterior-posterior or front-to-back) chest X-ray photograph.



Figure 2 – Transversal CT slice of the chest.

This new technique of *computed tomography* (CT) reconstructs a cross-sectional image of the body from a 'virtual pile of X-ray photographs'. A tomographic image is an image of a *slice* through the body. The word 'tomography' comes from the Greek: *tomos* means slice, *graphein* stands for 'to write'. So, tomography literally means 'writing slices'. Structures and lesions previously impossible to visualise can now be seen with remarkable clarity. The principle behind CT is shown in Figure 3: a thin collimated beam of X-rays passes through the body to a detector that measures the transmitted intensity. The collimator is a set of narrow lead tubes or an array of small holes in a lead plate, resulting in a thin straight beam of X-rays. Measurements are made at a large number of points as the source and detector are moved past the body together. The apparatus is then rotated slightly about the body axis and again scanned. This is repeated at, for example, 1° intervals for 180° . The intensity of the transmitted beam for the many points of each scan, and for each angle, are sent to a computer that reconstructs the image of the slice. The image is presented on a computer monitor.

Note that the imaged slice is perpendicular to the long axis of the body. For this reason, CT is sometimes called *computed axial tomography* (CAT), although the abbreviation CAT, as in CAT scan, can also be read as *computer-assisted tomography*.

But how is the image formed? We can think of the slice to be imaged as being divided into many tiny picture elements or *pixels*, which could be squares as in Figure 4. For CT, the width of each pixel is chosen according to the width of the detectors and/or the width of the X-ray beams. This determines the resolution of the image, which is typically about 2 mm. An X-ray detector measures the intensity of the transmitted beam after it has passed through the body. Subtracting this value from the intensity of the beam at the source, we get the total absorption. Note that only the total absorption along each beam line can be measured: the absorption by all the pixels in a line. To form an image, we need to

determine how much radiation is absorbed at *each* pixel – how that can be done will be discussed below. We can then assign a ‘greyness value’ to each pixel according to how much radiation was absorbed. The image, then, is made up of tiny spots (pixels) of varying shades of grey, as is a black-and-white television picture. Often the amount of absorption is colour-coded. The colours in the resulting ‘false colour’ image have nothing to do, however, with the actual colour of the object.

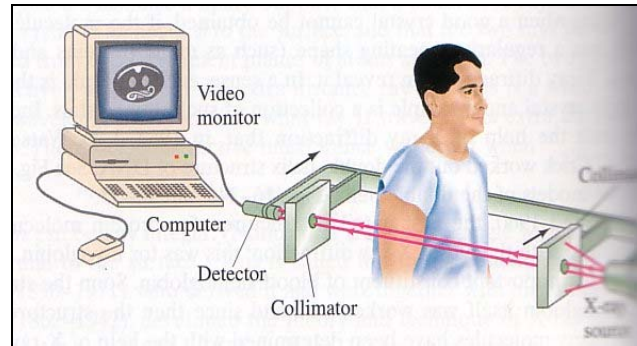


Figure 3 – Tomographic imaging with an X-ray beam.

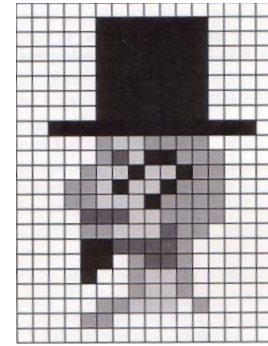


Figure 4 – Example of an image made up of many small squares called pixels (picture elements). This one has rather poor resolution.

Finally, we must discuss how the ‘greyness’ of each pixel can be determined even though all we can measure is the total absorption along each beam line in the slice. It can be done only by using the many beam scans made at a great many different angles. Suppose the image is to be an array of 100×100 elements for a total of 10^4 pixels. If we have 100 detectors and measure the absorption projections at 100 different angles, then we get 10^4 pieces of information. From this information, an image can be reconstructed, but not precisely. If more angles are measured, the reconstruction of the image can be done more accurately.

There are a number of mathematical reconstruction techniques, all of which are complicated and require the use of a computer. To suggest how it is done, we consider a very simple case using the so-called *iterative technique*. The verb ‘to iterate’ is derived from the Latin expression for ‘to repeat’. Although this technique is less used now than the more direct ‘Fourier transform’ and ‘back projection’ techniques, it is the simplest to explain. Suppose our sample slice is divided into the simple 2×2 pixels as shown in Figure 5. The number in each pixel represents the amount of absorption by the material in that area

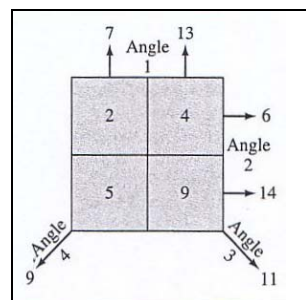


Figure 5 – A simple 2×2 image showing true absorption values and measured projections. Note that in real situations, the true absorption values are not known. These values have to be calculated from the measured projections.

(say, in tenths of a percent): that is, 4 represents twice as much absorption as 2. But we cannot directly measure these values – they are the unknowns we want to solve for. All we can measure are the projections – the total absorption along each beam line – and these are shown in the diagram as the sum of the absorptions for the pixels along each line at four different angles. These projections (given at the tip of each arrow) are what we can measure, and we now want to work back from them to see how close we can get to the true absorption value for each pixel. We start our analysis with each pixel being assigned a zero value (Figure 6a). In the iterative technique, we use the projections to estimate the absorption value in each pixel, and repeat for each angle. The angle 1 projections are 7 and 13. We divide each of these equally between their two pixels: each pixel in the left column gets $3\frac{1}{2}$ (half of 7), and each pixel in the right column gets $6\frac{1}{2}$ (half of 13) (Figure 6b). Next we use the projections at angle 2. We

calculate the difference between the measured projections at angle 2 (6 and 14) and the projections based on the previous estimate (top row: $3\frac{1}{2} + 6\frac{1}{2} = 10$; same for bottom row). Then we distribute the difference equally to the pixels in that row. For the top row we then

have (left and right, respectively): $3\frac{1}{2} + (6 - 10)/2 = 1\frac{1}{2}$ and $6\frac{1}{2} + (6 - 10)/2 = 4\frac{1}{2}$. And for the bottom row (left and right, respectively): $3\frac{1}{2} + (14 - 10)/2 = 5\frac{1}{2}$ and $6\frac{1}{2} + (14 - 10)/2 = 8\frac{1}{2}$. These values are inserted as shown in Figure 6c. Next, the projection at angle 3 gives (upper left and lower right, respectively): $1\frac{1}{2} + (11 - 10)/2 = 2$ and $8\frac{1}{2} + (11 - 10)/2 = 9$. Finally, the projection at angle 4 gives (lower left and upper right, respectively): $5\frac{1}{2} + (9 - 10)/2 = 5$ and $4\frac{1}{2} + (9 - 10)/2 = 4$. The result, shown in Figure 6d, corresponds exactly to the true values in Figure 5.

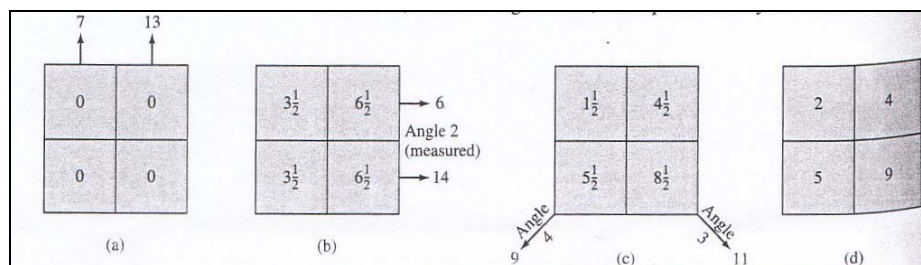


Figure 6 – Reconstructing the image using projections in an iterative procedure.

To obtain these four numbers exactly, we used six pieces of information (two each at angles 1 and 2, one each at angles 3 and 4). For the much larger number of pixels used for actual images, exact values are generally not attained. Many iterations may be needed, and the calculation is considered sufficiently precise when the difference between calculated and measured projections is sufficiently small. The above example illustrates the ‘convergence’ of the process: the first iteration (b to c in Figure 6) changed the values by 2, the last iteration (c to d) by only $\frac{1}{2}$.

Figure 7 illustrates what actual CT images look like. It is generally agreed that CT scanning has revolutionised some areas of medicine by providing much less invasive, and/or more accurate diagnosis.

Computed tomography can also be applied to ultrasound imaging, magnetic resonance imaging (MRI) and to emissions from radioisotopes in nuclear medicine.

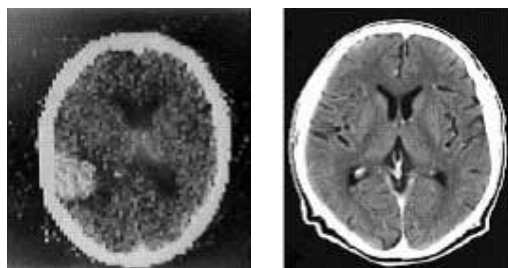


Figure 7 – Two CT images, with different resolutions, each showing a cross section of the brain. The left photograph is of low resolution. The right photograph is of considerably higher resolution.

Reference – Adapted from Giancoli, Douglas C. (1998), *Physics: principles with applications* (5th edition, pp. 784-787). Upper Saddle River, NJ: Prentice Hall.

The CT scanner

Computed tomography is an imaging technique that produces cross-sectional images, representing in each pixel the local X-ray attenuation properties of the body. The first experimental set-up of Hounsfield in 1970 worked with the so-called translation/rotation principle. A thin beam of X-rays was generated through the use of a collimator and a single detector element was used to measure the attenuated intensity. By translating this set-up, different positions were measured. After an entire set of parallel measurements had been acquired, the set-up was rotated to acquire the next parallel projection. This principle was used in what is now called the 1st generation of CT scanners (Figures 3 and 8).

The 2nd generation CT scanners differed only slightly from that initial design in that a small number of measurement values could be obtained simultaneously. In Hounsfield’s first commercially available scanner a total of 180 projections were obtained in steps of 1° with 160 measurement values each. The acquisition of those 28.800 measurement values took five minutes. From that data an image of 80×80 pixels was reconstructed. With such a scanner, a head examination requiring six slices took about half an hour. Therefore,

physicists were aiming at shortening the examination times. This was achieved with the introduction of the 3rd generation CT scanners: a 1D array of detector elements positioned on an arc covers the entire measurement field and acquires a complete 'fan-beam' projection (Figure 8). This not only avoided the slow translation movements, but also improved the efficiency of using the output of the X-ray tube. As figure 9 shows, a modern 3rd generation CT scanner is a machine consisting of a donut-shaped gantry with a big hole. Head, body, arms or legs have to be in the middle of the scanner to make a cross-sectional image. The patient is moved in and out on a motor-controlled table. The slice thickness is usually 0.5 to several mm and the spatial resolution (in the cross section) is roughly 1 mm at 512×512 pixels per slice.

Within the ring of the CT scanner an X-ray tube is placed opposite a detector array with up to 1200 detecting elements, which receive the photons that went through the patient. If one measurement has been done this way, the source and detector rotate over a small angle (roughly 1°) and a new measurement is taken. The scanner repeats this procedure until a rotation of 180° has been reached. Then all thousands of measurements for reconstructing one slice have been done. The table on which the patient lies can then move a little further through the ring for measuring a new slice.

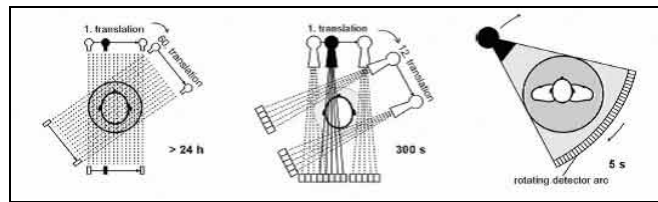


Figure 8 – CT scanner generations. Left: 1st generation – pencil beam (1970). Centre: 2nd generation – partial fan beam (1972). Right: 3rd generation – fan beam (1978).



Figure 9 – A 3rd generation CT scanner.

Spiral CT – In 1987 continuously rotating gantries were introduced to shorten examination times even more: *spiral* or *helical* CT. Up to that time, power supply to the rotating gantry and data transmission out of the gantry was performed via cables. Therefore, the direction of rotation had to be reversed after each scan, substantially slowing down the acquisition of a series of images and making the system rather vulnerable for mechanical cable damages. These drawbacks were overcome with the introduction of slip ring technology for the power supply and optical transfer for data transmission. The patient is moving slowly (1-3 mm/s) and continuously while the scanner rotates constantly at about 1-3 rotations/s. Spiral CT has the important advantage to be fast: modern scanners can collect and reconstruct a high-resolution slice of 512×512 pixels within half a second.

Multi-slice CT – In 1998, several manufacturers introduced *multi-slice* CT (MSCT) systems. This new technique allows the simultaneous acquisition of multiple images by using 2 to 16 detector arrays next to each other.

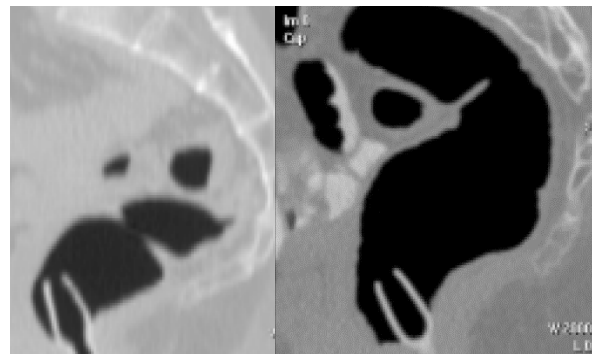


Figure 10 – Two CT images resulting from different techniques. Left: reconstruction from a spiral CT dataset (3 mm slice thickness). Right: reconstruction from a multi-slice dataset (1 mm slice thickness). Note the marked improvement in image quality.

With the simultaneous acquisition of several slices, imaging times are shortened significantly. A four-slice system with a 0.5 s rotation makes it possible to take a CT of the lungs in a few seconds while the patient holds his breath. Other motion artefacts 'disturbing' the image, like the beating of the heart, become less too. Short scan times also

avoid the need to wait for tube cooling between scans.

Image reconstruction

Imagine splitting up a piece of different tissues into many mini-cubes and sending X-ray beams through them at different angles. Detector elements receive signals depending on the different attenuation coefficients μ in each cube along the distance they have to travel. The line of cubes consists of different tissues with different atomic densities (Figure 11).

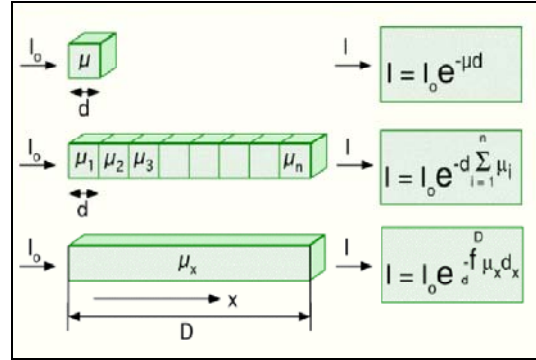


Figure 11 – The attenuated intensity I of the X-ray beams at the detector depends on the intensity I_0 at the source, the attenuation coefficients μ of the different tissues and the path length d .

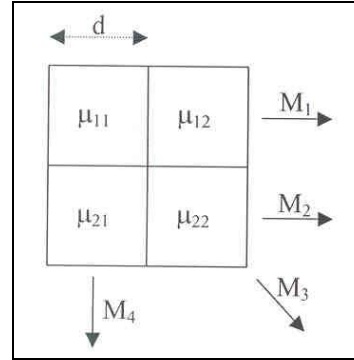


Figure 12

Figure 12 shows a very simplified example to explain how image reconstruction works. Suppose our patient slice contains only four pixels. In such a case we are dealing with four unknown attenuation coefficients (μ_{11} to μ_{22}). Of this four-pixel object four transmission intensities (I_1 to I_4) are measured. Assume that every pixel has a uniformly distributed absorption coefficient. The size of the pixels is given by d . To calculate the absorption coefficients of the four pixels, we have four equations with four unknowns:

$$\begin{aligned} I_1 &= I_0 e^{-(\mu_{11}d + \mu_{12}d)} \\ I_2 &= I_0 e^{-(\mu_{21}d + \mu_{22}d)} \\ I_3 &= I_0 e^{-(\mu_{11}d\sqrt{2} + \mu_{22}d\sqrt{2})} \\ I_4 &= I_0 e^{-(\mu_{11}d + \mu_{21}d)} \end{aligned}$$

This simple problem of four equations with four unknowns can easily be solved, but one can imagine that larger images as used in daily clinical routine (512×512 pixels = 262144 unknown μ values) need highly sophisticated algorithms to solve so many equations. The most widely used algorithm is the filtered back-projection method, using Fourier transform. Scientists are still searching for better algorithms nowadays.

Hounsfield units – To honour Hounsfield for his work the mean X-ray attenuation within one pixel (also known as CT number) is expressed in Hounsfield units (HU). Measured values of attenuation are transformed into CT numbers using the international Hounsfield scale:

$$\text{CT number} = 1000 \cdot (\mu - \mu_{\text{water}}) / \mu_{\text{water}}$$

In this expression μ is the effective linear attenuation coefficient for the X-ray beam. This scale is so defined that air and water respectively have the following CT numbers: -1000 and 0 HU.

Clinical use

Compared to the projection images in conventional X-ray photography, the slice images give a much better contrast between different tissues. This is one of the main advantages of CT. This imaging technique is applied to obtain anatomical images of all parts of the human body. CT is often used to detect cancer or follow the growing of tumours over time. As tumours are often well supplied with blood vessels (as they need much supply due to their fast growth), the use of contrast agents can make them visible (Figure 13). Timing is therefore important in CT imaging too. A dynamical serial measurement can be studied to detect the changes in blood vessel filling after the contrast agent is applied.

Also measuring the size of the bladder or detecting water in lungs is usually done with CT scanners. As CT is based on X-ray attenuation, a contrast agent with iodine (in blood

vessels) or barium (in the intestines) is often used, like in conventional X-ray imaging and Digital Subtraction Angiography (DSA) (see: Scientific/Technological fact sheet – X-ray Photography). Because CT is more sensitive to small intensity differences, small diffused concentrations outside blood vessels or cavities can also be detected.

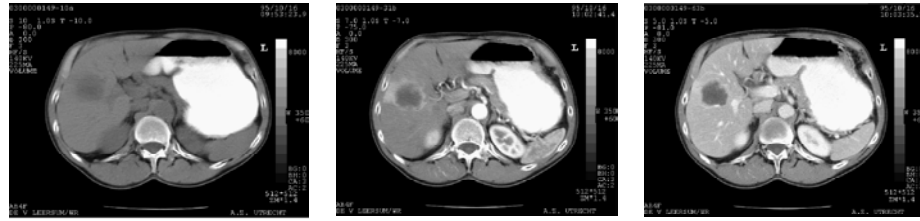


Figure 13 – Timing study: the liver tumour is visible due to the abnormal perfusion in the tumour.

3D images – From a whole stack of CT scans one can make a real 3D reconstruction, for example of the inner ear (Figure 14). One can even make a ‘virtual endoscopy’. In such a case the intestines in the patient are emptied and slightly filled with air. Then a large series of high resolution CT scans is made. The radiologist can make a virtual flight (as on a very small airplane) through the intestine canal to search for polyps in the full nine meters of intestines. Another way of visualising the intestine canal is the ‘unfolded cube view’, in which all six viewing directions are projected in one cubical view (Figure 15).

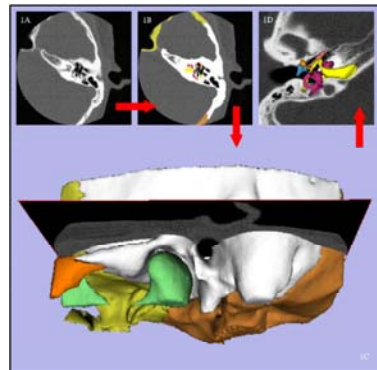


Figure 14 – Consecutive CT slices of the inner ear can be used to construct a 3D image of different details of the auditory organs, like the stapes and anvil.

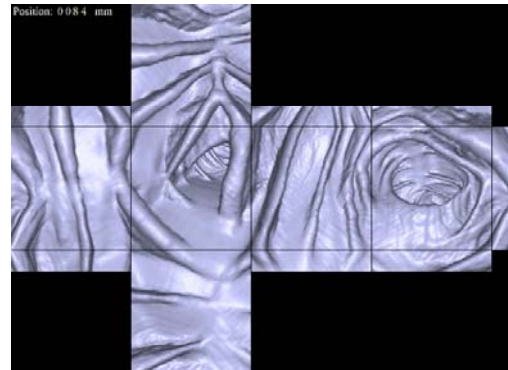


Figure 15 – Cube view of the intestines.

Real 3D CT images are also often used in radiotherapy clinics, where patients with cancer are treated by high intensity radiation at the tumour location. To make a treatment plan for the patient one can calculate by image processing on the CT slices which radiation dose will be delivered where. Clinical physicists play an important role in the dose calculations and radiotherapy treatment planning, avoiding radiation of vital organs as much as possible.

Radiation safety

Radiation doses in CT are relatively high. For example, the effective dose of a head scan is 2 mSv, of the thorax 10 mSv and of the abdomen 15 mSv. This is a factor 10 to 100 higher than radiographic images of the same region, but the diagnostic content of the CT images is typically much higher. Some scanners use a lower tube current and a higher voltage to reduce the dose. However, there is still some risk to a developing foetus. CT scans are therefore not recommended during pregnancy.

Physics principles involved

- X-rays: transmission and absorption, attenuation coefficient

Additional links

<http://science.howstuffworks.com/cat-scan.htm>