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Excerpt

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## Section 1

## Fundamentals of cross-sectional imaging

### Chapter

# 1

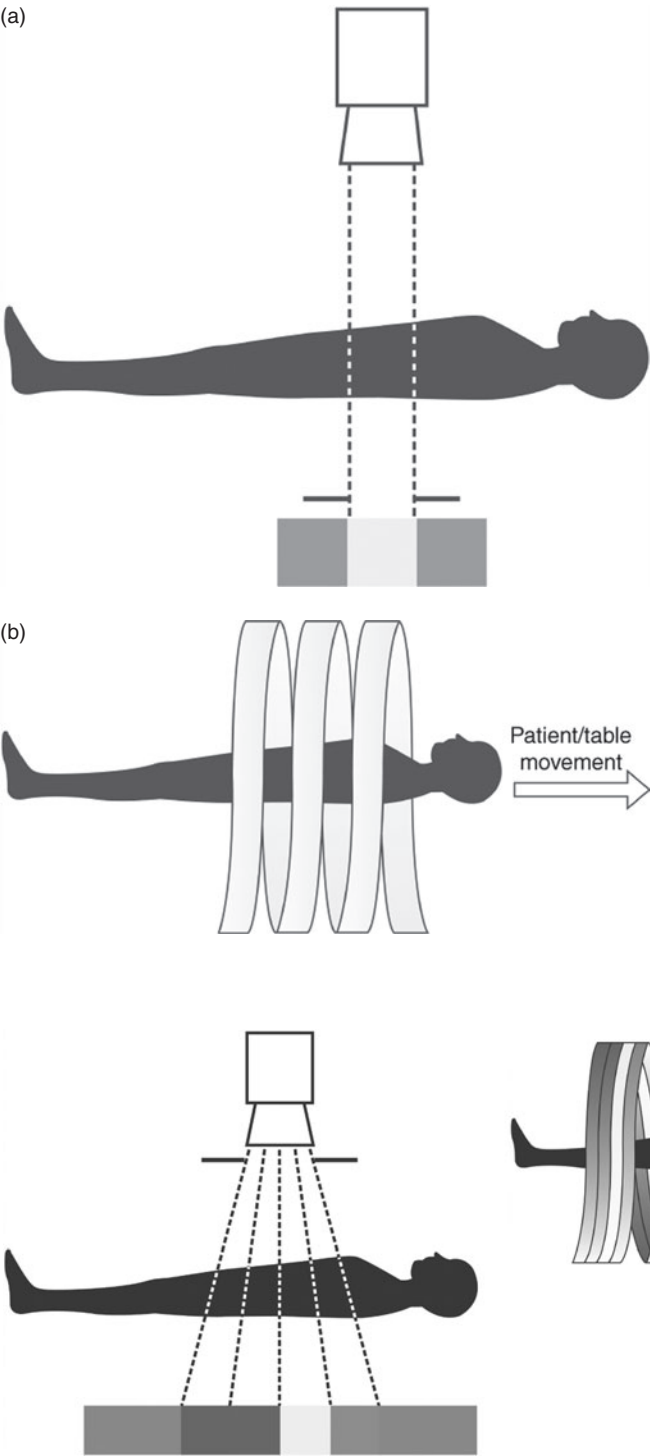
## Computed tomography

### History

- In the early 1970s, Sir Godfrey Hounsfield's research team produced the first clinically useful computed tomography (CT) scans.
- The original scanners took approximately 6 minutes to perform a rotation (one slice) and 20 minutes to reconstruct it (Fig. 1.1a). Despite many technological advances since then, the principles remain the same.
- On early scanners, the tube rotated around a stationary patient, with the table moving to enable further acquisitions. The machine rotated clockwise and counter-clockwise as power was supplied via a cable.
- Modern-day helical or spiral scanners obtain power via slip ring technology, thus allowing continuous tube rotation as the patient moves through the scanner automatically (Fig. 1.1b). This allows a large amount of data to be acquired in a single rotation, with the benefits of faster scanning, faster patient throughput and fewer patient movement artefacts.
- New multi-slice scanners use existing helical scanning technology, but have multiple rows of detectors to acquire multiple slices per tube rotation. The faster imaging with multi-slice scanners allows a larger volume of coverage and multiphase scanning during intravenous contrast administration (Fig. 1.2). This, coupled with improved spatial resolution, allows organ-specific as well as peripheral vascular assessment, leading to the advent of CT angiography and virtual endoscopy.
- Advanced computer processing power allows reconstructive techniques, such as three-dimensional and multiplanar reformatting, providing us with additional tools with which to improve diagnostic accuracy and aid clinical management.

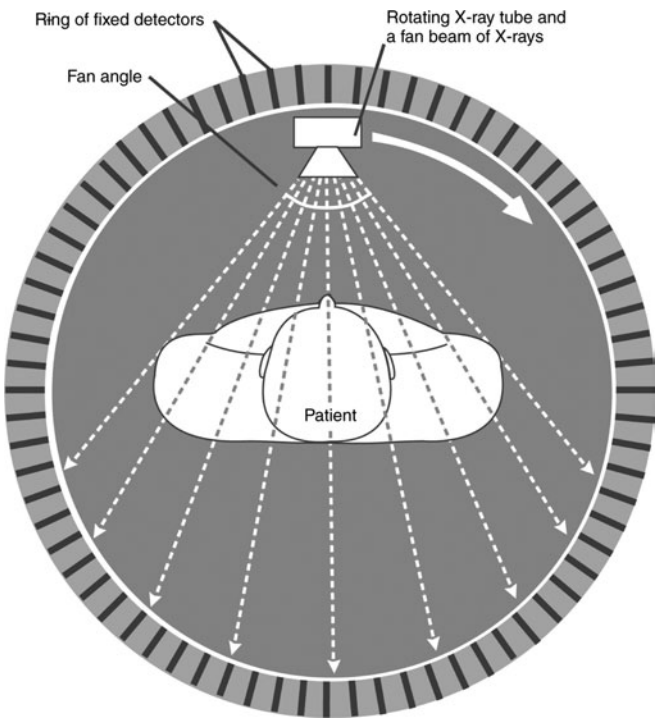
### Technical details

- The x-ray tube produces a narrow fan-shaped beam of collimated x-rays, which pass through the patient to reach a bank of detectors opposite the source (Fig. 1.3).
- X-rays are attenuated differentially by the patient, depending on the tissues through which they pass. Low-density tissues such as fat and aerated lung absorb fewer x-rays, allowing more to reach the detector. The opposite is true for dense tissues such as bone.



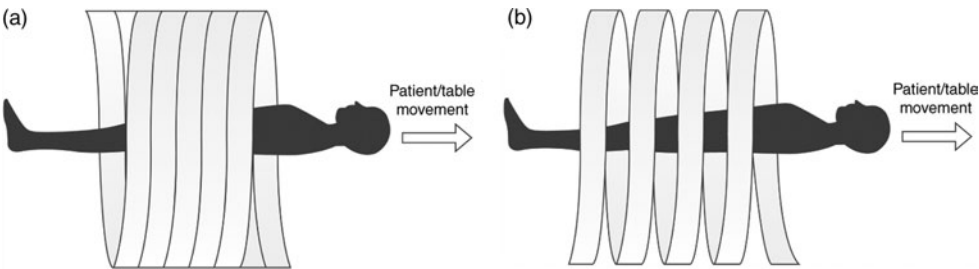
**Fig. 1.1.** Diagrams showing (a) a single-slice scanning system and (b) a single-slice helical CT scanning system, where the x-ray tube continues to rotate as the patient moves through at a constant rate.

**Fig. 1.2.** Multi-slice helical CT scanner with four detectors.



**Fig. 1.3.** Helical-ring system demonstrating the path of the x-rays from source to detector.

- The amount of transmitted x-ray radiation received by the detector provides information about the density of the tissue through which it has passed.
- A CT slice is divided up into a matrix of squares, e.g.  $256 \times 256$ ,  $512 \times 512$  and  $1,024 \times 1,024$ . The slice thickness determines the volume of these squares: these are called voxels. Using mathematical calculations, the degree to which a tissue absorbs radiation within each voxel, the *linear attenuation coefficient*  $\mu$ , is calculated and assigned a value related to the average attenuation of the tissues within it – the *CT number* or *Hounsfield unit (HU)*.
- Each value of  $\mu$  is assigned a greyscale value on the display monitor and is presented as a square picture element (pixel) on the image.
- Spiral scanners acquire a volume of information from which an axial slice is reconstructed, as above, using computer technology. Slices are created from the data during the reconstruction phase.
- *Pitch* is defined as the distance moved by the table (in millimetres) during one complete rotation of the x-ray tube, divided by the slice thickness in millimetres. In general, increasing pitch (increase in table speed with a fixed slice thickness) reduces the radiation dose to the patient (Fig. 1.4). This, in turn, reduces the amount of radiation reaching the detector for interpretation, with the net result of reduced image resolution. A compromise usually exists between limiting the patient's radiation dose and diagnostic image quality.



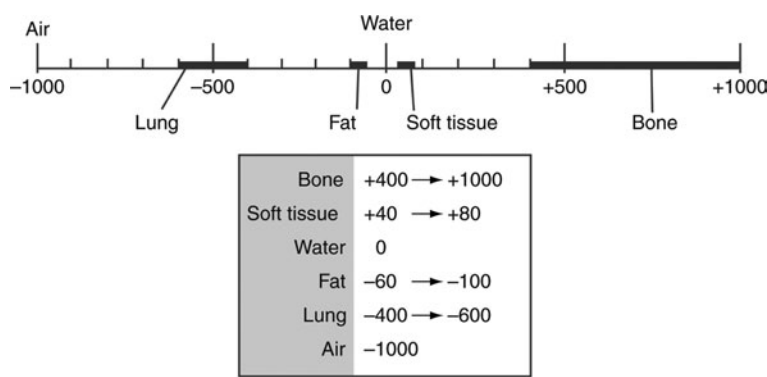
**Fig. 1.4.** Low pitch vs. high pitch: (a) low-pitch scanning with the table moving less for each tube revolution, resulting in a sharper image; (b) higher-pitch scanning, resulting in stretching of the helix as the table moves more for each revolution, leading to a loss of image quality.

## Windowing and greyscale

- Modern CT scanners are able to differentiate in excess of 2,000 CT numbers; however, the human eye can only differentiate around 30 shades of grey.
- To maximize the perception of medically important features, images can be digitally processed to meet a variety of clinical requirements.
- The greyscale values assigned to process CT numbers on a display monitor can be adjusted to suit the requirements of particular applications.
- Contrast can be enhanced by assigning a narrow interval of CT numbers to the entire greyscale on the display monitor: this is called the *window technique*. The range of CT numbers displayed on the whole greyscale is called the *window width* and the average value is called the *window level*.
- Changes in window width alter contrast, and changes in window level select the structure of interest to be displayed on the greyscale, i.e. from black to white.
- Narrowing the window compresses the greyscale to enable better differentiation of tissues within the chosen window. For example, in the assessment of CT of the head, a narrow window of approximately 80 HU is used, with the centre at 30 HU. CT numbers above 70 (i.e.  $30 + 40$  HU) will appear white, and those below  $-10$  (i.e.  $30 - 40$  HU) will appear black. This allows subtle differences in tissue densities to be identified.
- Conversely, if the window were widened to 1,500 HU, then each detectable shade of grey would cover 50 HU and soft-tissue differentiation would be lost; however, bone–soft tissue interfaces would be apparent.
- In practical terms, the window width and level are preset on the workstation and can be adjusted by choosing the appropriate setting, i.e. a window setting for brain, lung, bone, etc.

## Tissue characteristics and contrast medium

- Unlike conventional radiography, CT has a relatively good contrast resolution and can therefore differentiate between tissues which vary only slightly in density (Fig. 1.5). This is extremely valuable when assessing the brain, as grey and white matter differ only slightly in density.



**Fig. 1.5.** Graphical representation showing Hounsfield scale and CT numbers for different types of tissue, water and air.

- In body CT, radio-opaque contrast agents are used to improve the quality of the study, depending on the clinical setting. This may be in the form of dilute barium (intraluminal contrast) or iodinated agents, which can be used both intravenously and intraluminally.
- Intravenous contrast agents opacify blood vessels, increasing the density of vascular abdominal organs, thereby improving the contrast between lesions and normal structures. Using multi-slice scanners, the passage of contrast through the arterial, parenchymal and venous systems can be assessed with multiphase imaging. On the other hand, intraluminal contrast agents distend the gastrointestinal tract to aid detection of luminal lesions.
- Intravenous contrast medium can also be used in both head and neck and body imaging to reveal abnormalities which are either vascular or disrupt the normal parenchyma. These can often be difficult to observe on an unenhanced study.
- The densest structure in the head is bone, which appears white on the CT scan. This is followed by acute haematoma, which is more dense than flowing blood due to clot retraction and loss of water. The hyperdensity of blood is due to the relative density of the haemoglobin molecule. Over time, blood appears isodense and then hypodense, compared with brain parenchyma due to clot resorption. The brain can be differentiated into grey and white matter due to the difference in fatty myelin content between the two. Typically, white matter (higher fatty myelin content: HU  $\approx$  30) is darker than the adjacent grey matter (HU  $\approx$  40).
- In body CT, bleeding sites are identified by focal areas of contrast medium extravasation, which appear as high-density linear streaks or blushes due to the relative high density of contrast medium. Alternatively, bleeding may also be indicated by increased density of intra-abdominal fluid (>50 HU) due to the mixing of blood or dense contrast medium with fluid of a density similar to that of water, i.e. <20 HU.
- Fat and air have low attenuation values, with a negative Hounsfield unit, and can readily be identified. Subtle pockets of air are best identified on lung windows, as will be demonstrated in later chapters. These are often difficult to appreciate in the abdomen against adjacent fat, which also appears dark.

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- Cerebrospinal fluid has a similar attenuation value to water, near to zero Hounsfield units, and appears black.
- Pathological processes may become apparent due to the oedema within, or adjacent to, an abnormality. Oedema is less dense than normal brain parenchyma. In the body, this is often detected by increased density within the intra-abdominal fat surrounding an affected organ.

## Image artefacts

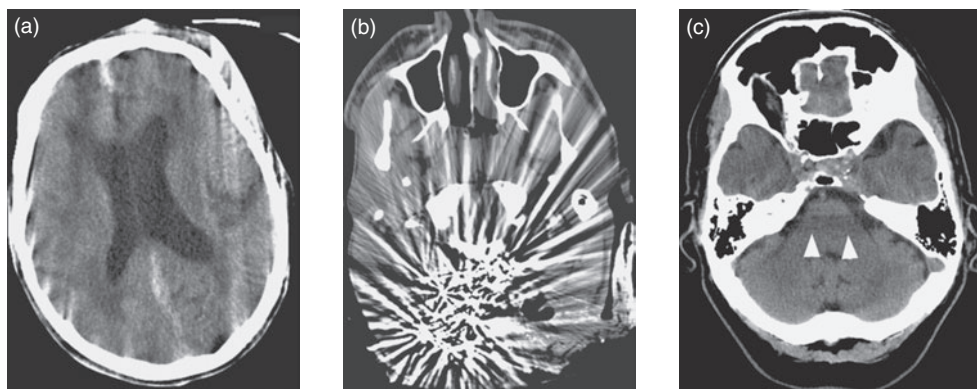
- An artefact is a visual impression in the image of a feature that does not actually exist in the tissue being imaged. Artefacts are important to recognize so that they are not confused with pathology. Artefacts may occur due to scanner malfunctions, patient movement and the presence of extrinsic objects within the slice being scanned, e.g. a metallic foreign body.
- Fortunately, many artefacts have now been reduced or eliminated through advances in CT technology and speed.

## Movement artefacts

- Occur with voluntary and involuntary patient movement.
- Result in streak patterns (Fig. 1.6a).
- Can be reduced by patient co-operation, quicker scan times and software compensation.

## Partial volume artefacts

- The CT number reflects the average attenuation within the voxel and, thus, if a highly attenuating structure is present within the voxel, it will raise the average attenuation of the whole voxel.



**Fig. 1.6.** (a) Axial image of the brain demonstrating blurred outlines and streak pattern across the brain parenchyma secondary to movement artefacts. (b) Axial image of the brain demonstrating a star-shaped, high-density streak across the brain parenchyma secondary to metallic artefacts. (c) Axial image of the brain demonstrating bands of low attenuation across the pons secondary to beam-hardening artefacts from the skull base.

- Contamination can occur especially with thicker slices and near bony prominences. It is therefore important to always review the images above and below to assess for structures likely to cause partial volume artefacts.
- Can be reduced by using thinner slices (e.g. posterior fossa) and software compensation.

Metallic artefacts

- The attenuation coefficient of metal is much greater than any structure within the body. As a result, radiation is completely attenuated by the object and information about adjacent structures is lost.
- Produces characteristic star-shaped streak artefacts (Fig. 1.6b).
- Can be reduced by widening the window; at a cost to parenchymal detail.
- Software manipulation may help.

Beam-hardening artefacts

- Results from an increase in the average energy of the x-ray beam as it passes through a tissue.
- Think of CT as using a spectrum of radiation energy; low-energy radiation is filtered out by high-density structures such as bone, leaving higher-energy radiation that is less absorbed by soft tissues, causing a low-attenuation streak artefact.
- Characterized by linear bands of low attenuation connecting two areas of high density, such as bone, e.g. the posterior fossa in the brain (Fig. 1.6c).
- Can be reduced by using a filter to adjust the spectrum of radiation and by post-processing software.

Quantum mottle

- Image reconstruction in CT requires a sufficient number of radiation photons to strike the detectors.
- The following situations reduce the number of radiation photons, resulting in a photon-poor imaging technique, which produces a grainy CT image:
  - Reducing slice thickness to reduce partial volume artefacts.
  - Alteration of CT x-ray technique to reduce the patient’s exposure to radiation.
  - Patient’s body habitus limiting penetration of the photons.
- Quantum mottle can be reduced by increasing the slice thickness or increasing the energy of the photon, which will increase the number of artefacts and the patient’s radiation dose, respectively. A compromise between image quality, presence of artefacts and radiation dose is therefore necessary.



Section 1

Fundamentals of cross-sectional imaging

Chapter

2

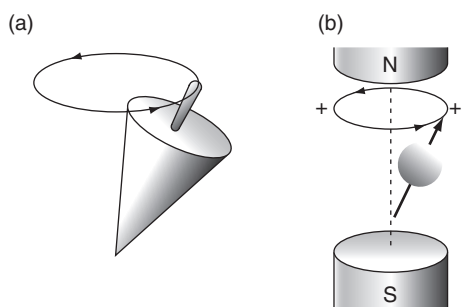
Magnetic resonance imaging

- Magnetic resonance imaging (MRI) is based on the principles of nuclear magnetic resonance (NMR), a spectroscopic technique used by scientists to study the chemical and physical properties of molecules at a microscopic level.
- In 1971, Raymond Damadian demonstrated that normal tissue and tumour tissue have different nuclear magnetic relaxation times. This raised the possibility of using magnetic resonance in clinical practice for disease detection.
- In 1977, the first MRI scan was performed on a human subject. It took almost 5 hours to produce one image.
- MRI is a rapidly developing imaging technique, which generates tomographic images safely and effectively.
- One of the major technological advantages of MRI is its ability to discriminate between different tissues according to their physical and biochemical properties. It therefore provides excellent soft-tissue contrast and is capable of demonstrating both anatomical features and flow-related phenomena. Unlike computed tomography (CT), MRI does not use ionizing radiation. Instead, it uses a powerful magnetic field to align the magnetization of some of the atoms in the body.
- The strength of a magnet in a MRI scanner ranges from 0.5 to 3 tesla (T), the unit of magnetic field strength. To put this into perspective, the Earth's magnetic field strength ranges from 0.00003 to 0.00007 T.
- In order to appreciate the complexities involved in image acquisition and interpretation of MRI, a basic understanding of MRI physics is required.

MRI physics

- MRI involves exploring the magnetic properties of human tissues. Hydrogen atoms, or protons, are essential components in human tissues; they provide the signal in MRI, which produces useful information.
- Hydrogen atoms are abundant in the human body, primarily in the form of water and fat. These protons have a single positive charge and spin on their own axis like the Earth. Additionally, protons also precess at a given frequency like a spinning top (Fig. 2.1) and this moving electric charge (i.e. an electric current) produces a magnetic field.





**Fig. 2.1.** (a) The motion of a spinning top describes how protons spin on their own axis and precess to produce a magnetic field. (b) Each proton possesses a positive charge. A moving charge (i.e. an electric current) induces a magnetic field.

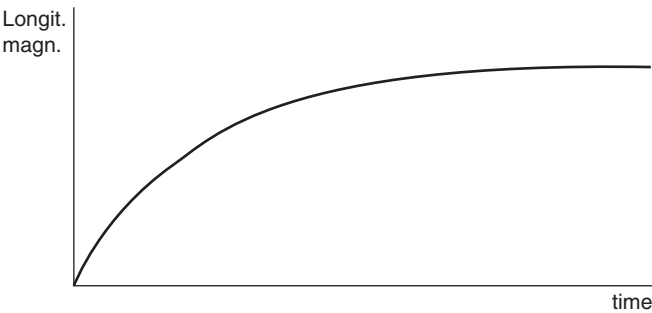
- The human body can therefore be thought of as having millions of charged spinning protons, or bar magnets. Normally, these bar magnets are randomly aligned so that their charges cancel each other out, and thus there is no net magnetic moment.
- In an MRI scanner, an external magnetic field runs down the centre of the tube. When a patient is placed on the MRI table, the protons within the patient align themselves along the external magnetic field. The protons can either align with their north and south poles parallel to the external magnetic field or anti-parallel to it.
- These states have different energy levels, with the anti-parallel state requiring more energy. The desirable state is the one that requires the least amount of energy; that is, parallel to the external magnetic field.
- The difference between the two states is very small when placed in a magnet; the actual difference in the number of protons depends on the strength of the magnet. For every 10 million protons aligned anti-parallel to the magnetic field, approximately 10,000,007 will be aligned parallel to it. The majority of the protons therefore cancel each other out, leaving only a few unmatched protons per million, which result in a net longitudinal magnetic vector.
- The protons in the patient precess at a frequency that is proportional to the strength of the magnetic field. The equation that allows us to calculate this is known as the Larmor equation:

$$\omega_0 = \gamma B_0$$

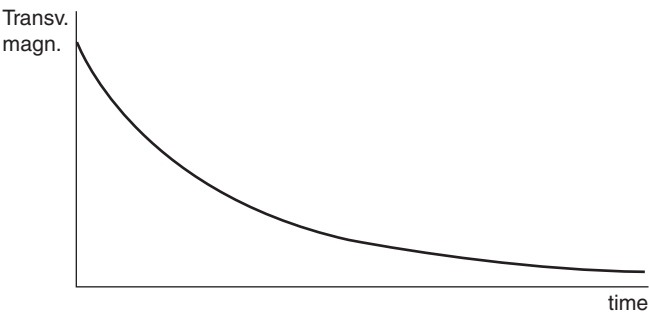
where  $\omega_0$  = angular precession frequency in megahertz (MHz),  $\gamma$  = gyromagnetic ratio, and  $B_0$  = external magnetic field in tesla (T).

- This means that as the external magnetic field strength increases, the precession frequency also increases. This relationship is determined by the gyromagnetic ratio ( $\gamma$ ), which is different for different materials. For protons, this value is 42.5 MHz/T.
- Once the patient is placed into the magnet, a radio-frequency (RF) pulse is then applied to the patient through a coil. The RF pulse has the same frequency as that of the protons in the patient, known as the resonance frequency. There are two important effects of the RF pulse:
  - Some of the protons are elevated to a higher energy level (i.e. align anti-parallel to the magnetic field).
  - The protons are forced to precess in synchrony (in phase).

- The first effect causes the longitudinal magnetization to decrease, as there are now more protons aligned anti-parallel to the external magnetic field. The second effect creates a new magnetization called the transverse magnetization.
- Once the radio-frequency pulse is switched off, two major processes occur:
  - First, the protons that were raised to a higher energy level by the RF pulse now resume their original lower energy level, and in so doing release energy to the surrounding lattice. This is known as the longitudinal relaxation, spin-lattice relaxation or T1 relaxation. The protons point parallel to the magnetic field once more and thus the longitudinal magnetization increases, eventually going back to its original value. If this were to be graphically represented with time versus longitudinal magnetization, a characteristic curve is produced, known as the T1 curve (Fig. 2.2).
  - The second process to occur is that the protons no longer precess in synchrony, or in phase. The original external magnetic field is not completely homogeneous, resulting in protons with slightly different precession frequencies. Further to this, protons are influenced by the tiny magnetic fields of neighbouring protons, thus causing different precessing frequencies. By losing phase coherence, the protons start to point in different directions and therefore transverse magnetization decreases. This is known as transverse relaxation or T2 relaxation. By plotting time against transverse magnetization, a further characteristic curve is produced, known as the T2 curve (Fig. 2.3).
- Recovery of longitudinal relaxation and loss of transverse magnetization results in a change in direction of the magnetic moment which, in turn, induces an electric current



**Fig. 2.2.** T1 curve: graph plot of longitudinal magnetization vs. time after the RF pulse is switched off.



**Fig. 2.3.** T2 curve: graph plot of transverse magnetization vs. time after the RF pulse is switched off.