521. X-ray computed tomography

Theoretical introduction

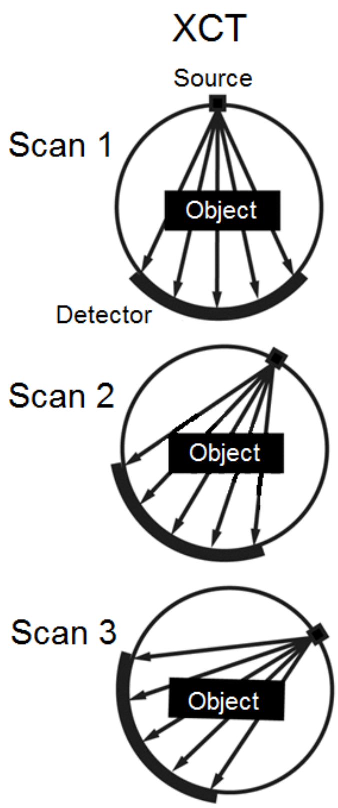
X-ray computed tomography (XCT) is one of the most important diagnostic methods in today medical practice. Furthermore, it is widely used in industry and scientific research to gain information on the internal structure of different objects without the need to destroy these objects.
The prerequisite which stimulated the development of tomography was the disadvantage of the standard X-ray investigation. The idea was to obtain not a single image, but rather a series of images seen at different orientations, and to reconstruct then the distribution of the density of the matter in a series of sliced by mathematical processing. Advantages of the computed tomography as compared to the traditional radiography are:

- absence of shadows and superpositions in the image;
- increased accuracy of measurements of geometrical proportions;
- sensitivity which is an order of magnitude higher than in radiography.

An image obtained by XCT is the two-dimensional projection of a 3D object, through which the radiation has passed. In this projection slices localized on different depths superpose on one another, so this image is also called a superposition image.

The principle of obtaining scans is the same for different CT methods; the only difference is in the radiation type used. An ultrasonic wave may be used instead of X-rays. In the X-ray CT, absorption of X-ray irradiation is registered, while in ultrasonic computed tomography (UCT), the decay time of ultrasound is measured. The image processing algorithms are the same in both methods.

In an X-ray CT investigation, X-rays from a source are passing through an object, and then the irradiation absorption level is registered in an array of points at the same time. The “source – sensor” system is then rotated around the object (patient) by a certain angle, and a new scan is acquired (see the figure). This action is repeated several times until the whole required angle range is swept.



Generation of X-rays

X-ray radiation is produced by an X-ray tube, which is schematically shown below (Fig. 1).

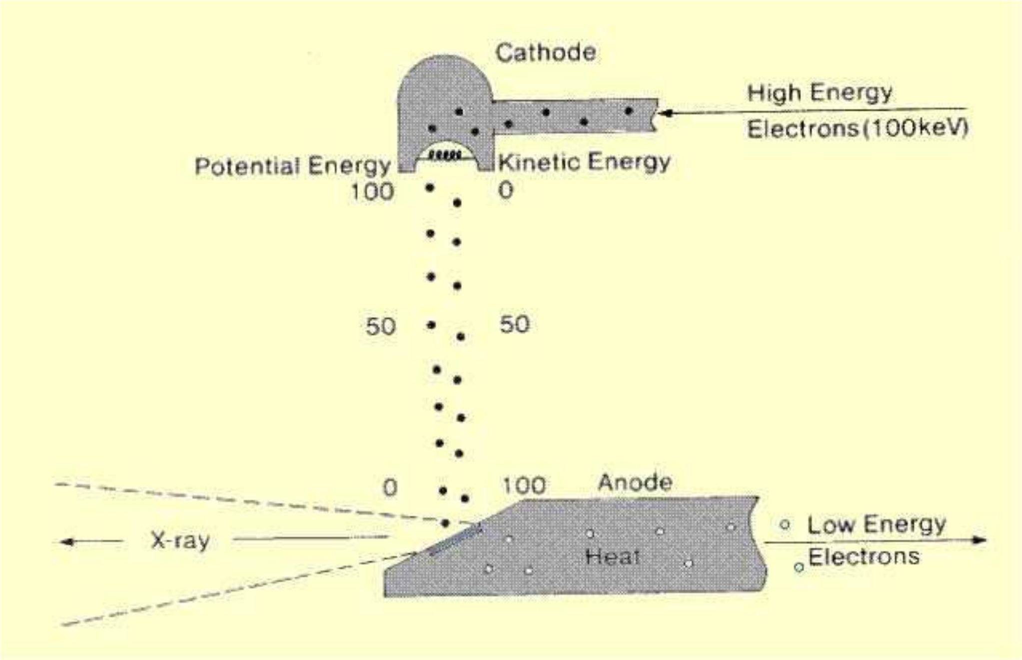


Fig. 1. Scheme and energy exchange within an X-ray tube.

The basic function of the cathode is to expel the electrons from the electrical circuit and focus them into a well-defined beam aimed at the anode. The typical cathode consists of a small coil of wire (a filament) recessed within a cup-shaped region. The filament of the cathode is heated in the same way as a light bulb filament by passing a current through it (it is not the same as the current flowing through the X-ray tube that produces the X-radiation!). During tube operation, the cathode is heated to a glowing temperature, and the heat energy expels some of the electrons from the cathode (so-called *thermionic emission*).

The anode is the component in which the X-radiation is produced. It is a relatively large piece of metal that connects to the positive side of the electrical circuit. The anode has two primary functions: (1) to convert electronic energy into X-radiation and (2) to dissipate the heat created in the process.

The ideal situation would be if most of the electrons created X-ray photons rather than heat. The fraction of the total electronic energy that is converted into X-radiation (efficiency) depends on two factors: the atomic number (Z) of the anode material and the energy of the electrons. Most X-ray tubes use tungsten, which has an atomic number of 74, as the anode material. In addition to a high atomic number, tungsten has several other characteristics that make it suited for this purpose. Tungsten is almost unique in its ability to maintain its strength at high temperatures, and it has a high melting point and a relatively low rate of evaporation. Sometimes anode body and its surface are made of different materials (molybdenum and graphite can be used to make the body, and the surface is made from Mo, W–Re alloy, Rh, Cu, etc.). Mo has an intermediate atomic number (Z = 42), which produces characteristic X-ray photons with energies well suited to mammography.

Not all area of the anode is involved in X-ray production. The radiation is produced in a very small area on the surface of the anode known as the *focal spot* (typically from 0.1 to 2 mm in size). The dimensions of the focal spot are determined by the dimensions of the electron beam arriving from the cathode. X-ray tubes are designed to have specific focal spot sizes; small focal spots produce less blurring and better visibility of detail, and large focal spots have a greater heat-dissipating capacity (thus increasing the lifetime of a tube).

As a result of the electron (or any other charged particle) decelerating in the electrostatic field produced by atoms (nuclei and surrounding electrons) in a material of the anode, so-called *braking radiation*, or *bremsstrahlung* is generated.

The fraction of the kinetic energy of an electron which will be transformed to an X-ray quantum and which will cause heating of the anode material are random, and therefore the beam of a large amount of electrons produce a continuous spectrum of the X-ray radiation. The highest-energy radiation quanta (with the shortest wavelength) are produced when the whole kinetic energy of the electron is transformed to radiation. We can write down the law of conservation of energy as

$$eU = h\nu_{\max} = hc/\lambda_{\min} ,$$

and hence $\lambda_{\min} = hc/(eU)$. If the wavelength is measured in Å (10^{-10} m), and voltage U in kV, then $\lambda_{\min} = 12.3/U$ (see Fig. 2a).

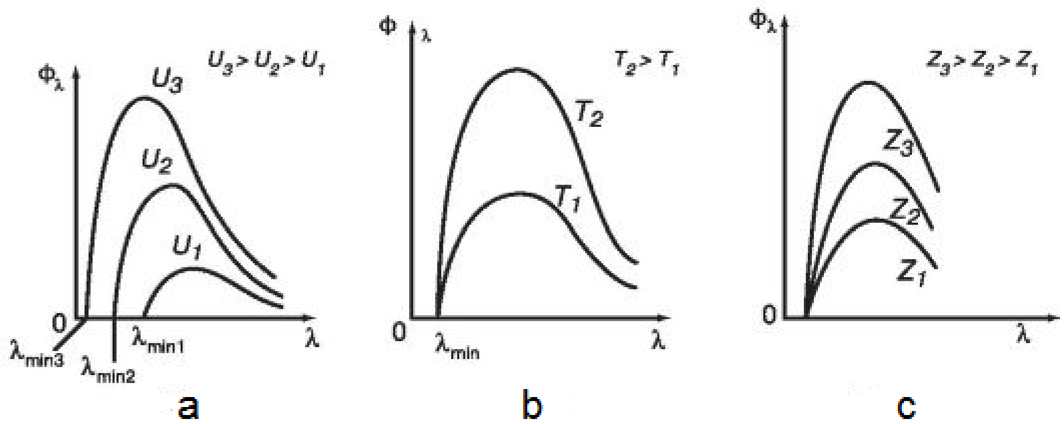


Fig. 2. Flow of X-ray radiation at (a) different accelerating voltages at the tube, (b) different cathode temperatures (i.e., emission currents), and (c) different anode materials (Z is the atomic number).

Short-wavelength radiation shows a higher permeability and is called *hard radiation*, in contrast to *soft radiation* (λ ≥ (0.1…0.2) nm, E ≤ (5…10) keV). Soft X-rays are easily absorbed in air; the attenuation length of 600 eV (~2 nm) X-rays in water is less than 1 micrometer (at this distance it decreases *e* times).

Thermal emission is more active, when the cathode is heated to larger temperatures. The spectral composition of the radiation is not changed, but the total intensity increases (Fig. 2b).

The intensity of radiation depends also on the material of the cathode, and it is obvious from Fig. 3c why heavy metals are used in practice for this purpose. The X-ray radiation flow depends on the anode material's atomic number *Z*, voltage at the tube *U*, and the current *I* passing through the tube (in the electron beam) as

$$\Phi = kIU^2Z\ ,$$

where *k* is a coefficient ($k = 10^{-9}\ V^{-1}$).

In addition to the continuous spectrum of bremsstrahlung, X-ray tubes can produce additional sharp peaks (line spectrum). This is so-called *characteristic radiation*, which is observed if the incoming electron has a kinetic energy greater than the binding energy of an electron within the atom. When this condition exists, and the collision between an incoming electron from the beam and an atom of the anode occurs, the electron is dislodged from the atom. When the orbital electron is removed, it leaves a vacancy that is filled by an electron from a higher energy level. As the filling electron moves down to fill the vacancy, it gives up energy emitted in the form of an X-ray photon. This is known as characteristic radiation because the energy of the photon is characteristic of the chemical element that serves as the anode material, independent of the molecular composition in which this atom is included. In the example of W, the electron can dislodge a tungsten K-shell electron, which has a binding energy of 69.5 keV. The vacancy is filled by an electron from the L shell, which has a binding energy of 10.2 keV. The characteristic X-ray quantum, therefore, has energy equal to the energy difference between these two levels, or 59.3 keV.

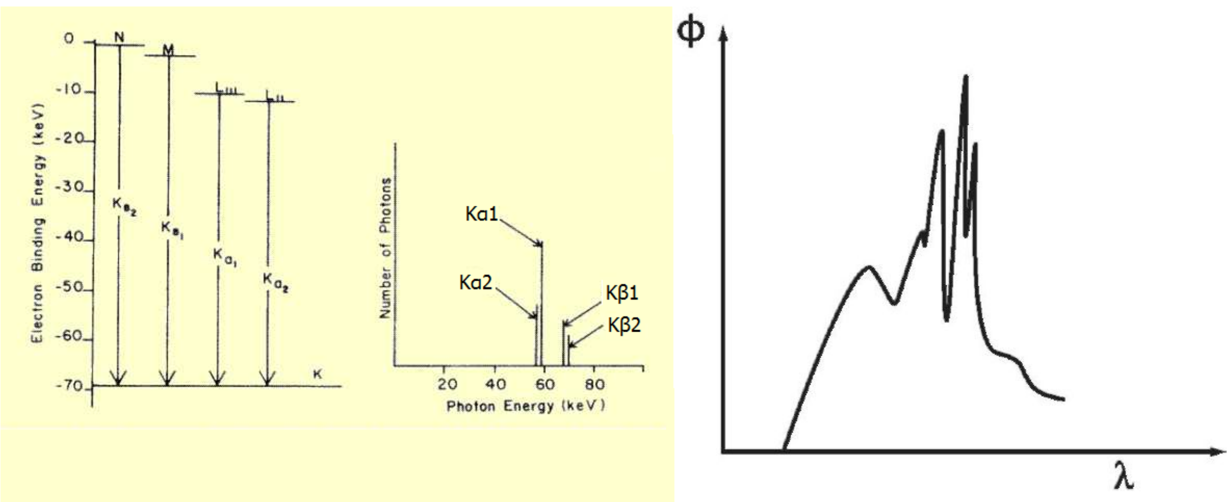


Fig. 3. Characteristic radiation: scheme of energy levels for W and a typical mixed spectrum of an X-ray tube at high voltages.

Frequency of characteristic radiation is related to the atomic number as $\sqrt{\nu} = A(Z - B)$ (Moseley's law), where A and B are constants that depend on the type of line (K, L, etc. in X-ray notation).

X-ray tomographic system

An X-ray tube together with a collimating system creates a narrow fan-shaped beam having a divergence angle of 30°–50°. The degree of attenuation of the X-ray beam passing through an object is recorded by detectors which transform the radiation into electrical signals. These analogue signals are then amplified by electronic units and converted to the digital form. Some materials are very effective in transforming X-ray radiation; most often two types of detectors are used: luminescent and gas detectors.

In the former type, a luminescent crystal is connected to a photomultiplier tube to transform light flashes into electronic signals. The amount of light generated is proportional to the energy of radiation which has been absorbed. Detectors of this kind were used in the 1 and 2 generation scanners. Their disadvantages are impossibility of close disposition of the detectors and persistence (afterglow effect).

A gas detector is an ionization chamber filled with xenon or krypton. Ionized gas allows electrons to come to tungsten plates, and thus electronic signals proportional to the incoming radiation intensity are generated. The plates are separated by gaps of 1.5 mm. Gas detectors were developed for the 3 generation scanners and provide high resolution and sensitivity. Their efficiency is close to 100% since they can be located very close to each other.

Solving the mathematical problem of reconstruction of a tomography image belongs to a class of ill-posed problems (1^{st} order operator equation). Exact and unambiguous solving of problems of this kind is, in general case, impossible.

Detector obtains data reflecting the interaction of X-ray radiation with the matter of which the studied object is made. Energy of photons decrease while they pass through the object due to photoelectric effect (absorption) and Compton effect (scattering). Attenuation coefficient for a narrow X-ray beam is defined by the linear attenuation coefficient μ, which is the feature of a certain material:

$$I = I_0 e^{-\mu d},$$

where d is the object's thickness, I_0 is the beam intensity emitted by the source, and I is the intensity measured by the detector.

The narrow beam is scattered by all voxels (volume elements) as shown in Fig. 4. The total attenuation coefficient for a beam passed through a certain series of voxels is $\mu_\Sigma = \mu_1 + \mu_2 + \cdots + \mu_{N-1} + \mu_N$, and it is this value which can be defined from the measurement, because $I = I_0 \exp(-\mu_\Sigma d)$.

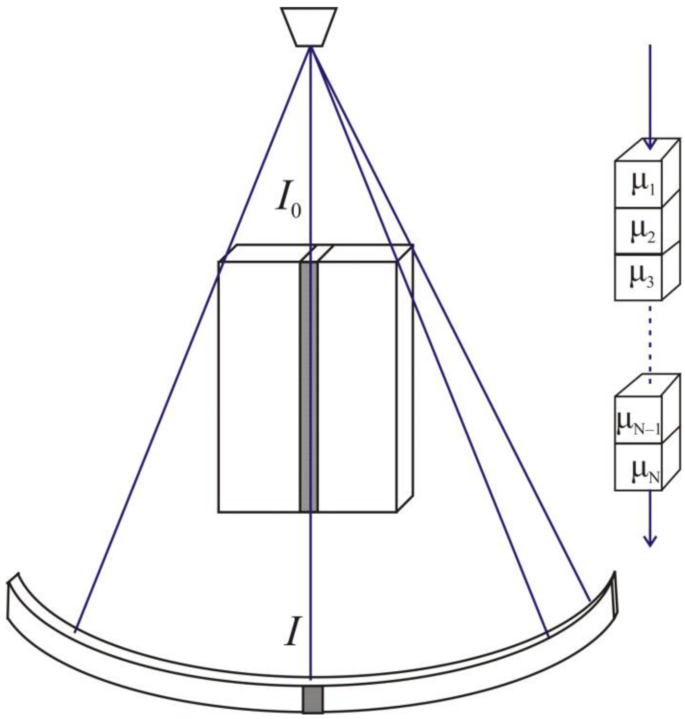


Fig. 4. Passing of an X-ray beam through a series of voxels.

It is possible to determine the absorption (attenuation) coefficients for each voxel by the method of inverse projections, which is based on obtaining information on absorption of X-ray radiation in many directions. Consider, as an example, the case of four voxels (Fig. 5, left).

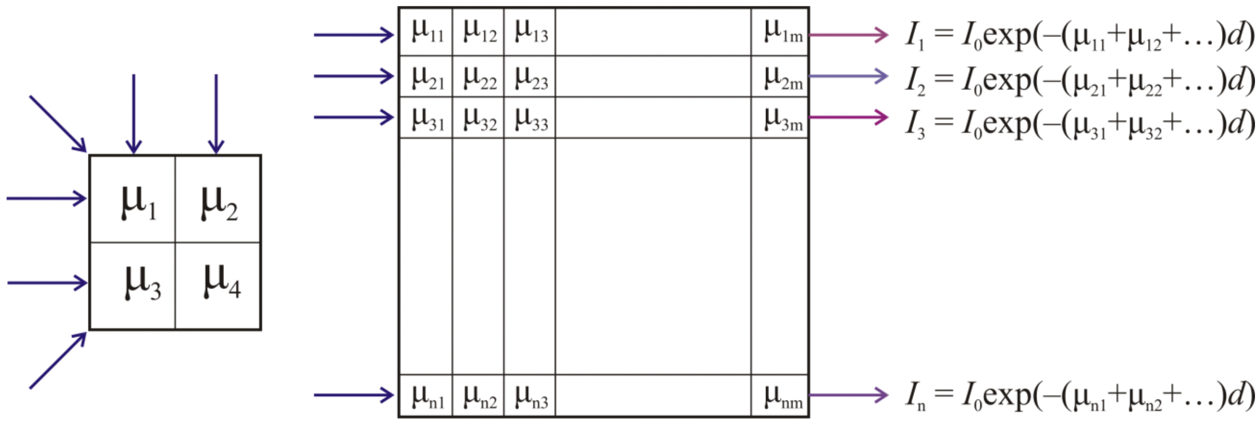


Fig. 5. Passing of an X-ray beam through different series of voxels in a slice.

This slice is irradiated in several orientations, and as a result we get a number of different total coefficients giving a system of equations:

$\mu_1 + \mu_2 = \mu_{12}$ (horizontal),

$\mu_2 + \mu_3 = \mu_{23}$ (diagonal),

$\mu_1 + \mu_3 = \mu_{13}$ (vertical),

$\mu_1 + \mu_4 = \mu_{14}$ (the second diagonal).

By solving this system, we obtain attenuation coefficients for the considered voxels. Each voxel is represented as a pixel on the image, whose brightness corresponds to the attenuation level provided by corresponding volume unit.

In practice computed tomography images are built of a quite larger number of pixels, and values of μ for corresponding quantity of voxels should be calculated. In modern instruments the digital matrix of the obtained image has usually the size of 512×512 or 256×256 pixels.

Computer processing of the image allows distinguishing more than one hundred attenuation levels, i.e., densities of the investigated tissues: from 0 for water or liquor to >100 for bones. This allows differentiating normal and pathological tissue regions with the accuracy of 0.5–1%, which is 20–30 times better that using standard radiography.

The design of an X-ray scanner has been developed for a long time. Five generation of tomography systems can be mentioned.

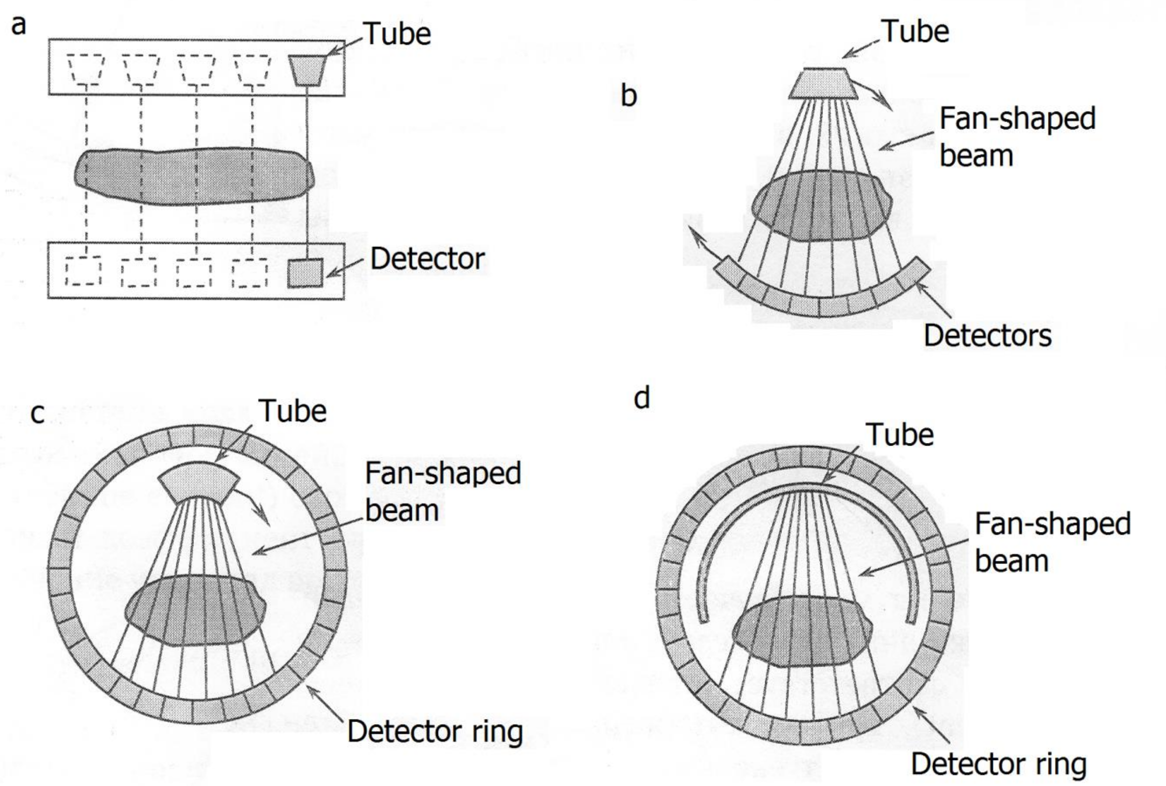


Fig. 6. Schematic representation of X-ray scanners.

Tomography instruments of the first generation, which appeared in 1973, had one narrow-beam X-ray tube (pencil beam) and one or two detectors, which moved synchronously along a ramp (Fig. 6a). The measurements were taken at 160 positions of the tube, then the ramp was rotated by 1°, and the measurements were repeated. The data was gathered for 45 minutes then processed on a special workstation during 2.5 hours. The small detectors rarely detected any scattered radiation, and this was a large advantage of this scanner.

The 2 generation instruments had several detectors working at the same time, and the X-ray tube produced a narrow fan beam (~30°). They used parallel scanning, as well as the 1 generation systems, but the range of the tube rotation angle increased to 30°, and the amount of linear displacement required was dramatically reduced. Total duration of obtaining one image decreased to 2–3 minutes per slice of a head CT slice. Resolution, however, suffered to some extent due to the new beam geometry and the fact that the detectors were exposed to more scattered radiation.

In the 3 generation systems (middle 1970's, Fig. 7b) the tube emitted a wide fan-shaped beam which irradiated a large set (about 400–1000) of detectors placed on an arc. The wide beam could reach the entire patient (slice) at one time. The improved construction (with a ring rail providing the power supply for the tube) allowed continuous rotation of the tube and detectors over 360°. This allowed avoiding the stage of moving the tube, and thus diminished the time required to get one image to 20 s and less. Due to the remarkable shortening of the procedure, scanners of this type were capable of studying moving parts of the body (lungs and abdomen) and made possible developing the helical data collection algorithm. They are sometimes called "rotate/rotate" scanners, unlike previous generations of "rotate/translate" systems. Nowadays, third generation CT scanners are still in existence.

Tomography systems of the 4 generation ("rotate/stationary" geometry, Fig. 7c) had a closed fixed ring of detectors (~5000) and an X-ray tube, which produced a fan-shaped beam and was turned around the patient inside the ring (outside disposition is also possible). By removing the detectors from the rotating gantry and putting them in a stationary ring around the patient, detectors were able to maintain calibration, which allowed suppressing so-called *ring artefacts* in the CT images. The scanning time for each slice was as small as 0.7 s, and an improvement in the image quality was achieved.

In early 1980's the electron-beam scanners (the 5 generation, Fig. 7d) were invented. A stationary cathode-ray (electron) gun placed behind the scanner ejects an electron beam, which passes through vacuum and is then focused and directed by electromagnets onto a tungsten target. The target has the shape of a circle arc (~210°) and is placed under the table for the patient. The targets are arranged in four rows, are very massive and have a flowing water cooling system. Opposite to the targets, an array of fast-response solid-body detectors is mounted, which is also stationary. Detectors form an arc of 216°. Scanners of this type are used in investigations of the heart, as they can produce an image in 33–50 ms (up to 30 frames per second!), and the number of slices is not limited by heat capacity of the tube. This generation was invented specifically to cardiologists; it was very expensive and not very versatile.

The 6 generation ("helical," or "spiral") scanners were introduced in practice in 1990's and combine the principles of the third and fourth generations with the slip ring technology to create a system that could rotate continually around the patient without being limited by electrical wires. They require much shorter acquisition times (i.e., as short as 30 seconds to scan the entire abdomen). The main drawback of helical CT scanners lies in the nature in which the data is collected. X-ray source and detector array rotate continuously as the patient table is moved progressively through the scanner, and thus no full slices (planar sections) of data are available. This problem can be compensated for through the reconstruction process.

The 7 generation scanners employ a new geometry of the X-ray beam. It does not pass through a narrow collimator and thus has a cone shape. Accordingly, detectors are placed in a planar array instead of a linear array, and cover a large area. A very large number of slices is acquired in a very short period of time, and a much higher level of sophistication in the reconstruction process is needed.

Laboratory work

Introduction

This laboratory work is dedicated to studying basic principles of an X-ray computed tomography scanner. A series of plastic details and a staffed frog serve as the objects of investigation. The goal is to obtain 3D scans of these objects, investigate their interior, find and measure the sizes of the bodies hidden within the study objects.

Accessories

- X-ray apparatus
- X-ray tube (with golden cathode)
- Chamber for placing the objects
- Model objects:
  1. plastic egg
  2. tower made of 5 square grey Lego cubes
  3. tower made of 3 rectangular coloured Lego cubes
  4. stuffed frog
- Computer (laptop)

Safety notes

The X-ray apparatus fulfills all regulations governing an X-ray apparatus and fully protected device for instructional use and is type approved for school use in Germany (NW 807/97 Rö).
The built-in protection and screening measures reduce the local dose rate outside of the X-ray apparatus to less than 1 µSv/h, a value which is on the order of magnitude of the natural background radiation.
Before putting the X-ray apparatus into operation inspect it for damage and to make sure that the high voltage is shut off when the sliding doors are opened (see Instruction Sheet for X-ray apparatus).
Keep the X-ray apparatus secure from access by unauthorized persons.
Do not allow the anode of the X-ray tube to overheat. When switching on the X-ray apparatus, check to make sure that the ventilator in the tube chamber is turning.

Preparing the experimental setup for measurements

1. Place the camera on a rail end opposite to the X-ray tube and secure it with the screw.
2. Put the object 1 (plastic egg) in the centre of the holder.
3. Close the glass doors of the apparatus and turn it on by the power button on the side panel.
4. Make sure that the apparatus and the camera are connected to the computer via a USB cable. Launch the “Computed tomography” programme.

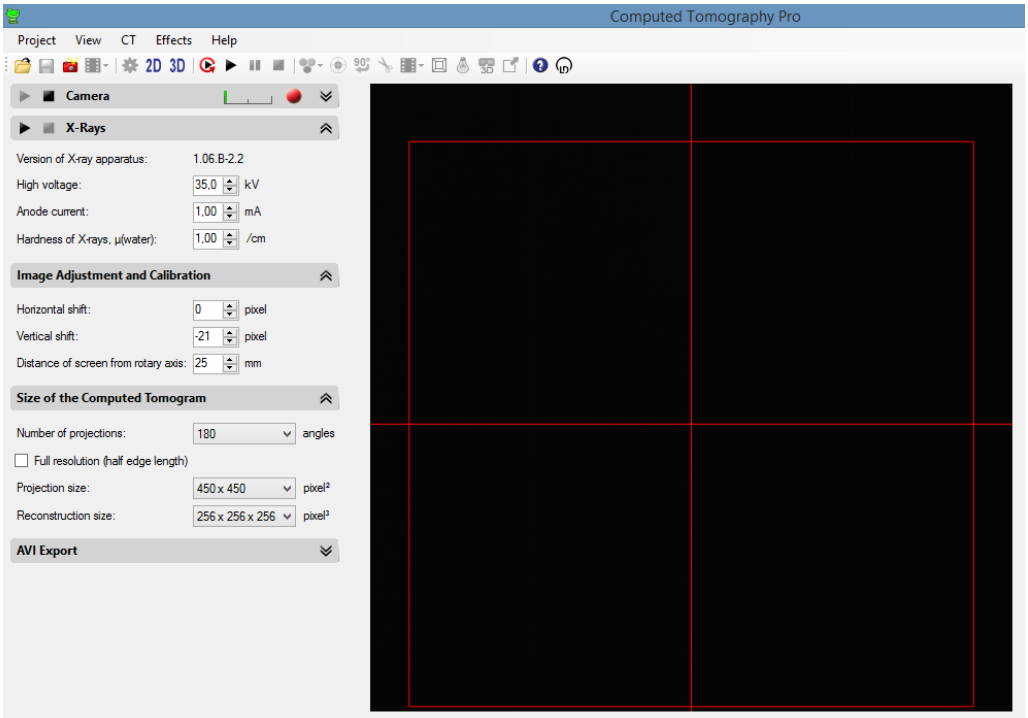


Fig. 1. Interface of the “Computed tomography” programme.

5. A red circle should flash on and off in the “Camera” line, indicating that the camera is connected.
6. Press the button ► in the “X-rays” line. The doors of the apparatus will be blocked, and the tube will start glowing. In the window on the right-hand side the image of the studied object will appear instead of the black square.
7. Move the centre of the red cross to the centre of the object using parameters “Horizontal shift” and “Vertical shift.” Set the distance from the rotation axis of the studied object to the camera in the line “Distance of screen from rotary axis.”
8. Choose the size of the recorded area (“Projection size” parameter) in the “Size of the Computed Tomogram” line so that the studied object’s image is within the red square. After that choose an appropriate “Reconstruction size” parameter (this value cannot be bigger than the projection size).
9. “Number of projections” parameter sets the number of rotation angles. The larger is this value, the better is the resolution of the image; however, the experiment duration also increases. Choose the value 180.

Algorithm of measurements

Exercise 1

1. Choose the CT menu in the upper bar of the programme window and press "Start CT scan."
2. In the appearing dialog, choose the folder to save your results and name of the experiment.
3. You can check experiment results using buttons 2D and 3D on the instrument bar.
4. Switch to the 3D mode. Move the cursor to the object's image. You can turn the image by pressing and holding the left button of the mouse. To zoom in and out, press and hold the right button and move the mouse forward or backward. Slices of the image can be changed by the roller of the mouse.
5. Colouring of the image may be set in the "Effects" – "Colour spectrum" menu.
6. Observe what happens when you modify the "Intensity" and "Transparency" parameters.
7. Write down what is inside the egg.
8. Choose the menu item "Reset Angle and Zoom" from the "Effects" menu; choose the perpendicular orientation of the egg by the button "Rotate by 90°." Move the cursor to the edge of the egg and press once the left button of the mouse. Using the ruler which will appear on the screen, measure the egg's diameter. Write down the value you obtain.

Exercise 2

1. Put the object 2 (grey Lego block) in the centre of the holder.
2. Repeat steps 5–9 and start the experiment.
3. In the "Effects" – "Colour spectrum" menu choose the "5 colours" regime.
4. Find the foreign body inside the object, describe its position (in the centre, on the side, etc.), measure its sizes (length, width, height).

Exercise 3

1. Put the object 3 (coloured Lego block) in the centre of the holder.
2. Repeat steps 5–9 and start the experiment.
3. In the "Effects" – "Colour spectrum" menu choose the "5 colours" regime.
4. Find the foreign body inside the object, describe its position (in the centre, on the side, etc.), measure its sizes (length, width, height).

Exercise 4

1. Put the object 4 (frog). Make sure that the sample on the platform does not touch the camera during rotation! Before you begin the measurements, show the setup to the teacher or engineer.
2. Set the "Number of projections" to 360, and tick the checkbox "Full resolution." Launch the experiment.
3. Investigate the object changing the "Intensity" and "Transparency" parameters. Obtain the reconstruction of the skin, skeleton, and viscera of the frog.