Ministry of Higher Education and Scientific Research

Al-Ma’moun University College

كليــة المــأمون الجامعــة

للعام الدراسي (2023–2024)

Physics of Computed Tomography

Lecture (1)

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Physics of Computed Tomography

Introduction and overview

Computed Tomography (CT) is a medical imaging technique that uses X-rays to obtain structural and functional information about the human body. The digital geometry processing can be used to generate a three-dimensional image of the internal structures of the human body from a large series of two-dimensional X-ray images taken around a single axis of rotation. It is also called a computed axial tomography scan. The origin of the word "tomography" is derived from the Greek word *tomē* ("cut") or "tomos" meaning "slice" or "section" and *graphein* meaning ("drawing"). A CT imaging system uses computer-processed X-rays to produce tomographic images or 'slices' of specific areas of the body, like the slices in a loaf of bread. Computed tomography (CT) has the capability to provide a different form of high-quality imaging. Computed tomography known as cross-sectional imaging is used for diagnostic procedures and visualization to guide therapeutic procedures in various medical disciplines. On the other hand, the radiation dose imparted to the patient's body during a procedure is a relatively high dose compared to radiography. But in spite of that, the computed tomography (CT) is now one of the most effective imaging methods and value to medical diagnosis and guidance of therapeutic procedures.
In CT scanning, both the **image quality characteristics** and the **radiation dose depend on and are controlled by the specific imaging protocol selected for each patient.** The image quality has a complex combination of many adjustable imaging factors and is influenced by many technical parameters for each procedure. Therefore, there is the need to manage the radiation dose for each patient and balance it with respect to the image quality requirements. This can be achieved by adjusting the protocol factors for each procedure.

Operating Steps

Devices that consist of the basic components of a patient table which the patient to be scanned, the table moves the patient into the gantry and the x-ray tube rotates around the patient. The scanner gantry contains the rotating portion that holds the X-ray tube generator and detector array. As x-rays pass through the patient to the detectors, A computer system acquires and performing the necessary calculations to go from measurements to a viewable image

A high-voltage x-ray generator supplies electric power to the x-ray tube, which usually has **a rotating anode** and is capable of withstanding the high heat loads generated during rapid multiple-slice acquisition. The x-ray tube generator, detector array, collimators, and rotational frame are housed in moveable frame as ring shaped unit called the **gantry** (Fig. 1)

Figure (1)

Different Generations of CT Scanners

The following describes the evolution of CT technology over the last 30 years. The "Generation" Race

– 1st Generation - single beam, translate-rotate
– 2nd Generation - multiple beam, translate-rotate
– 3rd Generation - fan beam, rotate
– 4th Generation - fan beam, fixed ring
– Fifth-generation CT (The electron-beam scanner)

- And many other generation , we will mention in the next lectures.

Basic Principles of CT

The tissues and materials generally differ in their ability to absorb X-rays, where some substances are more permeable to X-rays while some others impermeable and therefore the different tissues seem different when the X-ray film is developed.
For example, the dense tissues such as the bones appear white on a CT film while the soft tissues such as the brain or kidney appear gray while the cavities filled with air such as the lungs appear black.

Fundamentally a CT scanner when undergoing X-ray through the patient makes many measurements of **attenuation** through the plane of a finite thickness cross section of the body and obtaining information with a detector on the other side. The X-ray tube and detector array interconnected and rotated around the patient during the survey period. Then assemble the data that is obtained in digital computers and integrate it to reconstruct a digital image of the cross section to provide a cross sectional image (tomogram) that is displayed on a computer screen. That is a CT image is composed of *pixels* (picture elements).

Digital images

Visible light is essentially electromagnetic radiation with wavelengths between 400 and 700 nm. Each wavelength corresponds to a different color. On the other hand, a particular color does not necessarily correspond to a single wavelength. Purple light, for example, is a combination of red and blue light. In general, a color is characterized by a spectrum of different wavelengths.
The human retina contains three types of photoreceptor cone cells that transform the incident light with different color filters. Because there are three types of cone receptors, three numbers are necessary and sufficient to describe any perceptible color. Hence, it is possible to produce an arbitrary color by superimposing appropriate amounts of three primary colors, each with its specific spectral curve. In an additive color reproduction system, such as a color monitor, these three primaries are red, green, and blue light. The color is then specified by the amounts of red, green, and blue. Equal amounts of red, green, and blue give white. In practice, white light sources approximate this property. In a subtractive color reproduction system, such as printing or painting, these three primaries typically are cyan, magenta, and yellow. Cyan is the color of a material, seen in white light, that absorbs red but reflects green and blue, and can thus be obtained by additive mixing of equal amounts of green and blue light. Similarly, magenta is the result of the absorption of green light and consists of equal amounts of red and blue light, and yellow is the result of the absorption of blue and consists of equal amounts of red and green light. Therefore, subtractive mixing of cyan and magenta gives blue, subtractive mixing of cyan and yellow gives green, and subtractive mixing of yellow and magenta gives red. Subtractive mixing of yellow, cyan, and magenta produces black (only absorption and no reflection)

Note that equal distances in physical intensity are not perceived as equal distances in *brightness*. Intensity levels must be spaced logarithmically, rather than linearly, to achieve equal steps in perceived brightness.

Most digital medical images today use 4096 gray values (12 bpp). In the process of digital imaging, the continuous looking world has to be captured onto the finite number of pixels of the image grid. The conversion from a continuous function to a discrete function, retaining only the values at the grid points, is called *sampling*. Much information about an image is contained in its *histogram*. The histogram *h* of an image is a probability distribution on the set of possible gray levels (calculated according to Forurier transformation). The probability of a gray value *v* is given by its relative frequency in the image, that is,

$$h(v) = \frac{number\ of\ pixels\ having\ gray\ value\ v}{total\ number\ of\ Pixles}$$

Image quality

The *resolution* of a digital image is sometimes wrongly defined as the linear pixel density (expressed in dots per inch). This is, however, only an upper bound for the resolution. Resolution is also determined by the imaging process. The more blurring, the lower is the resolution. Factors that contribute to the unsharpness of an image are:

(1) The characteristics of the imaging system, such as the focal spot and the amount of detector blur
(2) The scene characteristics and geometry, such as the shape of the subject, its position and motion
(3) The viewing conditions.

Resolution can be defined as follows. When imaging a very small, bright point on a dark background, this dot will normally not appear as sharp in the imaging as it actually is. It will be smoothed, and the obtained blob is called the *point spread function* (PSF) (see the following Figure). An indicative measure of the resolution is the full width at half maximum (FWHM) of the point spread function. When two such blobs are placed at this distance or shorter from each other, they will no longer be distinguishable as two separate objects. If the resolution is the same in all directions, the *line spread function* (LSF), i.e., the actual image of a thin line, may be more practical than the PSF

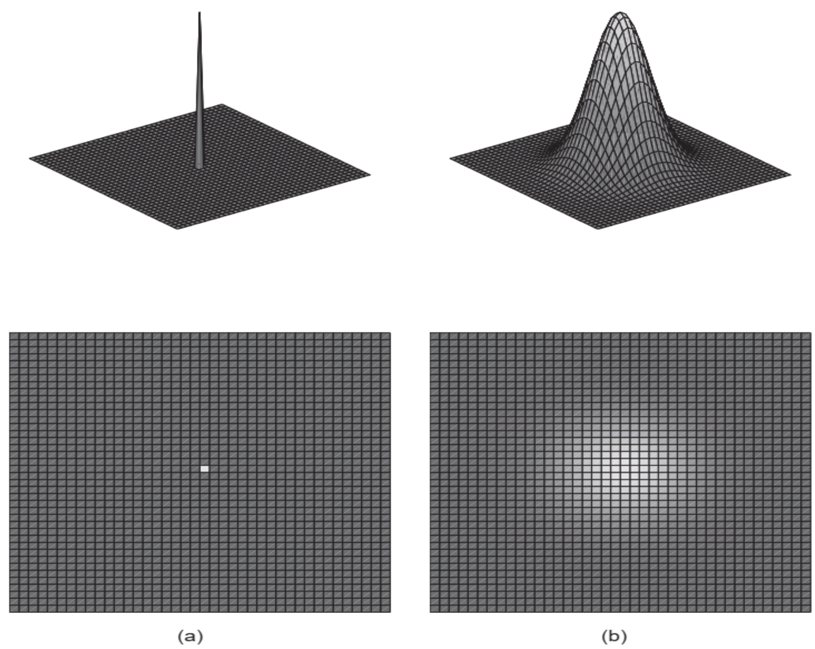


Figure 1.4 **(a)** Sharp bright spot on a dark background. **(b)** Typical image of (a). The smoothed blob is called the point spread function (PSF) of the imaging system.

Figure (2)

Instead of using the PSF or LSF it is also possible to use the *optical transfer function* (OTF) (see the following Figure). The OTF expresses the relative amplitude and phase shift of a sinusoidal target as a function of frequency. The modulation transfer function (MTF) is the amplitude (i.e. MTF = |OTF|) and the phase transfer function (PTF) is the phase component of the OTF. For small amplitudes the lines may no longer be distinguishable. An indication of the resolution is the number of line pairs per millimeter (lp/mm) at a specified small amplitude (e.g., 10%)

Contrast is the difference in intensity of adjacent regions of the image. More accurately, it is the amplitude of the Fourier transform of the image as a function of spatial frequency.
A third quality factor is image *noise*. The emission and detection of light and all other types of electromagnetic waves are stochastic processes. Because of the statistical nature of imaging, noise is always present. It is the random component in the image. If the noise level

is high compared with the image intensity of an object, the meaningful information is lost in the noise

. Each pixel on the image represents a measurement of the average x-ray attenuation of a box-like (small volume) element (voxel) extending through the thickness of the tissue section. In addition, in a real CT image, all tissues within a single pixel would be the same shade of gray (see figure 3). The image can be stored for retrieval and use later.

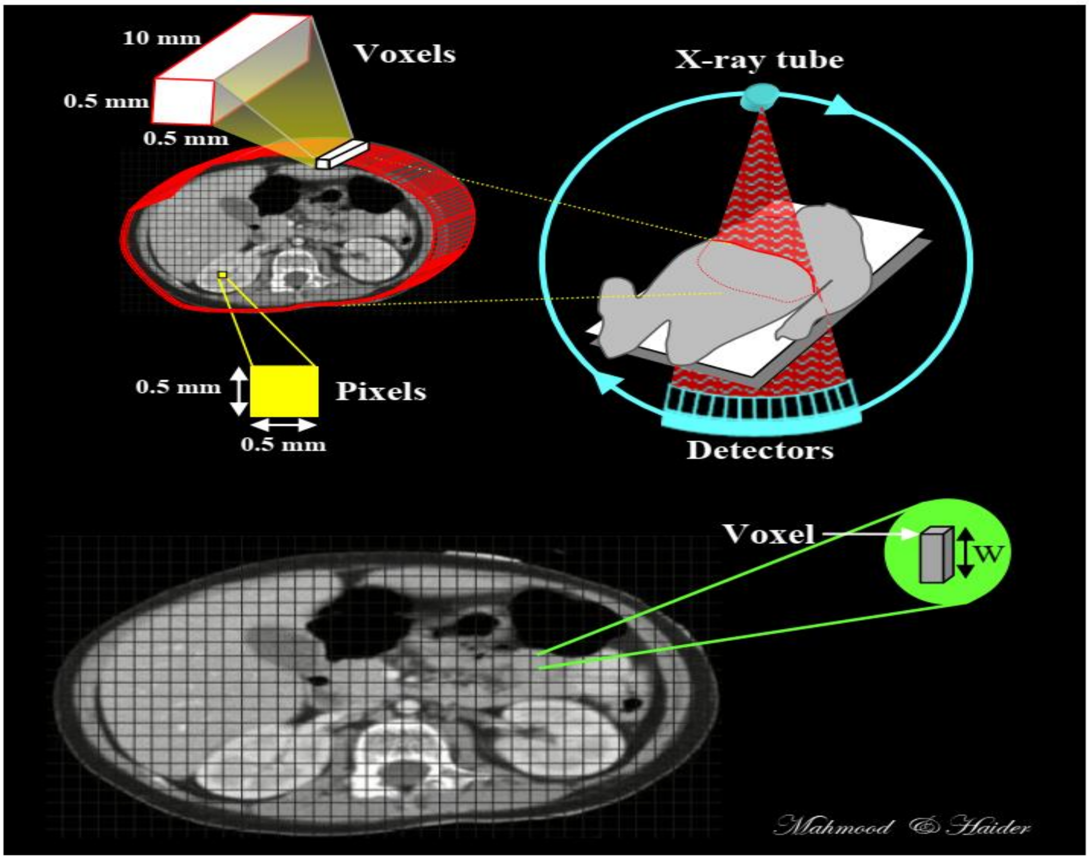


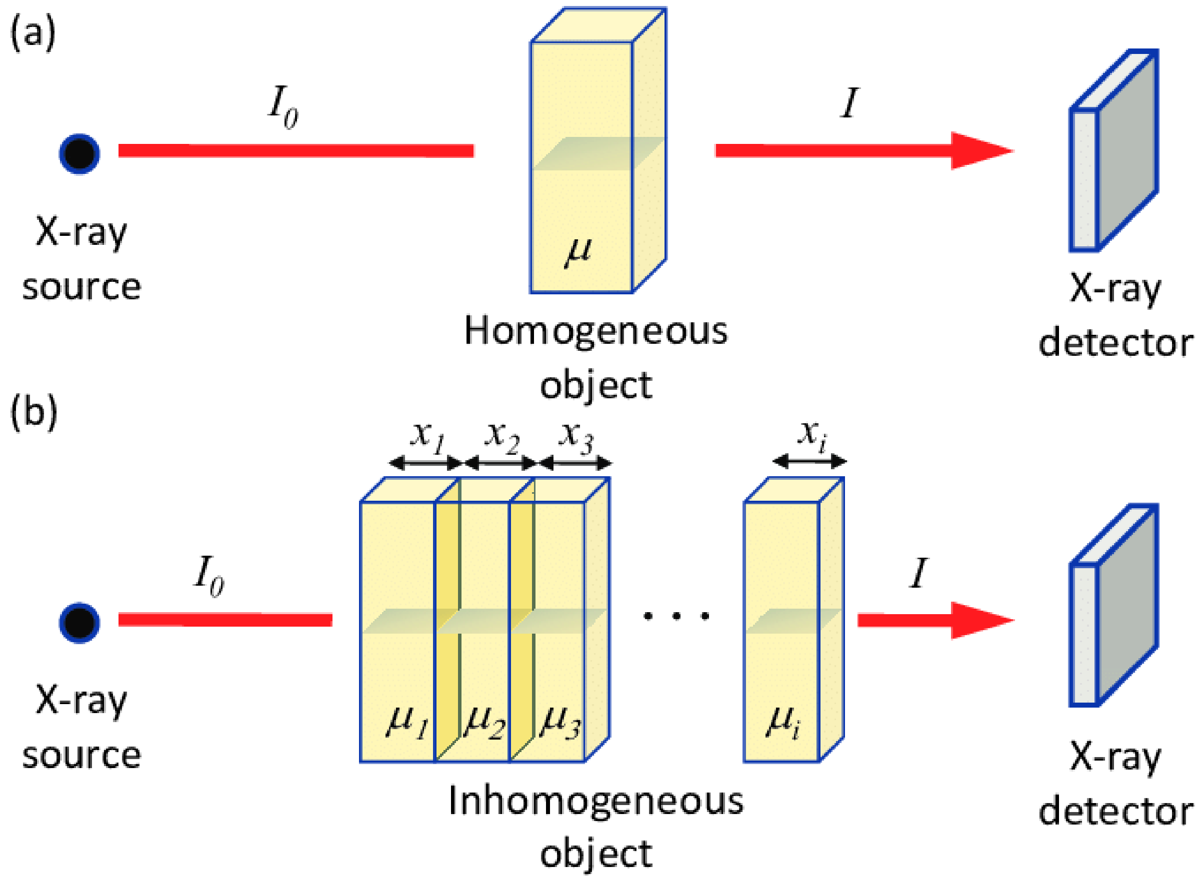
Figure (3)

When X-rays pass through the material subjected to attenuation which is the removal fraction of x-ray photons, as a result of tissue absorption and scatter, from the x-ray beam as it passes through matter. An attenuation measurement quantifies the fraction of radiation removed in passing through a given amount of a specific material of thickness Δx

Attenuation is expressed as:

$$I_t = I_o e^{-\mu \Delta x}$$

where, I_t and I_o are the x-ray intensities measured with and without the material in the x-ray beam path, respectively, and μ is the linear attenuation coefficient of the specific material.



Figure(4)

Attenuation of x-ray beam intensity in absorber system (nonhomogeneous layers like tissues).

Attenuation of an x-ray beam with a particular spectrum depends on two distinct properties of the tissue: the atomic number and the density of the attenuating material. When collecting one projection, it is not possible to determine what combination of atomic number and density resulted in the measured attenuation. Thus both very dense materials (e.g., bone) and materials with a high atomic number (e.g., iodine contrast media) produce strong attenuation. Both would have a similar appearance on CT even though they have very different physical properties. The key advantage of dual-energy and spectral CT techniques, is that they can be used to probe the attenuation arising from density and atomic number separately by making two different measurements of the same sample, object, or body part.

The transmission of the beam (the ratio $I:I_0$) determines the signal that is collected by the CT scanner detectors at each projection angle, and the attenuation differences along different paths through the patient's body result in the patterns that are formed in the data for each projection acquisition.