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RESEARCH ARTICLE

SIMPLE PHYSICAL PRINCIPLES AND MEDICAL USES OF COMPUTED TOMOGRAPHY (CT) SCAN A NEW ASSESSMENT

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Abstract

The development of X-ray computed tomography (CT) has been founded on the discovery of X-rays, the inception of the Radon transform, and the growth of X-ray digital data acquisition systems and computer knowledge. Dissimilar conventional X-ray imaging (general radiography), CT reconstructs cross-sectional anatomical images of the internal structures according to X-ray attenuation co-efficients (approximate tissue density) for almost every region in the body. This articleappraisals the essential physical principles and practical aspects of the CT scanner, including numerous notable evolutions in CT technology that resulted in the appearance of helical, multidetector, cone beam, portable, dual-energy, and phase-contrast CT, in integrated imaging modalities, such as positronemission- tomography—CT and single-photon-emission-computed-tomography-CT, and in clinical applications, including image acquisition parameters, CT angiography, image adjustment, versatile image visualizations, volumetric/surface rendering on a computer workstation, radiation treatment planning, and target localization in radiotherapy. The understanding of CT characteristics will provide more effective and accurate patient care in the fields of diagnostics and radiotherapy, and can lead to the improvement of image quality and the optimization of exposure doses.

Introduction:-

The discovery of X-ray radiation has been a major scientific breakthrough. This has been attributed to Wilhelm Conrad Röentgen (Germany) who conducted the first cathode tube experiments (Crookes tube) in November, 1895. This radiation has fluorescence characteristics, sensitizes the film, and penetrates opaque objects. Röentgen called this radiation type X-ray radiation. X-rays are irradiated and penetrate through three-dimensional objects. Xrays which penetrate through objects are recorded on films or detectors as two-dimensional images. This is generally referred to as radiography. Information disappears owing to overlap, tissues with minute absorption coefficients

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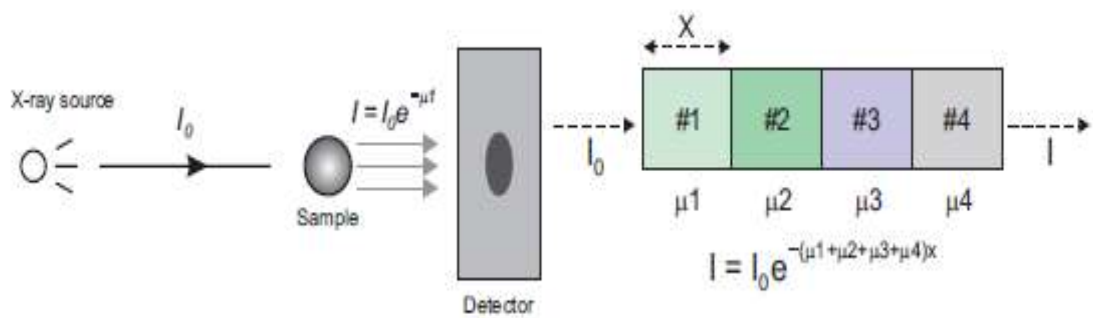
cannot be easily discriminated, and scattering X-rays cause adverse influences on imaging formation. The term "computed" in CT (computed tomography) indicates calculated or reconstructed, and the term "tomography" is a compound word comprising the term "tomo" (which meaning to "cut" or "section" in Greek) and "graphy" (which means "to describe" in Greek). CT scanner operations are based on X-rays. The typical energy used in general CT is in the range of 100 kV to 150 kV. The algorithms used for CT image reconstruction are based on mathematical foundations of the Radon transformation (Radon theorem) published in 1917 by Johann Radon (Austrian mathematician), who also provided the formula for the inverse transform. The Radon theorem states that the image reconstruction is possible from projections: an image distribution function could be obtained from an infinite set of its projections acquired by the rotational scanning. In the early 1960s, Allan M. Cormack, an American physicist with a South African origin, published the mathematical computation technique based on which the cross-sectional images of internal distributions could be calculated from projection of attenuation data owing to X-rays which penetrated the body at different angles based on the rotation of the X-ray source and electronic detector around a three dimensional object. The X-ray-based CT scanner was invented in 1972 by the British engineer Godfrey N. Hounsfield. Hounsfiled shared the Nobel Prize in Physiology and Medicine jointly with Allan M. Cormack for the development of CT in 1979. The first clinical CT scanners included a head scanner which was installed at the Atkinson Morley Hospital, London, UK, in 1972, and a whole-body scanner at Georgetown University Medical Center, USA, in 1976. In the early stages of head CT scanners, the image acquisition characteristics were associated with an acquisition time of 7 minutes, an image matrix of 80×80 pixels, a scan field of 25 cm, and a spatial resolution of 1.3 mm (≈ 4 lines per [lp]/ cm).

Basic Physical Principles ofComputed Tomography
X-ray attenuation through an object

The X-rays propagate through the sample wherein some of the X-ray photons are absorbed, and others are transmittedto the detector. The general form of X-ray attenuation is Ix=I0·exp–μx, where I0 is the X-ray intensity before the object is reached, Ix is the X-ray intensity after they penetrate through the object with a thickness x (where exp=e=the exponential coefficient=2.7182818…), μ is the X-ray linear attenuation coefficient, and x is the thickness of the absorbing material in chosen distance units, e.g., in mm. If X-rays pass through multiple objects with a total thickness x, the penetrated X-ray intensity can be expressed according to I=I0·e–μ1x×e–μ2x×e–μ3x×e–μ4x=I0·e–(μ1+μ2+μ3+μ4)x, known as the Lambert–Beer law. Therefore, the total linear attenuation coefficient can be written as the sum of contributions of μtot=μ1+μ2+μ3+μ4. Thus, the intensity of the X-rays measured on a position of the detector is proportional to the integral of the two-dimensional (2D) transparency of the object, and is exactly equal to the sum of the linear attenuation coefficient along their passage.



Computed tomography image data acquisition

To acquire a CT image, an object is placed on a table. An X-ray source (within the gantry of a CT scanner) rotatesaround the object, and X-rays which pass through the object are detected on the opposite side. The detections acquired at different angles are sent to the data acquisition system (DAS). These provide projection data to form tomographic images. The basic requirements of CT image acquisition include a) the fact that one tomographic image is reconstructed from X-ray projection data of the object acquired at various angles for general 360° (or 180°) rotations and b) during the scan, objects should be included in every projection dataset, and the object has to be still.

Computed tomography image reconstruction

Image reconstruction in CT is a mathematical process that calculates the 2D cross-sectional attenuation distributionfunction [f(x,y)] from a series of one-dimensional (1D) X-ray projections [P(r,θ)s] acquired as the line integrals at many different angles around the three-dimensional (3D) object. A projection, [P(r,θ)], is formed by the attenuated set of parallel X-rays (or isotropic X-rays emitted from a point source) through a 2D object of interest. A collection of projections at several angles during a single rotation of the X-ray source-detection system is called a sinogram. Effectively, this is a linear transform of the cross-sectional image of the object. Sinograms simply display all of the different projections for any slices stacked together. In reality, we can know the intensity I0 of the incident X-ray beam and can measure the intensity I of the detected X-ray beam. As a result, what we want to know are the distributions of individual X-ray attenuation coefficients of the object. Image reconstruction in CT involves the calculation of these individual coefficients in the internal structure of the object. There are a lot of methods that can be used for CT image reconstruction: matrix inversion, iterative, back projection, analytical, 2D Fourier transformation, and filtered backprojection methods in conventional CT; and 3D Radon transform, 3D filtered back projection methods in spiral CT or conical beam CT (CBCT), and nutating slice. Algorithms that reconstruct tilted image planes are adapted to the spiral path so that the rays are close to the image plane. The iterative reconstruction (IR) method is an algebraic reconstruction technique that begins with an image assumption: for a pixelated image value, a single ray in a particular projection is simply the sum of all the pixels the ray passes through on its way through the object, and its value is adjusted based on the difference of estimated and measured values until the two are in agreement. When CT was invented in the 1970s, the original reconstruction algorithm was the IR, which took approximately 45 minutes to reconstruct a single slice. Subsequently, a faster filtered back projection (FBP) algorithm was introduced, which has been used in CT imaging for more than 30 years. However, the FBP has problems associated with the relatively high noise levels. The backprojection method inversely “smears back” the projection across the image plane at the measurement angle for all of the projections. To improve the blurring problem that occurs in back projection, FBP was introduced with filtering to alter (or modify) the projections with a reconstruction Kernel before standard back projection was applied. The central slice theorem (CST) (also known as the projection slice theorem or Fourier slice theorem) is the theorem that relates the Radon with the Fourier transform, which is the most common reconstruction method currently used in the medical imaging field. Our goal is to identify a function f(x, y) of the attenuation coefficient map of the object in the x-y rectangular coordinate system from the P(r, θ) function of the projections (sinogram) in polar coordinates. It is always effective to know how various operations in the f(x, y) space are related to those in the Fourier F(u,v) space. The CST defines the relationship between the Radon transform of the object and the 2D Fourier transform of the sinogram P(r, θ). As a result, the 1D Fourier transform of the projection measured at an angle θ is the same as the radial slice acquired through the 2D Fourier domain of the object at the same angle. Finally, the f(x, y) attenuation coefficient distributions are obtained by the inverse Fourier transform of the calculated coefficient distributions of F(u,v) in the frequency domain.[1,9]

CT scan: normal appendix

CT scan: appendicitis

CT scan showing the liver

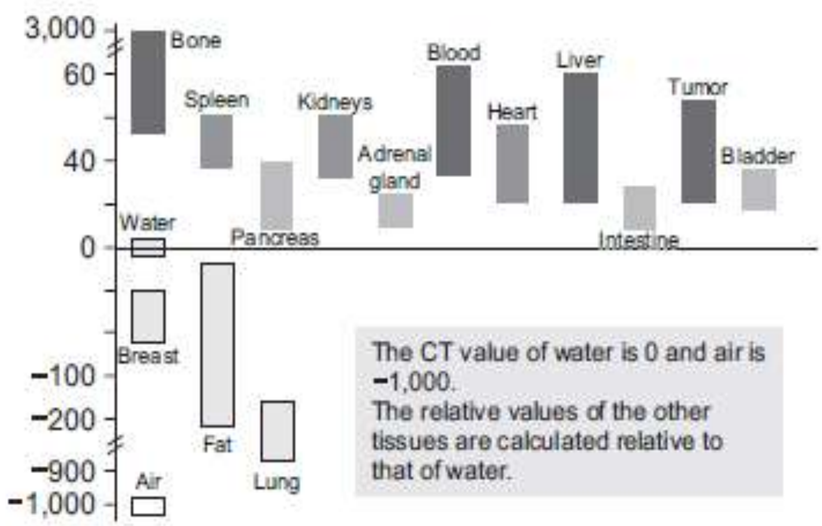
CT slice through mid abdomen

Computed tomography numbers/Hounsfield units

The image of a CT scanner is a digital image. It consists of a square matrix of pixels (picture elements), each of whichrepresents a voxel (volume element) of the interior structure of the object (patient). Given that a CT section has a finite thickness, each pixel actually represents a small volume element, or voxel (pixel area × slice thickness). The size of this voxel depends on the matrix size, the selected field of- view (FOV), and the section thickness. These pixel values are the average linear attenuation coefficients of the interior tissue corresponding to the spatial locations of these pixels, and the numerical pixel matrix corresponding to a spatial location in the image is converted into an image based on the allocation/transformation of the pixel values to a corresponding gray scale value. The typical CT image is constructed on a square matrix of 512×512 (262,144 pixels) and has an image depth of 12 bits (4,096 gray levels), but has increased to matrices composed of 1,024×1,024 or 2,046×2,048 (ultrahigh resolution CT) pixels in recent years. CT numbers (Hounsfield units [HU]) are defined as the attenuation values of the imaged tissues normalized to that of water. CT number (HU)=1,000× μpixel–μwater , μ water where μwater is the linear attenuation coefficient of water, μpixel is the linear attenuation coefficient of a given pixel. According to the definition of CT number (HU), the CT numbers of air and water range from −1,000 and 0, respectively. The CT numbers for various organs in humans.[15]

General classifications of computed tomography scanners

The CT scanner according to the data acquisition method is classified as traditional (axial) and spiral (helical) CT withthe fan-beam X-ray source used mainly for diagnostic purposes. In addition, CBCT scanners use the conical beam Xray source mainly for radiation therapy purposes. The conventional (traditional) CT scanner uses a fan-shaped X-ray beam, which takes a single slice image per scan and moves to the next slice scan position. The spiral (helical) CT scanner uses a DAS in which the fan-shaped X-ray beam and detector (or multidetector) move along a helical path relative to the object. In reality, the scanned object on the CT couch is moved to the bore of the scanner while the gantry rotates. CBCT scanner uses a conical beam X-ray beam which covers a large volume with a single rotation around the scanned object. Conversely, the multidetector CT (MDCT) is equipped with a 2D array of detector elements which replaces the linear array of detector elements (in a single slice CT) used in conventional and spiral CT scanners. The 2D detector array allows CT scanning to acquire multiple slices (or sections) of large volume simultaneously, and greatly increases the speed of CT image acquisition.[2,7]



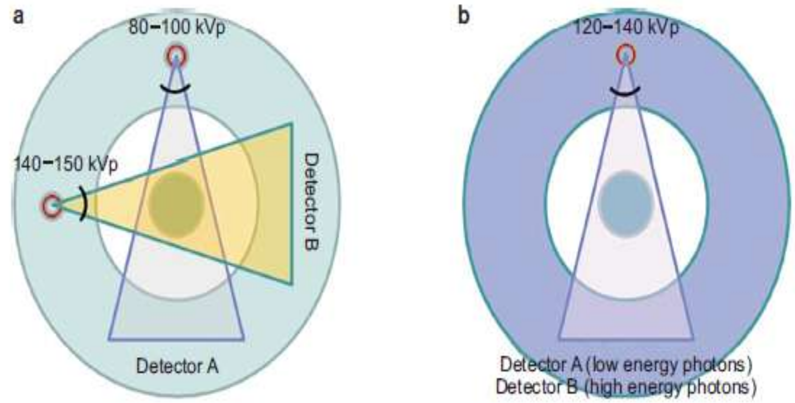
Components of computed tomography scanner

Basic, modern CT scanners are composed of the gantry equipped with the X-ray imaging acquisition system of Xraytube, high-voltage generator, filters, collimators, detector arrays, DAS, couch (patient table), operating console, and image reconstruction computer. The modern CT Xraytube has to sustain 60–80 kilowatts (kW) for up to 20 seconds on a focal spot which is as small as 1.3×10 mm. A highly stable, three-phase generator produces high voltages (generally between 120 to 140 kV) and supplies it to the Xraytube. The power capacity of the generator listed in kW determines the range exposure parameters, such as the Xraycharacteristics of kV and mA. The filter placed between the X-ray tube and the patient plays the role of removing low energy X-rays that do not contribute to image formation but increase patient dose. The collimator is placed between the filter and the patient.[6]

Its role reduces the radiation dose and restricts the scattered X-rays from the outer parts of the related slice. Different types of CT detectors were developed, including gas-filled detectors, scintillation detectors (solid-state detectors), and others. A gas-filled detector contains a single vessel filled with high-pressure gases (approximately 25 atm) of highatomicnumber elements (Krypton, Xenon, or Krypton and Xenon) and is divided in separate subdetectors. Early CT scanners used sodium iodide scintillation crystals (NaI) coupled with a photomultiplier tube. Most commonly used solid-state detectors convert the X-rays into visible light photons, which were converted into electrical signals through photodiodes. Solid-state detectors are also called scintillation detectors because of the use of crystals that fluoresce when X-ray photons interact with them. A photodiode transforms the scintillation lights to electrical signals. Patient couch (table) is made of carbon fibers to prevent it from interfering with X-rays due to their low-photon absorption properties. Patients lie on the couch and are moved through the gantry aperture to determine the scanning position and control the patients' movements after position settlement during examination. A CT couch should be strong and rigid to support weights up to 204 kg. The shape of the table used in radiation therapy is flat to reconstruct the patient image in the same condition as a treatment machine, while diagnostic CT uses a rounded table top for patient comfort. The gantry's bore size used in simulations of radiation therapy is bigger to accommodate situations in which the overall diameter of imaging volume is large owing to immobilization devices. Additionally, special patient postures are often needed for better treatments (85 cm for radiation treatment vs. 70 cm for diagnostic purposes). The operating console is the control center of the CT scan. It consists of a keyboard, multiple monitors, and computers which are used to operate the scanner, receive the data from the DAS, and reconstruct the CT image.[1,11,13]

Developments of computed tomography scanner

The first generation of CT scanners operated as follows. The X-ray pencil beam was measured by a single detector, and the X-ray tube and detector moved together (translation movement). The X-ray and detector then rotated together at a different angle, and these procedures were repeated for a single slice. The X-ray and detector then moved together to a different slice, and the process was then repeated. The second generation CT scanners were operated in the same way as the first generation scanners except for the simultaneous use of a row of up to 30 detectors incorporated with narrow fan X-ray beams. This method required approximately 5–90 seconds per slice to scan. The third generation CT scanner had a row of detectors which was wide enough to image the entire slice at a single angle orientation. Therefore, the translation movement was no longer necessary. The X-ray and detector rotated together at different angles, and this process was repeated until a single slice was scanned before acquisitions were conducted for a different slice (axial scanning). This method is most commonly used nowadays, and takes approximately 0.3 seconds to image a single slice. The fourth generation CT adopted a fixed complete ring of detectors. Therefore, the X-ray tube only rotated around a slice and moved to a different slice. This method is not commonly used today. The fifth generation CT electron beam scanners use a stationary X-ray tube and a stationary detector ring, thereby avoiding any mechanically moving parts like the case of the television tube. An electron beam is magnetically deflected onto a fixed array of tungsten anode targets which surround the patient. The magnetic deflection sweeps the electron beam over the target, thus creating an X-ray source that virtually rotates around the patient. Given the absence of mechanically moving parts, a sweep can be accomplished in as little as 50–250 ms. The sixth generation CT, helical (spiral) CT, was developed by Willi A. Kalender (German medical physicist) in 1989. The helical CT combined the priciple of the third and fourth CT generations with slip-ring technology. The patient was moved along the gantry direction through a contiually rotating X-ray beam and detector system in the gantry. Spiral CT is much more efficient in reducing the scanning time. The combination of continuous data acquisition with slip-ring-based volumetric data transmission and continuous table translation led to a considerable reduction of scanning time. The current slip-ring technology of X-ray tube and detector array system installed in the gantry can rotate at rates between 0.25 and 3 seconds per rotation. The seventh generation CT scanners, MDCT, was introduced in the late 1990s and early 2000s with a 2D multiple detector array which simultaneously scanned multiple slices in a single rotation. The development of MDCT has resulted in high-speed image acquisition which is proportional to the number of detector rows. The early, single-slice CT scanner had one row of detectors. However, nowadays all CT scanners are multislice detector arrays and have 8–320 rows of detectors. Multidetector row (or multislice) CT uses multiple rows of CT detectors instead of only one (even single-detector row CT has multiple detectors along the gantry, but only has one row, i.e., it only reconstructs a single slice along the z-direction). The advantage of MDCT is faster scanning, as more parts of the body are contained in each turn of the gantry. For MDCT, multiple slices are reconstructed per projection by the multiple rows of detectors. In MDCT, the pitch is calculated by dividing the tablemovement by the entire beam width. This yields numbers similar to the single-slice version: a pitch of 1 correspondsto contiguous helices.[6]



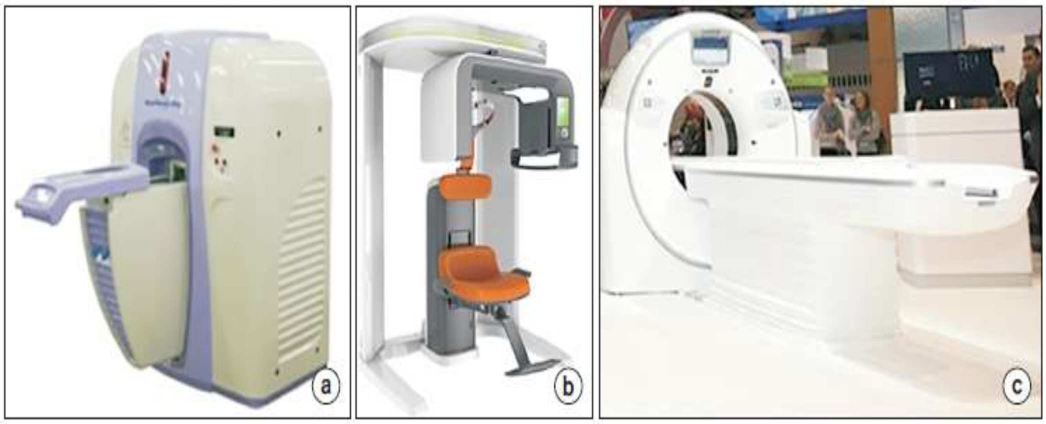
Clinical Applications of Computed Tomography

Computed tomography scanning parameters

Parameters responsible for the specifications and performances of X-ray CT scanners are closely related to eachother, and are various. The CT scanning time generally depends on the type of examination and the performance specifications of CT scanner. Short scan times are required to reduce the artifacts attributed to the patient's physiological motions, including the respiratory, digestive tract, peristaltic, and others. Gantry rotation time for a helical CT is defined as the amount of time required to complete one full rotation (360°) of the X-ray tube and detector around the object. The advances in technology have shortened the gantry rotation time to values as low as approximately 0.3 seconds. The common CT examination requires a scan length of 240 mm for the head and 400 mm for the chest or abdomen. In general, CT slice thickness (slice collimation) is typically 5 mm for a standard head CT scan, and ranges between 0.625 and 1.25 mm for both blood vessel and thin section facial bone CT, or 3D image processing. CT patient throughput is related to the number of examinations per hour, and indicates the ability of scan processing.

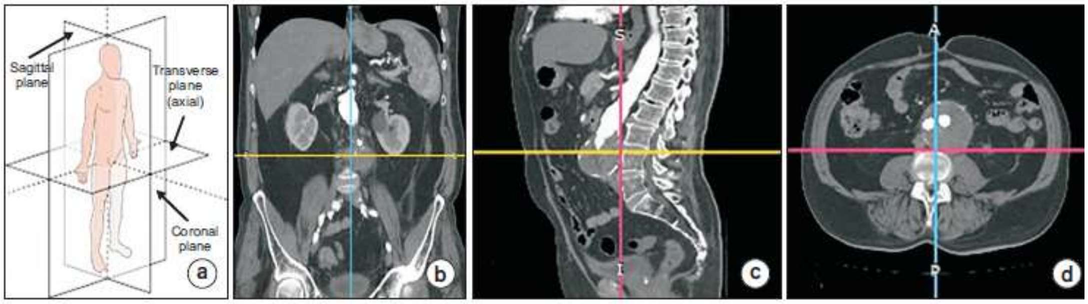
Causes and types of computed tomography image artifacts

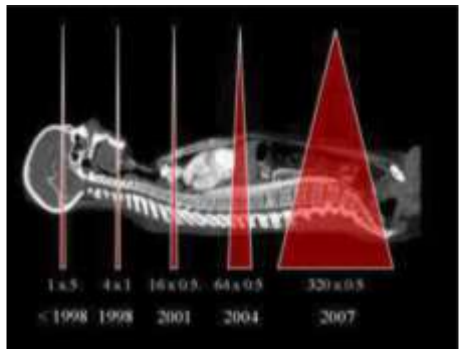
Fundamentally, image quality in CT, as in all other medical imaging modalities, is described based on spatial resolution,image contrast, image noise, and artifacts, which are closely related to each other. CT image quality is influenced by the imaging system (technical parameters), user (radiology technologist or radiologist during image acquisition and evaluation), and patient (patient's movement and size). Artifacts indicate features that appear in CT images that did not exist in the original clinic. This primary reason is attributed to the principal limitation of CT and the malfunction and maladjustment of CT itself. There are many different types of artifacts that originate from motion, metal, partial volume, beam hardening, technical defects, or operator errors. Motion artifacts are patient-based features produced by respiratory, muscular, or other movement by the patient during a scan. If patients move, motion artifacts can arise. Therefore, patient movements should be minimized to prevent these. Motion artifact can be minimized by using a motion artifact correction algorithm together with the control of breathing during image acquisition. Additionally, motion artifacts can be compensated for by the motion correction algorithm. Partial volume artifacts occur when tissues with considerably different X-ray absorption properties in complicated bone structures are included in the same CT slice. These result in a beam attenuation, which is proportional to the average value of each slice. This is because the very dense structures (bones) are only partially included in the slice, and thus result in high-contrast errors. These are attributed to the partial volume effect in the case of large structural variations toward the slice thickness direction. To reduce the partial volume artifacts, the use of a thin-slice thickness is recommended. Beam hardening is the phenomenon that occurs when an X-ray beam composed of different energies passes through a dense object, thus resulting in the selective attenuation of lower energy photons. The beam hardening effect is conceptually similar to a high-pass filter in that only higher energy photons are left to contribute to the beam. Thus, the mean beam energy is increased ("hardened"). The beam hardening effect gives rise to streaking (dark band) artifacts, which appear as multiple streaking bands positioned between two dense tissues or cupping artifacts, thus causing the edges of an object to appear brighter than the center. To effectively reduce the beam hardening effect, the X-ray spectrum has to minimize lower energy photons by adopting a metal filter before the CT scan. In this respect, CT scanners are equipped with metal artifact reduction algorithms. Metal streak artifacts are caused by multiple mechanisms, including beam hardening, scatter, Poisson's noise, motion, and edge effects in the scanned objects, including hair pins, clips, and metals inside the patient. Metals, such as gold-implanted objects within a human body absorb the X-ray beam almost completely, thus producing “radiation shadows” which generate pronounced dark and bright streak artifacts over the entire reconstructed image. This can only be avoided via a gantry tilt that excludes the disturbing metallic objects from the slice plane. Ring artifacts occur owing to the miscalibration or technical failure of one or more detector elements in a CT scanner. This is also caused by the insufficient X-ray intensity or the contamination of a contrast medium on the detector cover. They occur close to the isocenter of the scan and are usually visible on multiple slices at the same location. The ring artifacts appear as concentric rings superimposed on CT images. The ring artifact effects can be remedied by recalibrating the scanner’s malfunction detector system or replacing the malfunction detector elements.



Computed tomography image reformatting

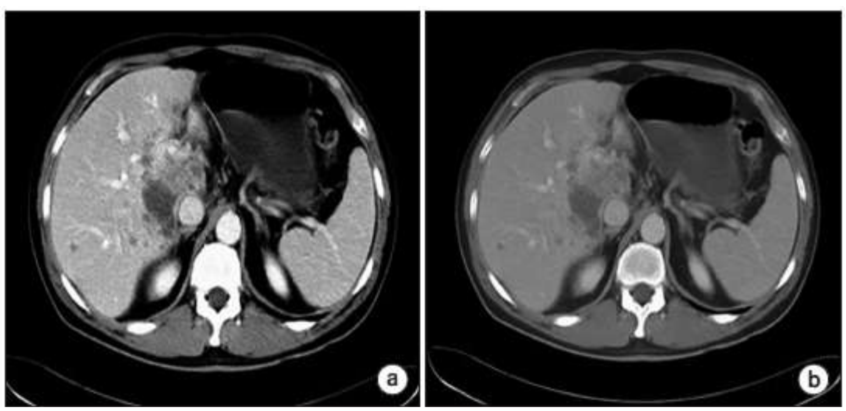
The most common application of X-ray CT refers to crosssectional (axial) images, which are used for diagnostic andtherapeutic purposes in various medical fields. The modern CT workstations are specifically designed for navigationand visualization of various reformatted images, such as the techniques of multiplanar reconstruction (MPR), maximum intensity projection (MIP), minimum intensity projection (MinIP), cine (movie) mode, shaded surface display (SSD, also known as surface rendering), 3D volume rendering, and virtual endoscopy (VE), such as angioscopy, bronchoscopy, and colonoscopy. The MPR technique involves the image processing tasks needed to convert thin-slice data from volumetric CT in the axial plane from another plane (such as coronal, sagittal, or oblique) to control the window level (WW) and center the image to view whichever structures needed with the aid of various software and a workstation. From the 3D reconstruction of a sequence of tomographic images, the MPR display can be used to generate interactive slices in Cartesian planes (axial, sagittal, and coronal), or even in oblique planes. This means that the physician can obtain views of the patient’s internal structures that provide greater clarification compared with the images from theoriginal sequence.The MIP technique is a method for 3D data that projectsthe voxels with the maximum intensity (highest Hounsfieldnumber) that fall in the way of parallel rays traced fromthe viewpoint to the plane of projection in the 2D viewingplane to identify all the hyperdense structures (boneor contrast material-filled structures) in a volume. TheMinIP technique is a 2D visualization method that enablesdetection of low-density structures in a specific volume.The MinIP images, mainly used to diagnose lung diseases, are multiplanar slab images produced by displaying onlythe lowest attenuation value encountered along a ray cast through an object toward the viewing plane. The surfaceshaded display represents a visualization technique, which is well established for 3D imaging of sectional image databy using edge detection image processing algorithms. SSD provides a realistic 3D view of the surface of a structureof interest within the specified volume by selecting attenuation thresholds. The volume rendering is a type of datavisualization technique, which creates a 2D projection from 3D imaging of sectional image data by means of a specifiedsoftware. In volume rendering, transparency, color, and shading are used to allow a better visualization of thevolume to be shown in a single image. Volume rendering is primarily conducted for better visualization of the humananatomy, surgical treatment planning, as well as for medical teaching.Sagittal





Clinical challenges of computed tomography scanner technology

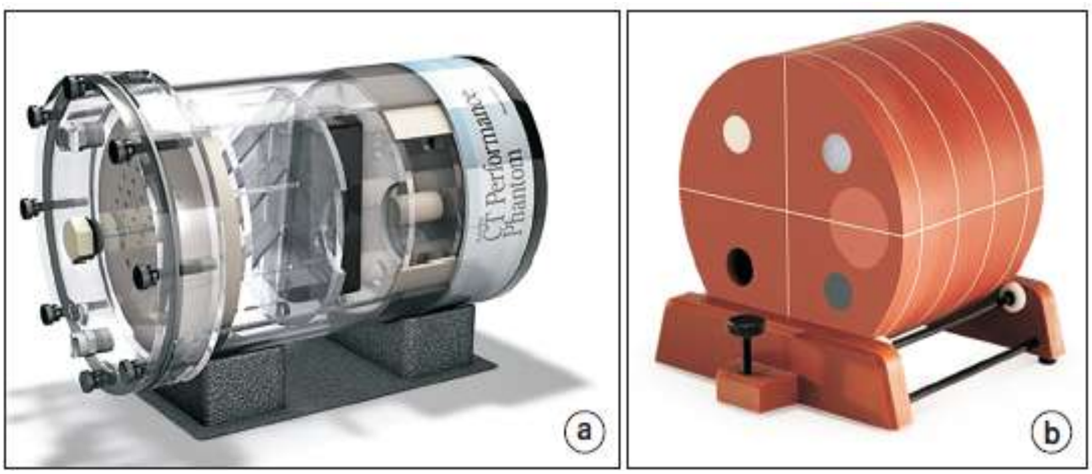
CT Angiography (CTA) is a type of medical examination that combines a CT scan with an injection of a special contrast media (dye) to enhance the signal intensity from arteries, veins, and tissues throughout the body. CTA



can be used to visualize the vessels of the heart, aorta, and other large blood vessels, lungs, kidneys, head and neck, and arms and legs. CTA images can be 3D reconstructed so that the cerebral vessels and accompanying pathology can be rotated and viewed from all angles. Contrast media are used to image structures that are not normally visible, or are not visible very distinctly. Intravenous contrast agents are used to enhance organs and visualize blood vessels. Oral contrast agents are used to visualize the digestive tract. The contrast media for CTA is injected through an intravenous line in the arm or hand. In general, CT contrast media is a colorless and transparent iodine compound, that is, a substance that X-rays cannot penetrate. In one CT exam, theamount of approximately 100–150 mL is injected into a vein at the rate of 2–3 mL/s with a CT contrast injector. The contrast media can lead to adverse reactions such as nausea, vomiting, and/or hives, uncommon allergic reactions, brief loss of consciousness, bronchospasm, kidney damage, and others. Before an external beam radiotherapy treatment planning process, the radiation oncologists plan to determine the type of cancer, its exact location, the size and shape of the cancer, the cancer staging, the critical structures closely located to the cancer, the radiation dose needed in the can cer treatment, and the patient's general health and medical history. CT simulators (CTSim) dedicated to radiotherapy are based on general diagnostic CT scanners, with few additional modifications. CTSim typically include a laser positioning and marking system as a reference for patient positioning, a larger bore size to accommodate patient immobilization devices, a flat tabletop to replicate the radiotherapy treatment unit couch, and a virtual simulation software workstation. The CTsim 3D imaging data of the patient's anatomy provide various information for more accurate delineation of the tumor and the surrounding normal tissues. In addition, the pixel HU values inherently include the associated tissue density information used to compute radiation dose distributions, which is a necessity for 3D radiotherapy treatment planning for X-ray radiation treatments. The conversion is accomplished with a phantom calibration, and annual checks are required. State-ofthe- art CTSim are also capable of acquiring 4D imaging data. The 4D CT simulations are applicable for the internal organs affected by respiratory motion. These advanced CT imaging simulation techniques allow accurate tumor localization, treatment visualization, and subsequent treatment delivery.

Quality control and imaging phantom of computed tomography

The periodic quality control (QC) (quality assurance) of CT is important to maintain proper performance of electromechanical components, optimum image quality, and reasonable radiation doses to the patients. The CT QC program is classified as the roles and responsibilities of the radiologist, radiologic technologist, and medical physicist, and the routine test cycle of weekly tests (some perform daily), monthly tests, semi-annual tests, and annual tests. The ACR recommended in the CT QC manual the minimum QC test frequencies, such as the annual tests of qualified medical physicist surveys, including participation in the review of clinical protocols with the CT protocol and management team, scout prescription and alignment light accuracy, table travel accuracy, radiation beam width, low-contrast performance, spatial resolution, CT number accuracy, artifact evaluation, CT number uniformity, dosimetry, CT scanner display calibration, radiological technologist QC tests of the water's CT number and standard deviation (daily), artifact evaluation (daily), wet laser printer QC (weekly), visual checklist (monthly), dry laser printer QC (monthly), and gray level performance of CT scanner acquisition display monitors (monthly). The CT performance phantoms used to CT image acceptance testing.



Conclusions:-

The understanding of CT characteristics will provide more effective and accurate patient care in the fields ofdiagnostics and radiotherapy, and can lead to the improvement of image quality and the optimization of the exposed doses.

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