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Medical imaging

Intended for third-year undergraduate students in Biomedical Engineering

Authored by

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Biography

Dr. Amel Korti is a university lecturer at Abou Bekr Belkaid university, Technology faculty, in Tlemcen, Algeria. She is a member of the biomedical engineering research laboratory. She is affiliated with the medical imaging team.

Dr. Amel Korti, a well-respected university lecturer specializing in medical imaging, brings theoretical knowledge and real-world insights to this comprehensive guide with her extensive academic background and practical experience.

In addition to her role as a university lecturer, Dr. Amel Korti has authored several influential biomedical papers, handbooks and books. Her publications, characterized by clarity and depth, serve as valuable resources for students, researchers, and professionals alike. Driven by a commitment to advancing knowledge in the biomedical field, she continues to inspire the next generation of biomedical engineers through her dynamic teaching and impactful literary contributions.

Avant-propos

Welcome to the fascinating world of medical imaging! In this comprehensive guide, we embark on a journey through the intricate landscapes of the human body, exploring the marvels of modern technology that allow us to visualize, diagnose, and understand the intricacies of our anatomy.

As the field of medical imaging continues to evolve at a breathtaking pace, it becomes imperative for healthcare professionals, students, and enthusiasts to stay abreast of the latest advancements. This coursebook serves as a bridge between the fundamentals and cutting-edge technologies in the realm of medical imaging.

From traditional X-rays to state-of-the-art magnetic resonance imaging (MRI), ultrasound and computed tomography (CT) scans, each chapter delves into the principles, applications, and clinical significance of different imaging modalities. The goal is to provide a comprehensive resource that not only educates but also sparks curiosity and a deep appreciation for the remarkable tools at our disposal.

Whether you are a seasoned healthcare professional seeking to enhance your skills or a student taking your first steps into the world of medical imaging, we hope this coursebook serves as a valuable companion on your journey.

Thank you for joining us on this exploration of medical imaging. May this coursebook inspire curiosity, deepen understanding, and ignite a passion for the incredible intersection of medicine and technology.

Authored by

Dr. Korti Amel

Preface

This coursebook explains the applied mathematical and physical principles of medical imaging and image computing. It offers something for everyone interested in the captivating world of medical imaging.

This course will cover five medical imaging modalities: projection radiography, X-ray computed tomography (CT), positron emission tomography (PET), magnetic resonance imaging (MRI) and ultrasound imaging. It also focuses on the acquisition and reconstruction of medical images too.

This course is the fruit of fifteen years' experience; it is an ideal handout for a one-semester course in medical imaging.

This course is dedicated to students registered in Biomedical Engineering in L3 GBM (semester 5) for all specialties.

LEARNING OUTCOMES

By the end of this course, the student will be able to:

1. Explain the basic principles of all medical imaging techniques;

2. Demonstrate an understanding of approaches to tomographic (CT, PET) medical image reconstruction;

3. Read and understand scientific documents in the field of medical imaging.

Abbreviation

CT	Computed Tomography
CR	Computed Radiography
DR	Digital radiography
PSP plate	Photostimulable phosphor plate
PMT	Photomultiplier tubes
SPECT	Single photon emission computed tomography
PET	Positron emission tomography
FBP	Filtered back projection
CWD	Continuous-wave Doppler
RF	Radio Frequency
MRI	Magnetic resonance imaging
NMR	Nuclear magnetic resonance
B0	Strong magnetic field
B1	Radiofrequency magnetic field
TR	Repetition time
TE	Echo time

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General introduction

Medical imaging is a remarkable field that plays a pivotal role in modern healthcare. It is the science and technology of visualizing the human body's internal structures, organs, and tissues to diagnose, monitor, and treat various medical conditions. This field has revolutionized medicine by providing non-invasive ways to peer inside the body, enabling healthcare professionals to make informed decisions about patient care.

Imagine a world where doctors can see the intricate details of your organs, blood vessels, and bones without the need for surgery. This is precisely what medical imaging accomplishes, and it has transformed the way healthcare is delivered.

This coursebook serves as your gateway into the captivating world of medical imaging. Whether you are a biomedical engineering student, a medical student, a healthcare professional, or simply someone intrigued by the wonders of technology and the human body, this course is designed to provide a comprehensive understanding of the field.

This manuscript is divided into six chapters:

The first chapter introduces radiography, where we describe the essential components of the device and the interaction of Xray with matter.

The second chapter explains the fundamental concepts of CT scanning, including its various components and operation.

The third chapter explores the fundamental concepts of PET scanning, detailing its various components and operation.

The fourth chapter discusses the reconstruction of tomographic images from diverse data sources.

The fifth chapter covers essential components of ultrasound devices and the interaction of ultrasonic waves with matter.

The end chapter is dedicated to magnetic resonance imaging, where we describe the essential components of the device and the image reconstruction process.

This manuscript concludes with a general summary.

By the end of this course, you will have a solid foundation in medical imaging, enabling you to appreciate its significance in healthcare and its potential for advancing the field of medicine.

Chapter 1
X ray radiography

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1. Introduction

Radiography is a traditional X ray imaging, enabling physicians to visualize anatomical strucures, diagnose diseases, and guide medical interventions. The field continues to evolve, incorporating cutting edge technologies to improve image quality, safety, and patient care. In the following section, a brief overview of the history of radiography is presented.

2. History

In 1895, the X ray was discovered by Wilhem Conrad Roentgen while experimenting with cathode rays. He notices that these mysterious rays can penetrate objects and create images on a photographic plate.

In 1896, Roentgen takes the first X ray image of his wife's hand, revealing the bones and her wedding ring. This ionic image demonstrates the potential of X ray for medical imaging.

In 1970-1980, Computed Radiography (CR) and Digital Radiography (DR) technologies begin to replace traditional film-based radiography, offering advantages in image storage, processing, and manipulation.

In 1990-present, Digital X-ray technologies continue to evolve, with advancements in image quality, dose reduction, and image sharing through Picture Archiving and Communication Systems (PACS).

Radiography refers to the medical imaging technique of using X rays to create images of the internal structures of the human body. Radiography allows healthcare professionals, particularly radiologists, to visualize bones, tissue, and organs to detect and diagnose a variety of medical conditions. In this chapter, we will present the various components of the X-ray machine and their operating principles.

3. Components of the X-ray machine

A radiography system comprises : an X ray source, an X ray detector, a collimator and an examination table on which the patient is placed for imaging (see Figure 1) :

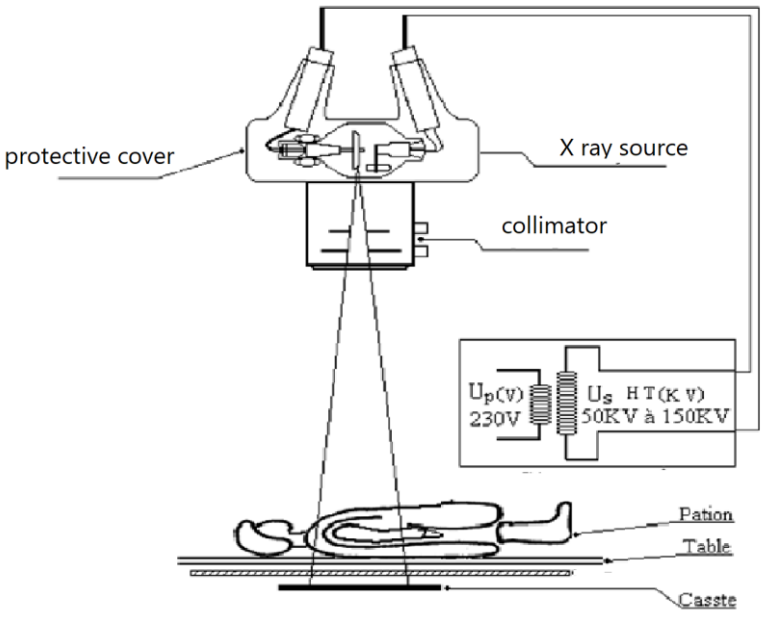


Figure 1.1. Radiography system components

3.1.X ray source

The X-ray source, often referred to as the X-ray tube, generates the X-ray photons that pass through the patient's body (Figure 2). It produces a controlled and focused X-ray beam with adjustable energy levels (kVp) and intensity (mA) that can penetrate the body's tissues and create detailes images of the internal structures.

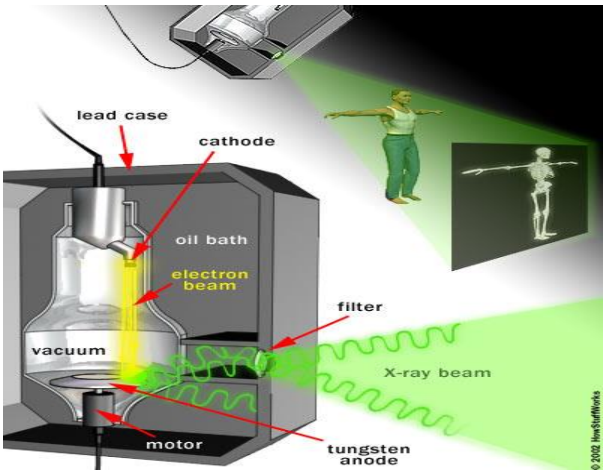


Figure 1.2. X ray source

An X-ray consists of a cathode and an anode (Figure 3) :

Cathode is the negative electrode, it consists of a heated filament that emits electrons when heated. These emitted electrons form a cloud around the filament.

Anode is the positive electrode.

X-rays can be produced by the anode through two main mechanisms: Bremsstrahlung radiation and characteristic radiation. They can both occur in the heavy atoms of tungsten. Tungsten is often the material chosen for the anode of the x-ray tube.

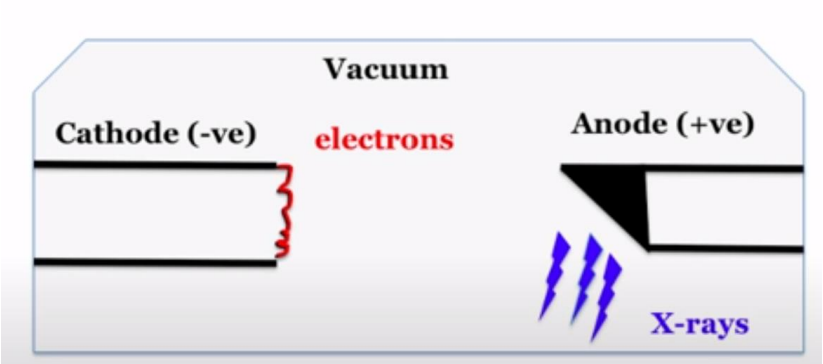


Figure 1.3. Inside the X ray source

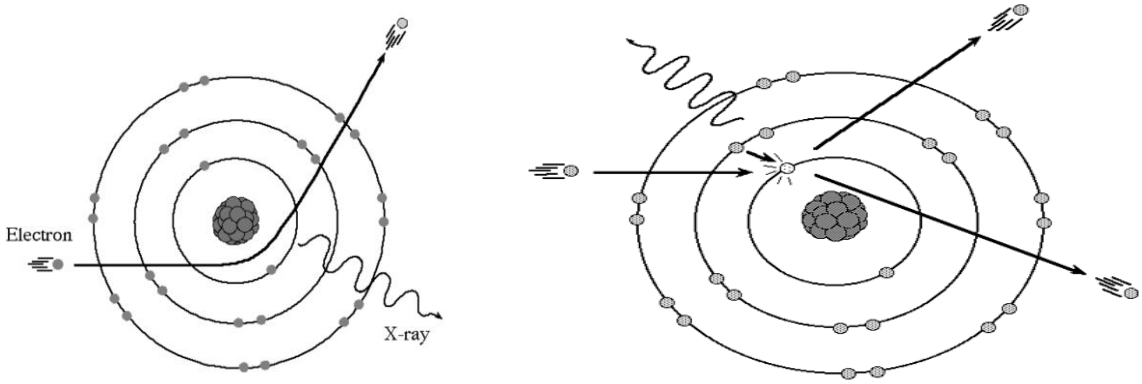
- **Bremsstrahlung radiation :**

A stream of high speed electrons is produced by the cathode. These electrons are accelerated towards the positively charged anode. As the high speed electrons approach the positively charged nucleus of the anode material, they experience a strong electric field. The field causes the electrons to be deflected and decelerated. As the electrons are slowed down or change direction, they lose kinetic energy, which is emitted as X-ray photons (Figure 4a).

- **Characteristic radiation**

Characteristic radiation is produced when an electron from an outer shell fills a vacancy in an inner shell of an atom's electron configuration (Figure 4.b).

When an incident electron collides with an atom in the target material, it can dislodge an inner-shell electron from its orbit. The atom becomes ionized, and an electron from a higher energy level falls into the vacancy created by the ejected electron. As the electron transitions, it releases energy in the form of an X-ray photon with a specific energy characteristic to the energy difference between the two electron orbits.



a) b)

Figure 1.4. X ray production by : a) Bremsstrahlung radiation b) characteristic radiation

The exact proportion of each type of radiation depends on factors such as the tube voltage (kVp) and the target material used in the anode. At lower tube voltages, characteristic X-rays may be more prominent, while at higher voltages, bremsstrahlung X-rays dominate.

X-ray tubes are made up of rotating or fixed anodes. The choice between rotating and fixed anodes depends on the specific application and the requirements of the X-ray system.

3.1.1 Fixed anodes X-ray tubes

In a fixed anode X-ray tube, the anode target is fixed in position and does not rotate during X-ray production (Figure 5). They are typically used in applications where lower power and less frequent imaging are sufficient.

Fixed anodes are commonly found in portable X-ray units, dental X-ray systems, and other settings where high-power imaging is not the primary requirement.

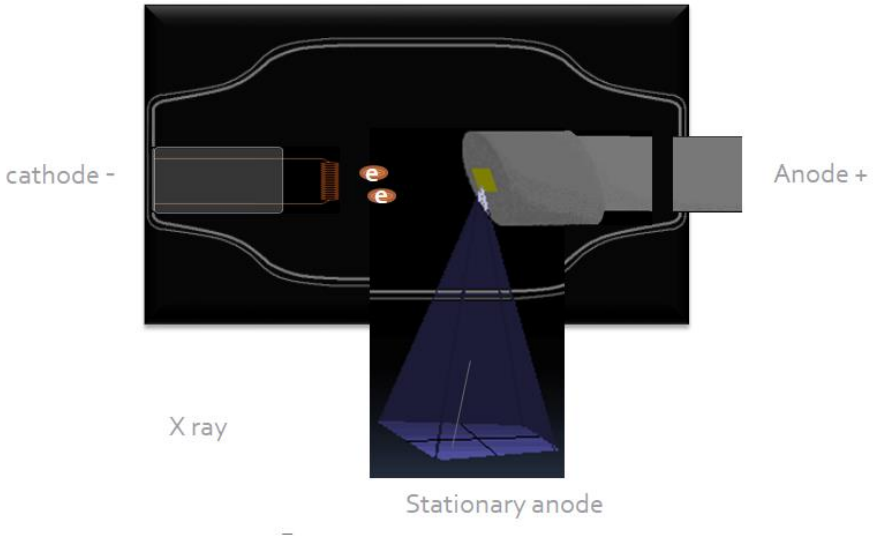


Figure 1.5. Fixed anode X-ray tube

3.1.2 Rotating anode X-ray tubes

In a rotating anode X-ray tube, the anode target is mounted on a rotating disk (Figure 6). The rotation serves two main purposes:

- Heat Dissipation: The rotation helps dissipate the heat generated during X-ray production. The high-speed rotation spreads the heat over a larger area, preventing overheating of the anode.
- Increased Power: The rotating anode allows for higher power X-ray production, enabling the generation of higher-energy X-ray beams and faster imaging.

Rotating anodes are particularly beneficial for CT scan, fluoroscopy, angiography, and other procedures that require rapid and continuous X-ray exposure.

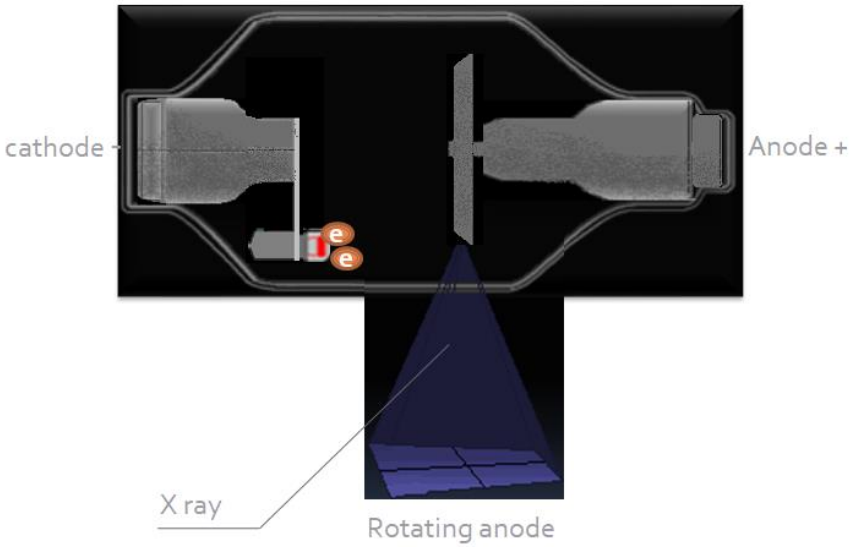


Figure 1.6. Rotating anode X-ray tube

3.2.Collimator

The collimator is a device attached to the X-ray tube that shapes and restricts the X-ray beam to the desired area of interest. It helps reduce unnecessary radiation exposure to surrounding tissues and improves image quality.

The Figure 7 explains how radiography works, from Xray source to the aquired image.

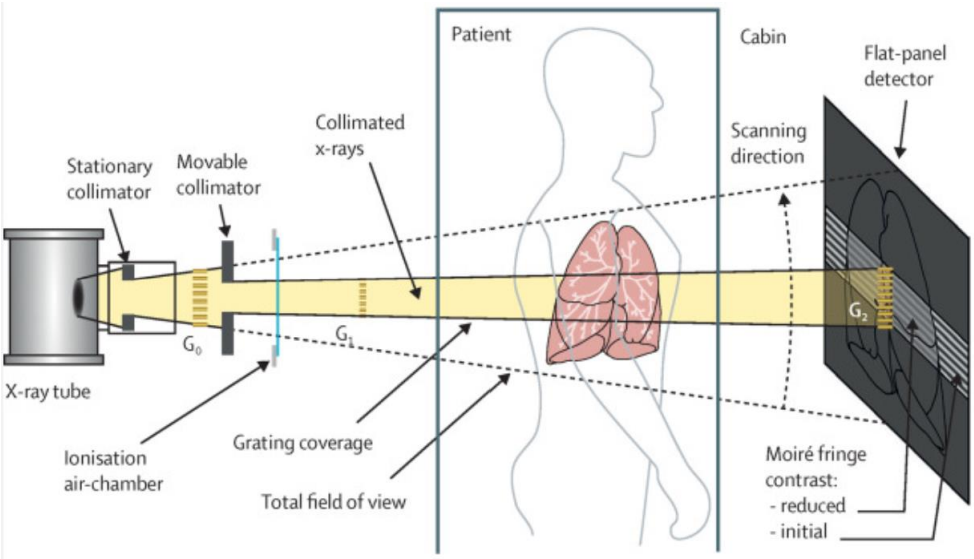


Figure 1.7. Radiography operating principle

When X-rays pass through the body, they interact with matter through several different processes.

3.3.Interaction of X-Rays with Matter

As X-rays pass through matter, they undergo different types of interactions that ultimately contribute to the creation of diagnostic images

Understanding these interactions enables us to understand how X-rays are absorbed, scattered and transmitted, leading to the formation of radiographic images. Here are the main interactions of X-rays with matter :

3.3.1. Photoelectric effect

In this interaction, an X-ray photon (green photon in Figure 8a) interacts with an inner-shell electron of an atom and is completely absorbed. The energy of the absorbed photon is transferred to the electron, causing it to be ejected from its shell. This process results in ionization, where the atom loses an electron and becomes ionized. The ejected electron is referred to as a photoelectron. The probability of the photoelectric effect is higher for higher atomic number (Z) materials and lower X-ray energies.

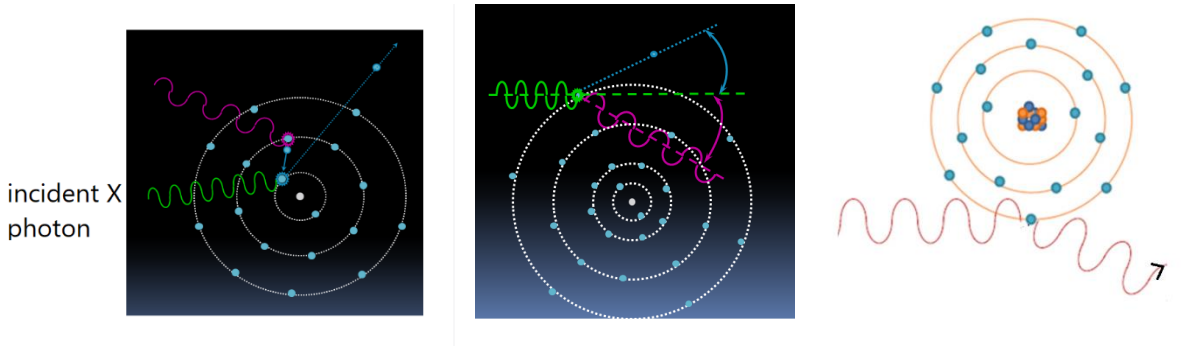
3.3.2. Compton scattering

During Compton scattering, an X-ray photon collides with an outer-shell electron of an atom (Figure 8b). The photon transfers some of its energy to the electron, causing the electron to be ejected from its shell (ionization). The scattered photon changes direction and exits the material at a reduced energy. Compton scattering is more likely to occur at higher X-ray energies and is responsible for the majority of scattered radiation in diagnostic radiography.

3.3.3. Coherent scattering (Thomson scatter)

Coherent scattering is a low-energy (less than 10 keV) interaction where the X-ray photon interacts with the atom's entire electron cloud. The photon's energy is momentarily absorbed by the atom, causing the electron cloud to vibrate. The photon is then re-emitted with the same energy but in a different direction (Figure 8c).

Coherent scattering has minimal impact in diagnostic radiography due to its low energy and limited contribution to image formation.



a) b) c)

Figure 1.8. Interactions of X-rays with matter : a) Photoelectric effect ; b) Compton scattering, c) Coherent scattering

In radiography, the dominant interaction between X-rays and matter depends on the energy of the X-ray photons and the type of tissue or material being imaged. The photoelectric effect is more pronounced at lower X-ray energies and with materials that have high atomic numbers (Z). It is the primary interaction for soft tissues and dense materials like bone. Photoelectric absorption contributes significantly to image contrast, especially in higher atomic number materials, as it removes X-ray photons from the beam.

Compton scattering becomes more significant at higher X-ray energies. It is the primary interaction for lower atomic number materials, such as soft tissues. Compton-scattered photons contribute to image fog and may result in reduced image contrast and clarity. Compton scattering also contributes to scattered radiation, which can affect image quality and radiation dose to patients and healthcare workers.

Subject contrast in radiography refers to the visual difference in X-ray attenuation between different tissues within the human body being imaged.

The attenuation of X-rays as they pass through matter in radiography can be described using the Beer-Lambert law, also known as the exponential attenuation law. This law relates the intensity of the X-ray beam before and after passing through a material to its thickness x and the material's linear attenuation coefficient μ (see Figure 9a) :

$$I=I_0\exp(-\mu x)\rightarrow \mu=1/x\ \ln(I/I_0) \quad (1)$$

Where : I is the intensity of the X-ray beam after passing through the material and I_0 is the initial intensity of the X-ray beam before passing through the material.

For heterogenous object (see Figure 9b) :

$$I=I_0\exp[-(\mu_1+\mu_2+\ldots+\mu_n)x] \quad (2)$$

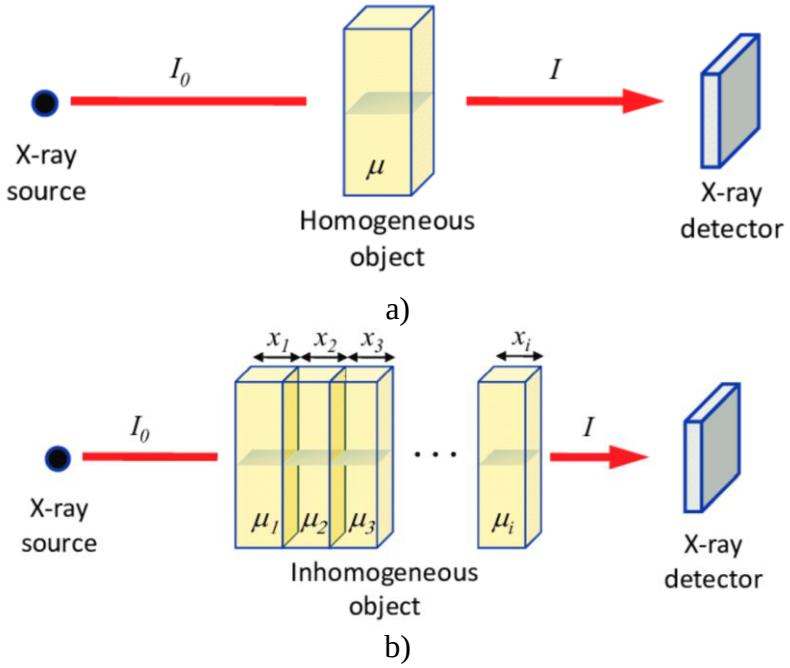


Figure 1.9. Schematic of X ray radiography : a) attenuation through homogenous medium and b) summation of attenuation through heterogenous medium

Subject contrast is produced by differential absorption (attenuation coefficient) of the various tissues imaged. All photon interactions (shown in previous section) within the patient represent a degree of absorption. Subject contrast can be thought of as the percentage of attenuations.

In radiography, the image produced is based on the differential attenuation of X-rays by different tissues in the body. Higher attenuation results in darker areas on the radiographic image.

X-ray images appear in various shades of white and grey. Because bones and metal objects are solid, less radiation passes through them, making them appear white on the radiograph. On the other hand, skin, muscle, blood and other fluids, and fat are grey because they allow most radiation to pass through. Areas where there is nothing to stop the beam of radiation, such as air, or even a fracture, appear black compared to surrounding tissue.

3.4.X-ray detector

The X-ray detector captures the transmitted X-rays that pass through the patient and converts them into electrical signals. They have evolved significantly over the years, transitioning from traditional film-based systems to Computed Radiography (CR) and digital technologies (DR) that offer improved image quality, efficiency, and dose reduction. Digital Radiography (DR) systems use flat-panel detectors, while Computed Radiography (CR) systems use phosphor plates that are later scanned to create digital images.

3.4.1 Film-Based Detectors

In traditional radiography, X-ray film was used to capture X-ray images. The film reacts to the X-rays, producing a latent image that needs to be developed before the image can be visualized.

The film based detectors consist of :

Radiographic film : X-ray film is a photosensitive material that reacts to X-ray exposure. When exposed to X-rays, the film emulsion undergoes a chemical change that creates a latent (invisible) image.

Intensifying screens : These screens are positioned on both sides of the X-ray film (see Figure 10) and contain fluorescent materials. When X-rays strike the intensifying screens, they convert X-ray energy into visible light, which exposes the X-ray film.

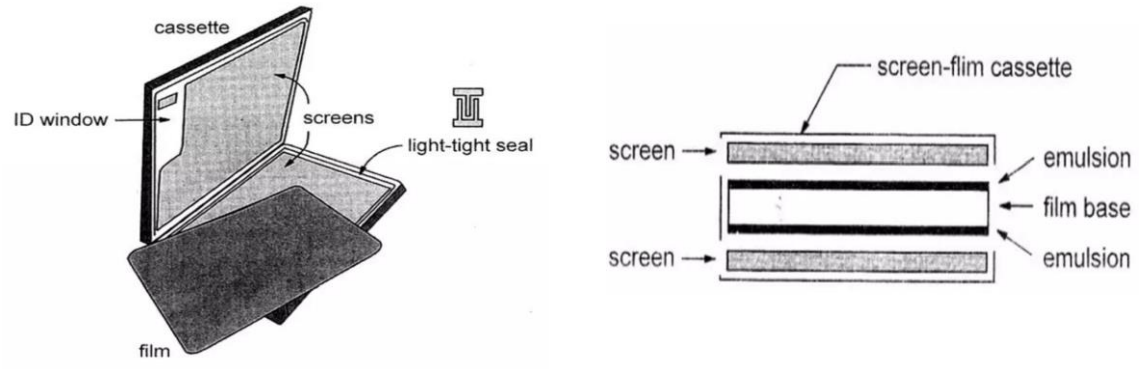


Figure 1.10. Film based detectors

The intensifying screens emit visible light photons proportional to the X-ray intensity they received. This light exposes the X-ray film's emulsion. The emulsion is the active component in which the image is formed and consists of many small silver halide crystals suspended in gelatin. The gelatin supports, separates, and protects the crystals. The typical emulsion is approximately 10 µm thick. The emitted light interacts with the X-ray film's emulsion, causing chemical changes in the film's silver halide crystals. These changes create a latent image that is not immediately visible.

The exposed X-ray film is processed in a darkroom using a series of chemical solutions. During development, the areas with greater X-ray exposure (darker areas on the image) undergo more extensive chemical reactions, leading to the formation of a visible image.

Recently, film-based systems have largely been replaced by digital technologies, namely Computed Radiography (CR) and Digital Radiography (DR).

3.4.2 Computed Radiography (CR)

Computed Radiography (CR) is a digital imaging technology that uses photostimulable phosphor (PSP) plates to capture X-ray images. CR systems offer a digital alternative to traditional film-based radiography, providing benefits such as rapid image acquisition, digital image processing, and efficient archiving.

The Figure 11 shows a schematic view of an X ray system equipped with a CR reader. Similar to traditional radiography, the patient is exposed to X-rays, which pass through the body and interact with the photostimulable phosphor plate.

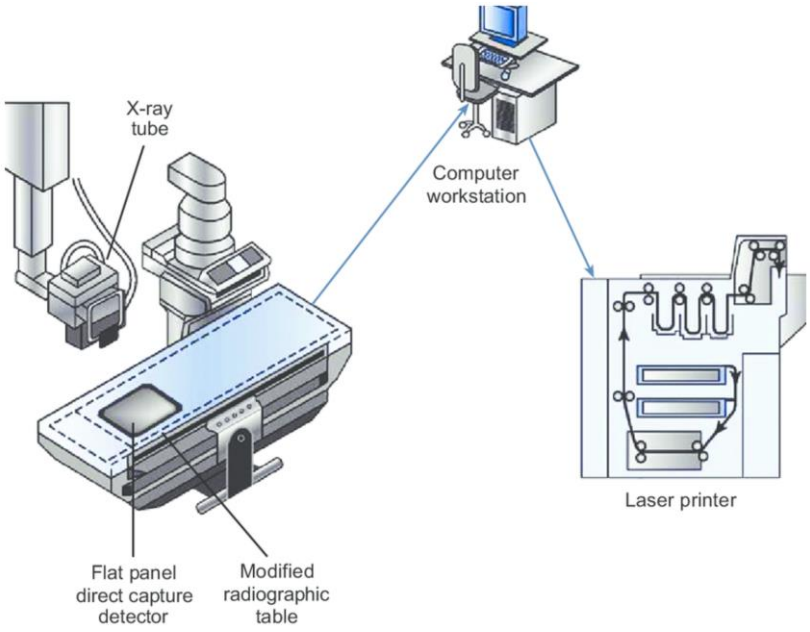


Figure 1.11. Schematic view of an X ray system equipped with a CR reader.

The X-ray photons are absorbed by the phosphor material in the plate, causing the electrons in the phosphor atoms to move to higher energy states. The absorbed energy is stored as trapped electrons in "electron traps" within the phosphor material. These trapped electrons create a latent image.

3.4.2.1 Photostimulable phosphor plate (PSP plate)

The PSP plate contains specialized materials that enable the storage of X-ray energy and its subsequent release as visible light during the scanning process. The main components of a photostimulable phosphor plate (Figure 12) include:

Phosphor layer : It consists of a layer of photostimulable phosphor crystals. These crystals have the ability to absorb X-ray energy and store it temporarily in the form of trapped electrons. Common materials used for the phosphor layer include barium fluorohalide compounds such as barium fluorobromide (BaFBr) or barium fluorochloride (BaFCl).

Protective layer : Above the phosphor layer is a protective layer that shields the phosphor crystals from physical damage and environmental factors such as dust, and scratches. It helps prevent scratches and contamination that could degrade image quality.

Reflective layer : The reflective layer enhances the efficiency of X-ray energy capture. It reflects emitted light during the readout process, preventing it from escaping the phosphor layer and allowing for efficient light collection.

Conductive layer : The conductive layer helps in discharging any residual electrical charges that might accumulate on the plate's surface during exposure.

Support layer: The support layer provides structural integrity to the PSP plate. It is typically made of a flexible material, such as polyester. The support layer holds the various layers together and ensures the plate's stability during handling and processing

Light Shielding Layer: The light shielding layer is positioned between the support layer and the backing layer. It is typically made of a material that absorbs or scatters light, preventing external light from reaching the sensitive phosphor layer.

Light exposure to the phosphor layer can cause unwanted noise or artifacts in the image, affecting its quality and accuracy. The light shielding layer helps maintain the integrity of the latent X-ray image stored in the phosphor layer until the plate is processed for readout.

Backing layer : The backing layer is usually a black material that reduces the amount of backscatter radiation during exposure.

It prevents X-rays from passing through the plate and interacting with the cassette or table beneath, reducing image artifacts

Antistatic layer :

Some PSP plates have an antistatic layer that helps reduce static electricity buildup and minimizes the attraction of dust or debris.

Barcode or identification layer : In some PSP plates, a barcode or identification layer is added for tracking purposes. This layer can store patient and image information, allowing for easy association between the image and patient data.

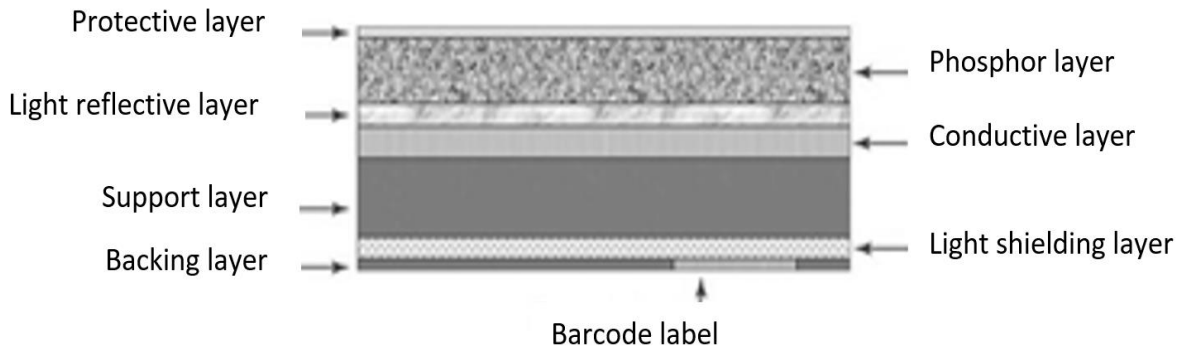


Figure 1.12. Photostimulable phosphor plate

After exposure, the phosphor plate is removed from the X-ray machine and placed in a CR scanner. The scanner uses a laser to scan the plate pixel by pixel, releasing the trapped electrons and converting the stored energy back into visible light.

The released light photons are collected by photomultiplier tubes or other light. The amount of emitted light is proportional to the amount of X-ray energy absorbed during exposure. The emitted light is converted into electrical signals and then into digital data through an analog-to-digital converter (ADC) for visualization on a computer monitor. Figure 13 explain image formation in CR.

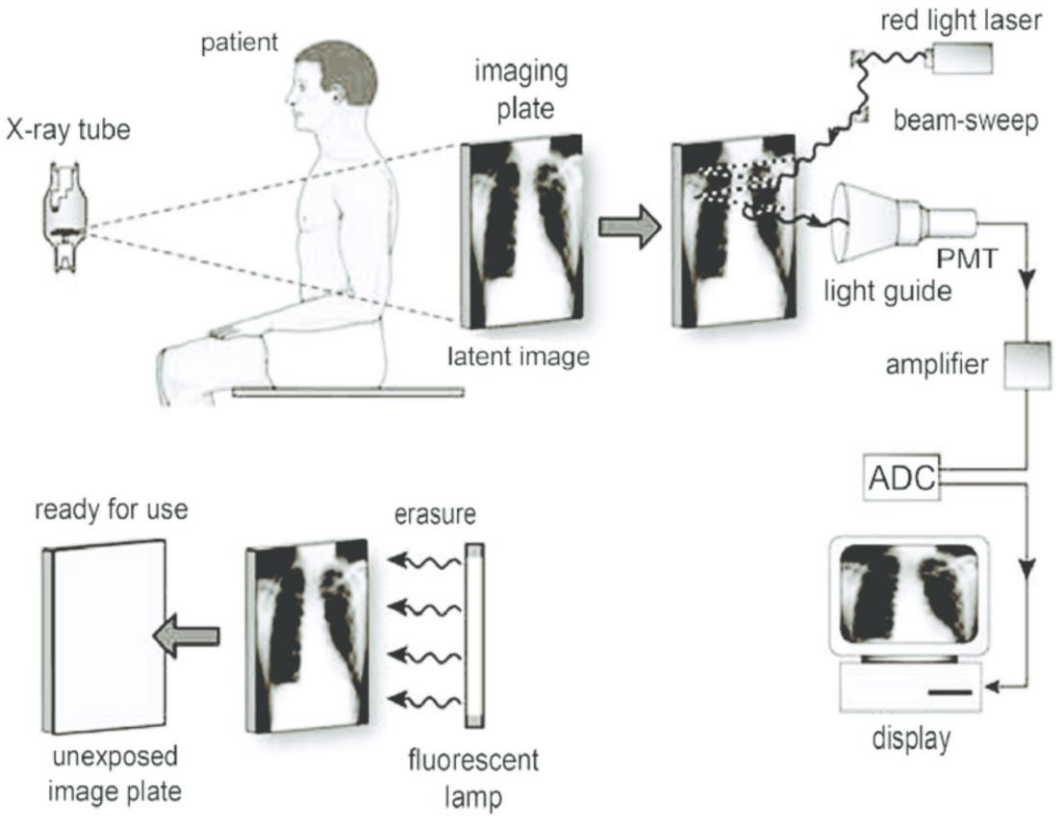


Figure 1.13. Image formation in CR [1]

Image processing techniques are applied to enhance contrast, correct artifacts, and optimize image quality.

In Computed Radiography (CR), erasure and the use of a fluorescent lamp are important steps in the process of reusing photostimulable phosphor plates for capturing new X-ray images.

3.4.2.2 Erasure

After a photostimulable phosphor plate has been exposed to X-rays and scanned to create a digital image, it must be "erased" before it can be used again for another exposure. Erasure refers to the process of removing the trapped electrons from the phosphor material, effectively resetting the plate for a new exposure. Erasing the plate ensures that any residual latent image from the previous exposure is removed and does not interfere with the quality of the new image.

Erasure is typically done using bright light or intense illumination. The phosphor plate is exposed to this light, causing the trapped electrons to be released and returning the phosphor material to a neutral state.

A fluorescent lamp is commonly used to provide the intense illumination needed for erasure in CR systems (Figure 13). The intense light emitted by the fluorescent lamp floods the phosphor plate, releasing the trapped electrons and erasing any residual latent image. This ensures that the plate is effectively "blank" and ready for a new exposure.

3.4.3 Digital radiography (DR)

Digital Radiography (DR) is a modern medical imaging technology that directly captures X-ray photons and converts them into digital signals, allowing for immediate image acquisition, processing, and display. The figure 14 shows a Digital Radiography equipment.

Figure 1.14. A Digital Radiography equipment.

DR detectors has revolutionized radiology by providing rapid and high-quality diagnostic imaging with enhanced efficiency, image manipulation, and dose reduction.

There are two main types of digital radiography detectors: Direct and Indirect.

3.4.3.1 Direct radiography detectors

Direct DR detectors directly convert X-ray photons into an electronic signal, which is then converted into a digital image. They offer high spatial resolution and dose efficiency. There are two common technologies used in direct DR detectors:

- **Amorphous selenium (a-se) detectors :** These detectors use a layer of amorphous selenium, which is a semiconductor material with high X-ray absorption efficiency. When X-rays hit the a-Se layer, they generate electron-hole pairs, and an electric field accelerates these charges to collect them at the detector's surface. The pattern of charges is then converted into a digital image.
- **Amorphous silicon (a-si) detectors :** These detectors use a layer of amorphous silicon, which acts as an X-ray sensor. When X-rays are absorbed by the a-Si layer, they release electron-hole pairs that are read out by thin-film transistors (TFTs) in an array (Figure 15 a). The TFTs convert the charge pattern into a digital image.

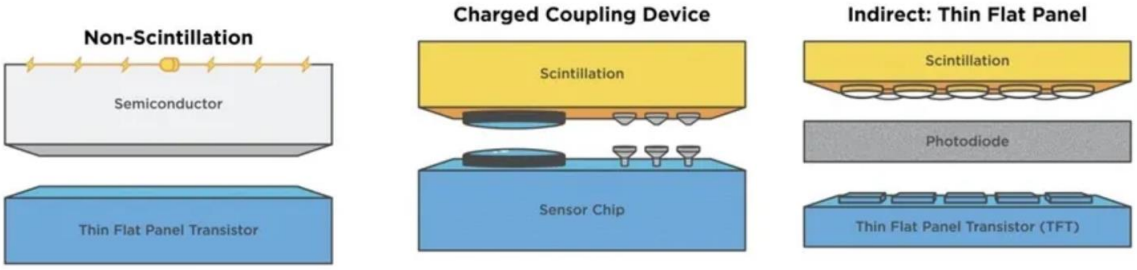
3.4.3.2 Indirect radiography detectors

Indirect DR detectors use scintillation materials to convert X-ray photons into visible light, which is then detected and converted into a digital image. There are two common types of indirect DR detectors:

- **Cesium Iodide (CsI) detectors :** These detectors use a layer of CsI scintillator, which emits light when exposed to X-rays. The emitted light is then detected by an array of

photodiodes or photomultiplier tubes, which convert the light signal into an electronic signal that is further processed into a digital image (Figure 15b).

- **Gadolinium Oxysulfide (Gd2O2S) detectors :** X-ray photons are first converted into visible light by a scintillator material, such as cesium iodide (CsI). The visible light is then detected by an array of photodiodes or other semiconductor devices, which convert it into electrical signals.



a) b)

Figure 1.15. a) Direct and b) indirect DR detectors

Digital radiography detectors have revolutionized the field of radiology by offering improved image quality, speed, and diagnostic capabilities compared to traditional film-based systems.

4. Conclusion

Radiography is a fundamental tool in modern medicine, allowing healthcare providers to visualize and diagnose a wide range of medical conditions with precision. Detectors used in medical imaging, including Digital Radiography (DR), Computed Radiography (CR), and traditional film-based radiography, play a crucial role in capturing and converting X-ray information into visible images.

Digital detectors, whether in DR or CR systems, offer numerous benefits, including immediate image acquisition, easy storage and retrieval, and the ability to enhance and share images digitally, making them the preferred choice for most modern radiographic applications.

Chapter 2
Computed Tomography (CT scanner)

Authored by

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1. Introduction

The principle of medical imaging is based on medical images in order to help the doctor determine and make the best decision for the correct diagnosis.

The X-ray technology provides a better view of the part of the human body on a flat surface, but they are limited to specific aspects, We know that the structure of the human body is a multilayered structure (one organ can be covered by another). This makes it very difficult to distinguish tissue within the same organ. Therefore, X-rays are never used to treat lesions or tumors but are currently helping to distinguish them because the images obtained are at a gray level (grayscale). Radiography gives us anatomical images of only the human body. It does not contain information about the physiology and biology of the living organ.

In order to solve the problem of the multilayer structure of the body, Godfrey Hounsfield in London presented the first scanner in 1971 with the installation of the first prototype «skull». The invention of X-ray tomography was one of the first breakthroughs in medical imaging and is the basis for the immense progress made in this field. This chapter presents an overview of the scanner, its components and generations, how the image can be reconstructed, and the limitations and risks of the scanner. We end this chapter with a conclusion

2. CT scanner definition

Computed tomography scanner also known as a CT scanner, is a medical imaging device that uses X-ray technology to generate detailed cross-sectional images of the body (see figure 1). It provides a valuable diagnostic tool for physicians to visualize internal structures and identify various conditions or abnormalities. the word “Tomos “ in Greek meaning slice or section and “graphia” that meaning write, there are many names for CT scan which is : CT(computed tomography) , Scanner, and CAT (computerized - assisted - tomography). Was the word “Scan “ defined as measurer and examiner.

Figure 2.1. CT scanner

According to the National Cancer Institute **(NCI),** a CT scanner is defined as follows: "A machine that uses X-rays to create detailed pictures of areas inside the body. The pictures are made by a computer linked to an X-ray machine. A dye may be injected into a vein or swallowed to help the organs or tissues show up more clearly."[2]

According to the Radiological Society of North America **(RSNA)**, it provides the following definition: a CT scanner is defined as follows: “Computed tomography (CT) is a specialized X-ray technology that produces cross-sectional images of the body. The CT scanner emits a series of narrow beams as it moves through an arc. This beam is rotated around the patient as the patient is moved through the scanner. The X-ray energy is absorbed differently by different tissues in the body, and the information about the energy absorption is used to generate images of internal structures.

These definitions should provide a clear understanding of what a CT scanner is and how it works.

3. CT scan components and generations

CT scanners work by rotating an X-ray source and detector around the patient's body. As the X-ray beam passes through the body, it is attenuated to varying degrees by different tissues. The detectors measure the intensity of the X-rays that emerge from the body, and this data is processed by a computer to construct detailed images. The following sections describe the different components and generations of CT scanners.

3.1. Components

CT scanners are available on the market as single-slice scanners, helical scanners, and multislice scanners. The gantry and patient table are the main components of CT scanners.

3.1.1 Gantry

The gantry is the ring-shaped part of the CT scanner, containing many of the components required for X-ray production and detection. It is a quality control system. Figure 2 shows the CT scanner Gantry components

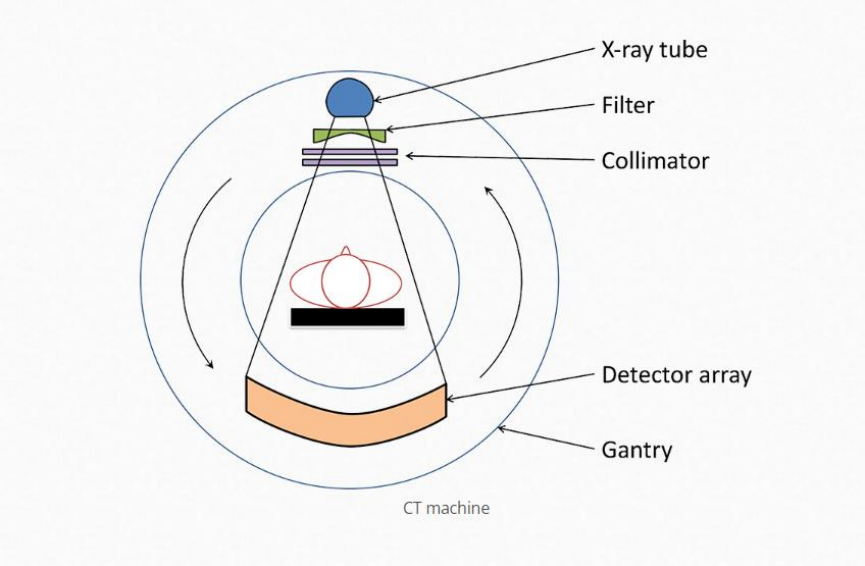
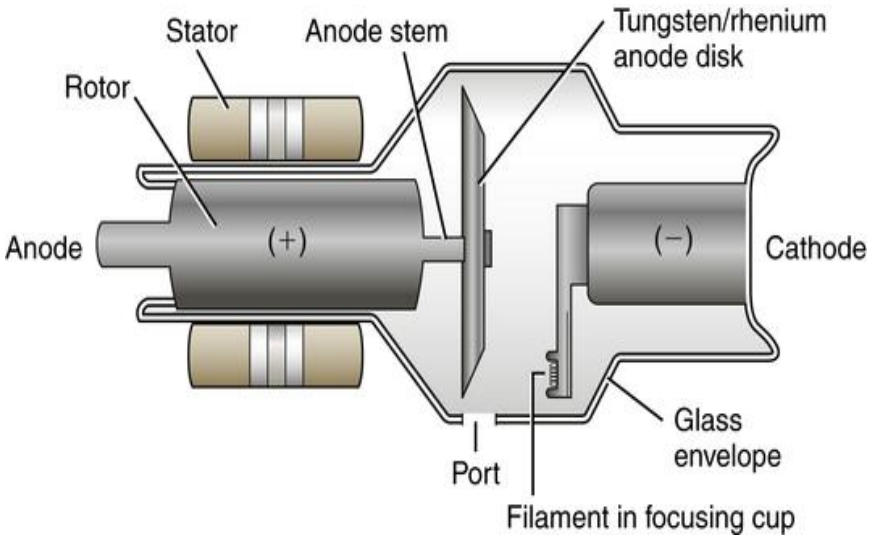


Figure 2.2. CT scanner Gantry components

3.1.1.1. X-ray tube

Although there are many different x-ray tube designs, they all have the same basic components. The X-ray tube design is used for general medical radiography. It purposes is an electronic vacuum tube that consists of an anode, a cathode, and an induction motor all enclosed in a glass or metal enclosure (envelop). Figure 3. provides a labeled illustration of this design. Recall that The anode is the end of the tube that is positive and the cathode is the end that is negative. The anode incorporates an anode target and an induction motor. The cathode is made up of the focusing cup and filament with its supporting wires [3],[4],[5].



a)

b)

Figure 2.3. a) Parts of the X-ray Tube, b) Photograph of X-ray Tube

The X-ray tube has the following characteristics:

- The X-ray tube uses intense pulse of X-ray and its performance must be stable.
- Tubes are operated for exposure time at high mA (e.g. 90 s, 120 kV, 200 mA).
- Heat exchangers are provided to cool oil, air, and to maintain gantry at low temperatures.
- Anode-cathode axis, parallel to the axis of rotation, to reduce heel effect.
- As focal spot increases the information spread over large number of detectors, and limit the resolution. High resolution CT uses small focal spot size.
- The anode is flat for easy heat dissipation and the angle is smaller than normal. While the cathode is angled and the focal spot position can be switched magnetically.

The X-ray tube must meet two essential qualities: High power (short exposure times) and fine focus (spatial resolution).

3.1.1.2 - High voltage generator

High-frequency generator with a capacity of 60 kW. The generator can give a tube current of about 800 mA @125 kV with a pulse duration of 2–4 ms. Three-phase generator produces high voltages (generally between 120 to 140 kV) and supplies it to the X-ray tube.

3.1.1.3 - Filtration and Collimation

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 the patient's dose. Filters made of aluminum or Teflon.

The collimator is located between the filter and the patient. CT scanner uses one or two collimators, which reduces patient dose and improves image contrast, by limiting scatter radiation, The Single slice scanner uses 2 collimators for pre and post-patient collimation. whereas the Multislice scanner uses only a single collimator as pre-patient collimation [3],[4],[5].

Note:

The pre-patient collimator was placed between the filter and the patient, while it limited the area of the patient and determined his dose.

The post-patient collimator was Placed after the patient and above the detectors. It restricts the X-ray beam seen by the detector array, reduces scatter and improves contrast.

3.1.1.4 – Detector

Detector converts X-rays to an electrical signal, there are two types of detectors:

3.1.1.4.1 gas-filled detector (Xenon gas detector)

The Xenon gas filled ionization chamber detector has high atomic number (54) and its K-shell binding energy is 35 keV. When X-ray falls on the detector, ionization takes place and electric charges are produced. These charges constitute an electric signal that is amplified and digitized. The digitized electronic signal is proportional to the incident X-ray intensity.

3.1.1.4.2 Solid state detector

Consists of a scintillation phosphor coupled to a photodiode. When X-ray falls on the detector light is produced, which is detected by the photodiode. The photodiode gives the electric signal that is digitized. The digitized electronic signal is proportional to the incident X-ray intensity .

On either side of the gantry opening, we'll find control panels used by X-ray technologists to control alignment lights, gantry tilt and table movement.

3.1.1.4. Control Console

Controls for selection of technique factors: movement of gantry, patient couch, image reconstruction and transfer, selection of kVp, mA, etc.
The control contains 2 monitors: one to annotate the patient data (patient name, age, etc.), and identify the images (technique and couch position), and the other one for the operator to view the image. It can manipulate the image, optimize diagnostic information, and adjust contrast and brightness.

3.1.1.5. Patient Couch

The couch supports the patient. It is motorized, allowing the patient to be positioned smoothly and unaffected by the patient's weight. It moves longitudinally through the gantry opening.

3.1.2. Computer

The computer is used to help of a microprocessor/array processor and has primary memory and the software includes planner and volumetric quantitative analysis and reconstruction of images in coronal, sagittal, and oblique planes.

3.2. CT scan Generations

The configuration of the x-ray tube to the detectors determines scanner generations (see Figure 4) [6], [7], [8]:

- **The first generation**: the CT scanner is a rotate/translate pencil beam system. It had two X-ray detectors and used parallel ray geometry (Figure 4.a). It took about 4.5 minutes per scan and it is rotated between translations to acquire 180 projections at 1° intervals.
  The advantage of the system is the efficient scatter reduction.
  The disadvantage includes; the amount of time it took to acquire the images and to reconstruct the images using the computer.

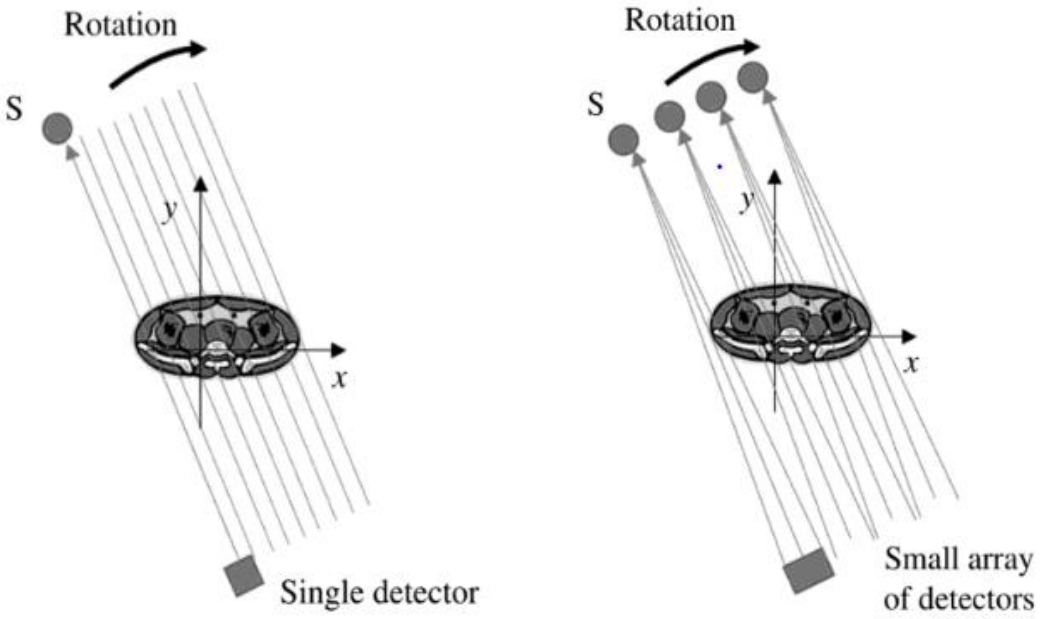
- **Second generation:** the second generation CT scanner is also a rotate/translate system with a narrow beam geometry of 10°. A linear array of 30 detectors was used to acquire more data and improve image quality. These scanners provided larger rotational increments and faster scans. The shortest scan time was 18 seconds per slice. A narrow fan beam allows more scattered radiation to be detected (figure 4.b).

- **Third generation:** The third generation scanner is a rotate/rotate system with wide beam geometry. The number of detectors has increased substantially (> 800 detectors) and the angle of the fan beam is increased to cover the entire patient. It eliminated the need for

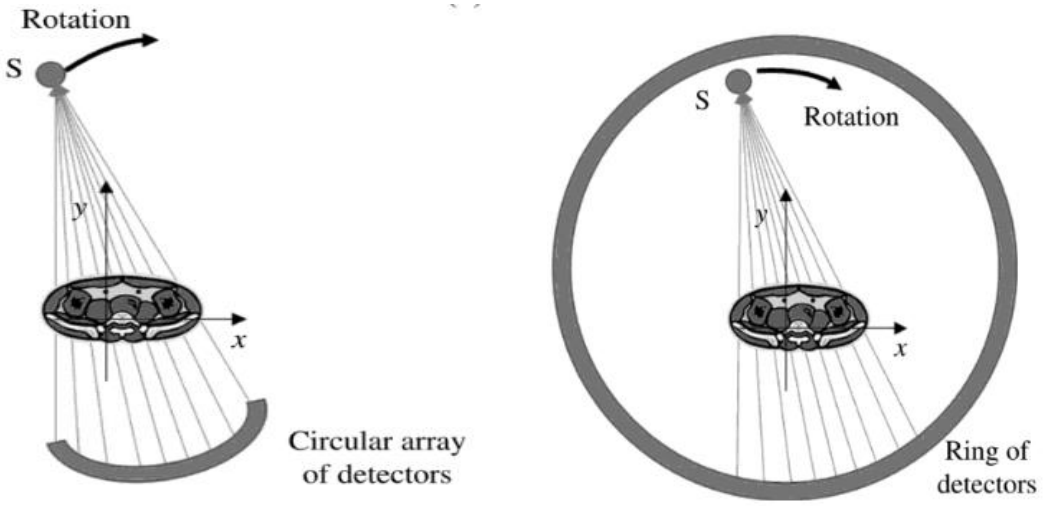
translational motion (figure4.c). The X-ray tube and detector array are mechanically joined and rotated together. Newer systems have scan times of the order of < 0.5 seconds.

- **Fourth generation:** The fourth generation scanners (figure 4.d) are designed to overcome the problem of ring artifacts. It has a stationary ring of about 4,800 detectors, and the X-ray tube has to move inside this detector. Since it is rotated continuously, a very fast scan time is possible. It has inter-scan delay times since the X-ray tube had to return to its starting position (home).

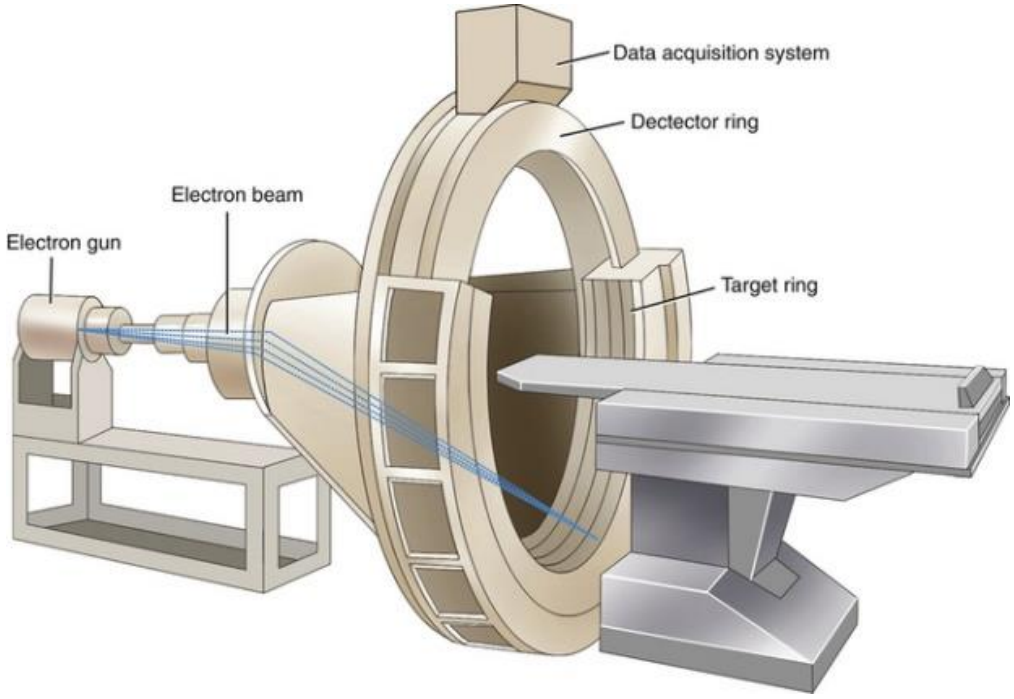
- **Fifth generation:** The fifth generation scanner (figure 4.e) is a stationary/stationary system, developed specifically for cardiac tomography imaging. No conventional X-ray tube is used, instead, a large arc of tungsten (210°) encircles the patient and lies directly opposite to the detector ring. It uses an electron gun that deflects and focuses a fast-moving electron beam along a tungsten target ring in the gantry. Since the detector is also in the form of a ring, it permits simultaneous acquisition of multiple image sections. The images are obtained in 50 ms times and can produce fast frame rate CT movies of the beating heart with minimum motion artifacts.
  The advantage is the speed of data acquisition. The whole heart can be acquired in 0.2 s.



(a) (b)



(c) (d)



(e)

Figure 2.4. CT scan generation : (a) first generation,(b)second generation,(c)third generation, (d)four generation, and (e) fifth generation.

4. Principle of the CT scanner and image reconstruction

When biological tissues are exposed to X-rays, they are attenuated according to an exponential function taking into account photoelectric absorption and diffusion by Compton effect. If I_0 is the incident flow of X-rays in a heterogeneous medium of attenuation coefficient (x), and I the flow out of the tissues, we have the following relationship:

$$I=I_0 \exp-(x) \qquad (1)$$

The CT scanner is based on the measurement of the different absorption coefficients of tissues crossed by an X-ray beam. Each tissue has its own absorption coefficient which depends on the density of the tissue and the energy of the beam passing through it. If we associate a gray level scale with this coefficient, we can obtain using retro projection algorithms a scanner image, corresponding to the image of a cross-section of the studied body [3].

The CT image is made up of pixels (picture elements). The average X-ray attenuation of a box-like (small volume) element (voxel) that extends through the thickness of the tissue section is measured by each pixel in the image. Additionally, in a real CT image, all tissues within a single pixel would have the same gray shade. Each organ has a unique attenuation value on the Hounsfield scale. For instance, water registers as 0 HU (Hounsfield Units), bones are typically around 1000 HU and appear white, while air is approximately -1000 HU and appears black (figure 5).

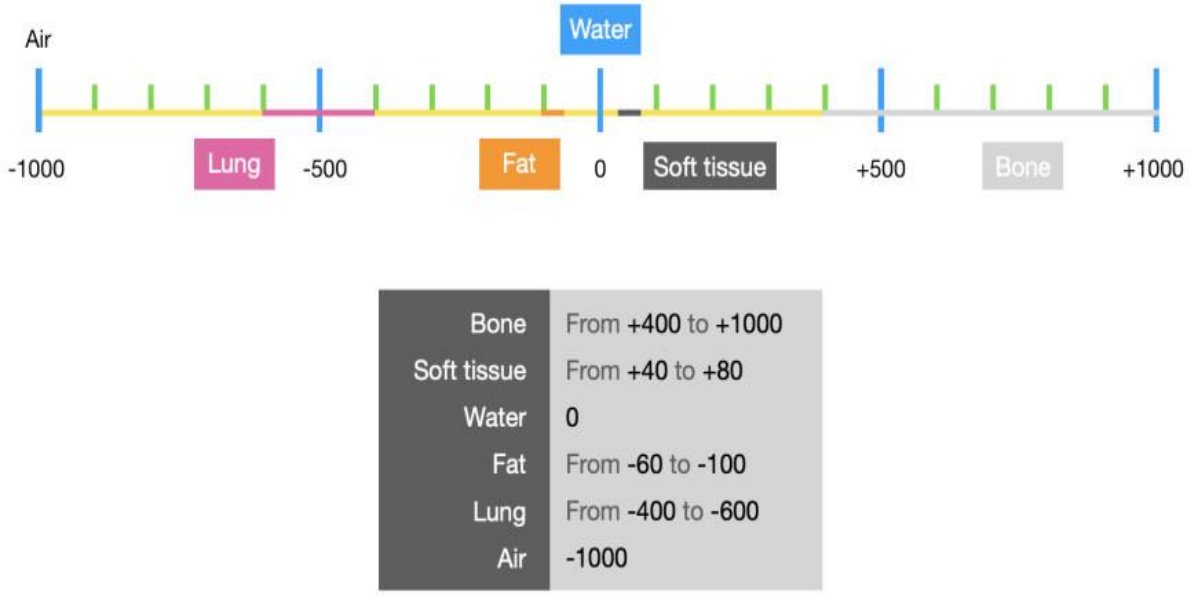


Figure 2.5. Hounsfield unit Scale

The image contains pixels (of length x) and voxels (volume). The X-ray beam which is transmitted through a voxel, is given by the relation:

$$I=I_0 \exp(-\mu x) \qquad (2)$$

Where I_0 is the number of initial X-ray photons, I is the number of transmitted photons, exp is the base of the natural logarithm (2.718) and μ is the linear attenuation coefficient of the voxel. The values of I0, I, and x can be measured and the only unknown is μ which represents the tissue.

CT scan image reconstruction involves creating detailed cross-sectional images of the body using data acquired from a CT (computed tomography) scan.CT scans can be used to create a three-dimensional representation by using a series of X-ray images taken from different angles

around the body [3],[6]. The CT scan data use important steps to produce an image, the image reconstruction follows the steps below :

1- Data Acquisition: During a CT scan, the X-ray machine rotates around the body, taking multiple X-ray images from various angles. These images are captured by a detector array.

2-Data Conversion: The X-ray images are converted into digital data, which represents the intensity of X-ray absorption at each point.

3- Preprocessing: Before reconstruction, certain corrections and enhancements are applied to the data to account for factors like X-ray beam divergence, patient motion, and beam-hardening artifacts.

4- Projection Data: The digital data is organized into a set of projection data or sinograms. Each sinogram represents the intensity measurements along a particular X-ray path through the body.

5- Filtered Back Projection: The most common method for CT image reconstruction is filtered back projection. In this process, each sinogram is mathematically manipulated to reverse the X-ray projection and reconstruct the attenuation values along each X-ray path.

6- Voxel Reconstruction: The reconstructed attenuation values are assigned to individual pixels, known as voxels, in a three-dimensional grid corresponding to the scanned region. Voxels contain information about the X-ray absorption properties of the tissues at specific locations.

7- Image Formation: Finally, the voxel data is processed to generate cross-sectional images that can be visualized on a computer monitor or printed on film. Advanced algorithms and techniques may be employed to improve image quality, reduce noise, and enhance specific structures of interest.

It's important to note that while this explanation provides a general overview of CT scan image reconstruction, the actual process involves complex mathematical algorithms and techniques to produce accurate and detailed images. Figure 6 illustrates the production of a CT scan image through the collection of information in a sinogram before generating the final image using the steps described earlier.

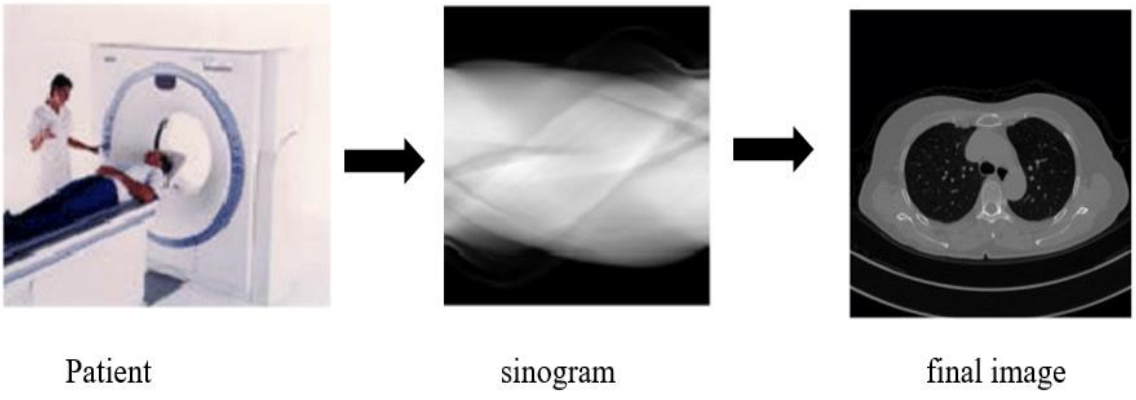


Figure 2.6. CT scan image production steps

5. Limitations and risks of CT scan

- The CT scan cannot show the functional parts of a human body.

- The CT scan is an effective and proven examination, but it does entail certain radiological risks, such as X-rays and the administration of contrast products, particularly for pregnant women.

-Risk of hematoma at injection site if there is an allergic from it.

6. Conclusion

Rapid advancements in computed tomography (CT) image quality reflect the ongoing pursuit of efficient and accurate image reconstruction methods while minimizing radiation exposure. Over the past decade, CT technology has seen significant improvements, and research in this field continues to evolve.

The core challenge in CT image reconstruction lies in computing the attenuation coefficients of various X-ray absorption paths (ray sums) obtained as a set of projection data. Minimizing X-ray exposure is crucial for the safety of individuals undergoing CT scans. While CT scans are cost-effective and fast, they may not provide the same level of anatomical detail as some other medical modalities.

The quest for optimal CT image quality remains at the forefront of medical imaging research, with the goal of striking a balance between precision, safety, and efficiency.

Chapter 3
Scintigraphy scan

Authored by

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1. Introduction

The domain of nuclear medicine for diagnostic imaging have developed over the last century, with the discovery of X rays by Rontgen and Becquerel who discover the natural radioactivity. Scintigraphy, also known as nuclear scintigraphy or nuclear medicine imaging, is a medical imaging technique that uses small amounts of radioactive material, called radiotracers or radiopharmaceuticals, to create images of various organs and tissues in the body. It is a non-invasive diagnostic tool that provides valuable information about the structure and function of internal organs [9]. In this chapter, we explore the fundamental concepts of nuclear medicine imaging, aiming to enhance understanding of this technology. We begin with a concise definition and discuss patient preparation before undergoing this examination. Subsequently, we delve into the operation of scintigraphy, explaining the principles of radioactivity and the essential components employed in gamma cameras to generate images. Additionally, we distinguish between scintigraphy, SPECT, and PET imaging techniques. Finally, we conclude the chapter by examining the applications, benefits, and associated risks of scintigraphy.

2. Scintigraphy definition

Scintigraphy is based on the administration of a small amount of a radioactive substance (radiotracer or radiopharmaceutical) into the human body, which travels to the target organ or tissue. This radiopharmaceutical emits gamma rays that are detected by a special camera called a gamma camera. The gamma camera is placed over the body to create an image of the distribution of the radiopharmaceutical within the body.

The scintigraphy images are used in a wide variety of medical conditions, with the most common applications being bone scintigraphy in oncology and cardiac scintigraphy in the study of myocardial ischemia. Scintigraphy is a minimally invasive technique that involves intravenous administration. It is considered safe and can be repeated as many times as necessary. Figure 1 provides an example of a Scintigraphy device.

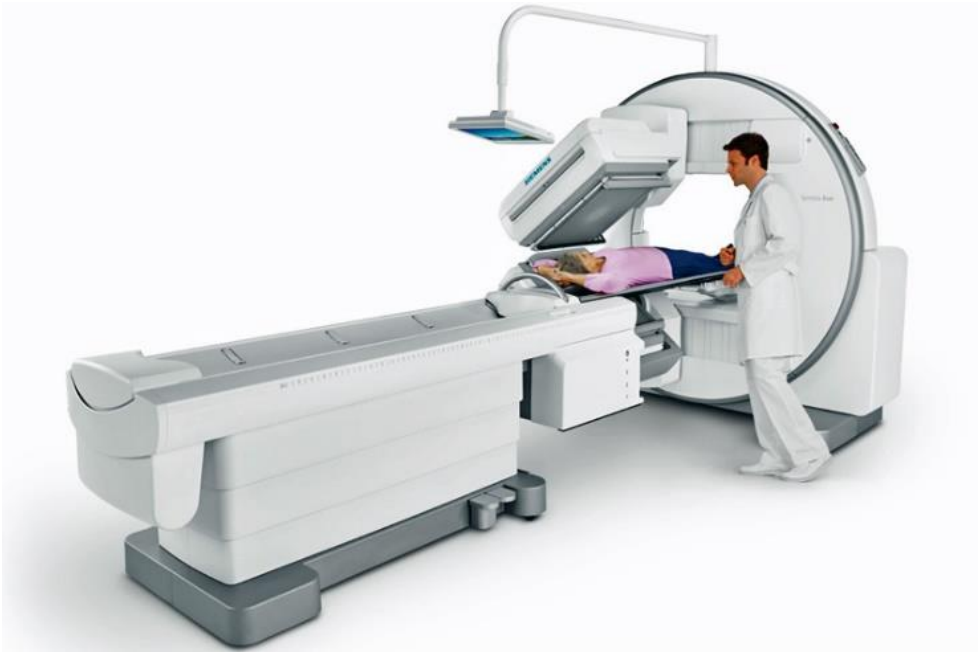


Figure 3.1. Scintigraphy device

3. Preparation of scintigraphy

Preparation for the scan depends on the specific type of examination and the medical condition being evaluated. The general guidelines for scintigraphy preparation may include:

1. Inform the health professional about any medical condition (allergy or medication being taken).
2. Wear comfortable clothing, preferably without metal zips or buttons.
3. Drink plenty of fluids before and after the test, to eliminate the radiopharmaceutical from body quickly.
4. Don't have to fast, except for thyroid scintigraphy, gastric emptying, and scanning with I-131.
5. Avoid caffeine, nicotine and strenuous exercise for at least 24 hours before the exam.
6. Some types of nuclear medicine imaging may not be recommended during pregnancy or lactation.
7. Depending on the type of test, it may be necessary to remain in the imaging center for several hours after injection, allowing for the absorption of the radiopharmaceutical and the acquisition of images at the optimal time.

4. Scintigraphy operation

Scintigraphy, an imaging technique that plays a crucial role in visualizing the internal functions and structures of the human body. By using small quantities of radioactive substances known as radiotracers or radiopharmaceuticals, scintigraphy provides healthcare professionals with a powerful tool for assessing organ function, detecting abnormalities, and assisting in the diagnosis and monitoring of a wide range of medical conditions. This imaging procedure involves the administration of radiotracers, their natural distribution in the body and the detection of gamma rays emitted using a gamma camera, which are then transformed into detailed images thanks to advanced computer processing. Figure 2 illustrates the basic principle of image formation in sintigraphy.

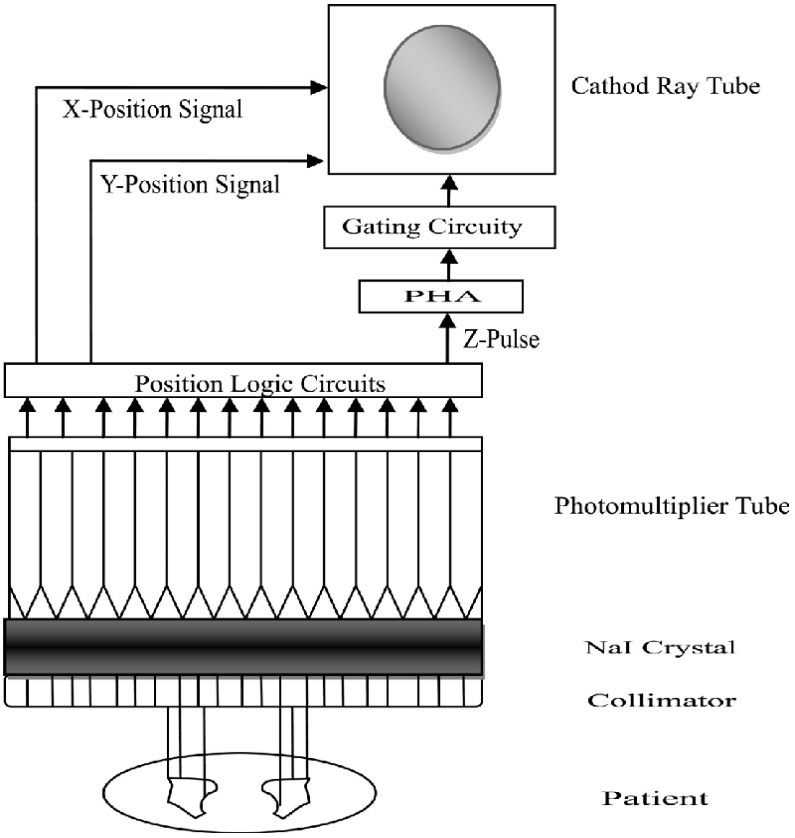


Figure 3.2. Basic principle of image formation in sintigraphy

4.1. Radioactivity

Radioactivity discovered by Henry Becquerel, is defined as a process in which nuclei of certain elements spontaneously disintegrate : transformation into another element by ejection of α, β or γ particle.

- Emission of α particle means loss of two protons and two neutrons.
- Emission of β particle means loss of an electron.
- Emission of a γ ray means no change in charge and mass, but only energy changes.

During disintegration, atoms of new elements called daughter are formed with different physical and chemical properties. Radioactivity remains unaffected due to the physical and chemical changes of the material, represented by:

$$N = N_0 e^{-\gamma t} \quad (1)$$

Where N is the number of radioactive nuclei given instant of time and N_0 is the number of radioactive nuclei at time t=0.

Radiotracer are used in small amounts to evaluate the function of different organs, it can be designed to seek out only desired tissues or organs, radiotracers help doctors to follow physiological functions and metabolic activity of the human body and learn more about organs. The table 1 presents the most commonly used radiotracers in scintigraphy.

Table 3.1. Different radiotracer used in scintigraphy

Radionuclide	***Half-life***	***Imaging decay***	***E_{max} (Positron or photon) (kev)***	***Production***
^{11}C	20.4 min	β^+	960	$^{14}N\ (p, \gamma)\ ^{11}C$
^{13}N	9.96 min	β^+	1190	$^{16}O\ (p, \gamma)\ ^{13}N$
^{15}O	2.07 min	β^+	1720	$^{14}N\ (d, n)\ ^{15}O$
^{18}F	1.8 h	β^+	640	$^{18}O\ (p, n)\ ^{18}F$
^{64}Cu	12.7 h	β^+	660	$^{64}Ni\ (p, 2p)\ ^{64}Cu$
^{67}Ga	78.3 h	γ	93, 184, 300, 393	$^{68}Zn\ (n, p)\ ^{67}Ga$
^{86}Y	14.7 h	β^+	1200	$^{86}Sr\ (p, n)\ ^{86}Y$
^{99m}Tc	6.0 h	γ	140	$^{99}Mo/^{99m}Tc$ generator
^{111}In	67.2 h	γ	171, 245	$^{111}Cd\ (p, n)\ ^{111}In$
^{123}I	13.2 h	γ	159	$^{121}Sn\ (\gamma, 2n)\ ^{123}I$
^{124}I	100.2 h	β^+	2100	$^{214}Te\ (p, n)\ ^{124}I$
^{125}I	60.1 days	γ	35.5	$^{124}Xe\ (n, \gamma)^{125}Xe/^{125}I$

Gamma radiation is similar to X rays in that it easily penetrates living materials. Gamma rays, also referred to as photons, travel at the speed of light. Gamma rays have sufficient energy to ionize matter and can cause cellular damage [10].
The gamma rays are often produced as a result of annihilation events.

- **Annihilation phenomenon**

Annihilation is a fundamental process in particle physics involving the interaction of particles and antiparticles, resulting in their mutual destruction and the conversion of their mass into energy. The annihilation of the positron and electron yields two gamma photons each of 511Kev. Figure 3 shows the production of gamma rays.

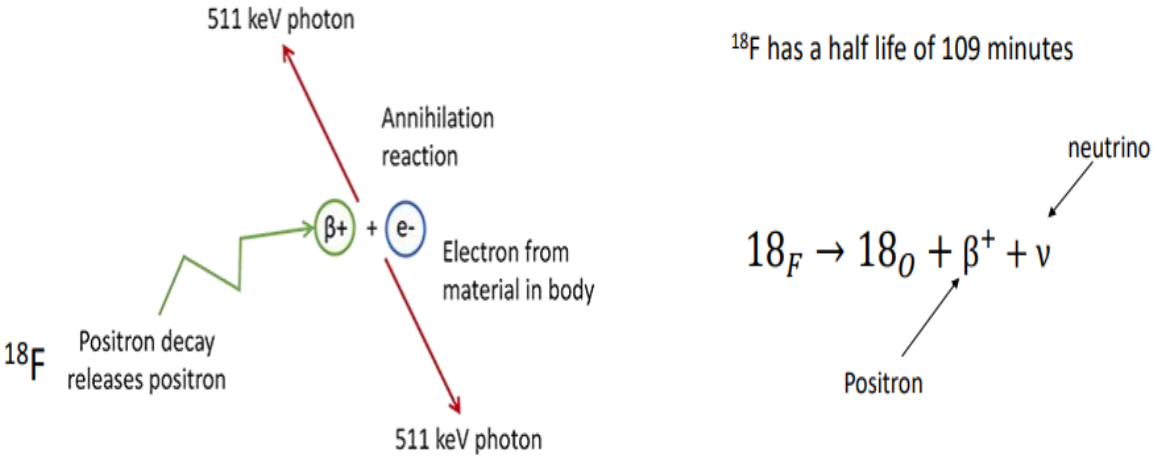


Figure 3.3. Production of gamma rays

The operation of a gamma camera involves several key steps, from the detection of gamma-ray emissions to the generation of images that provide valuable information about the distribution of radiotracers or radiopharmaceuticals in the body.

4.2. Gamma camera and its components

A gamma camera (γ-camera), alternatively known as a scintillation camera or Anger camera, is a device designed for capturing images of radioisotopes The applications of scintigraphy include early drug development and nuclear medical imaging to view and analyze images of the human body or the distribution of medically injected, inhaled, or ingested radionuclides emitting gamma rays [10].
The major components of a gamma camera are a collimator, a large-area NaI(Ti) scintillation crystal, a light guide, and an array of photomultipliers tubes (see figure 2) [11].

4.2.1 Collimator

When radiation is emitted from the patient, it can exit at any angle and strike the detector in a position that may not correspond to its point of origin. To address this issue, a collimator is employed, allowing only gamma photons traveling in parallel with the collimator to be accepted. Those traveling at an angle will intersect with the septum, typically made of lead, become absorbed, and consequently, not contribute to the image. Figure 4 illustrates this procedure.

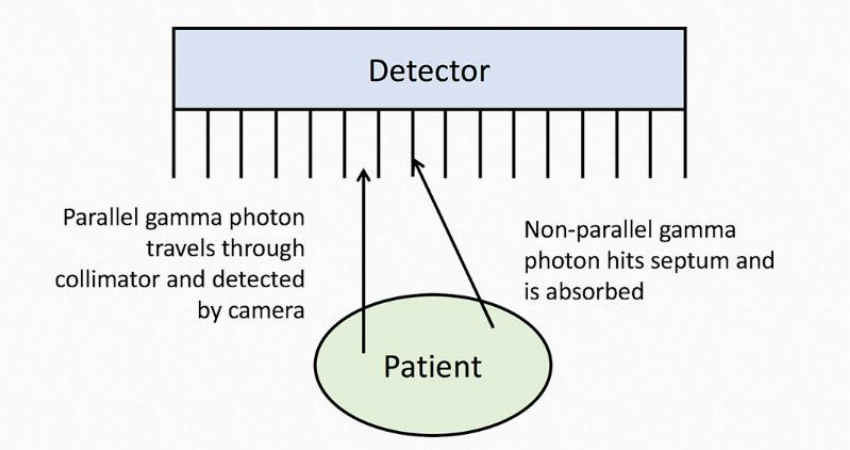


Figure 3.4. collimator

In the upcoming section, we will outline key characteristics of a collimator employed in gamma cameras.

-Features of the collimator:

The main characteristics of a collimator used in gamma cameras are as follows (see figure 5):

- **Parallel hole:** all holes are parallel to each other, most common design are Low Energy All Purpose (LEAP), Low Energy High-Resolution (LEHR) and Medium and High Energy Collimators.
- **Converging hole:** the holes are not parallel but focused toward the organ.

- **Diverging hole:** used to enlarge the field of view, example, portable cameras with small crystal.
- **Pinhole:** these collimators have a single hole with interchangeable inserts, used for small organ like the thyroid. There are designed for low energy isotopes [12].

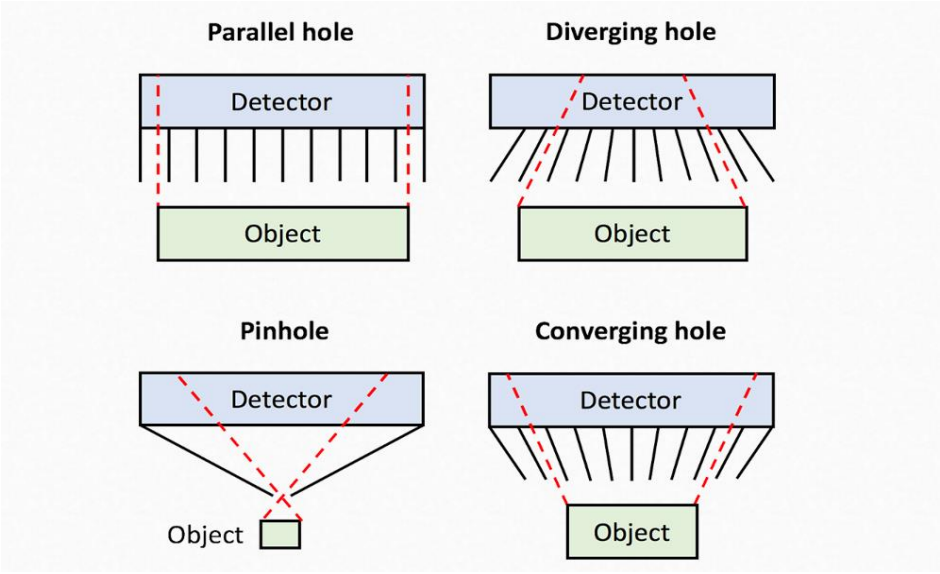


Figure 3.5. Types of collimators

4.2.2 Scintillation crystal

The scintillation crystal is a critical component of gamma cameras. Its primary function is to convert incoming gamma rays into light. When a gamma photon interacts, it releases light photons (mixture of visible and UV light) as shown in figure 6.
The scintillation crystal is composed of a single crystal of sodium iodide with a small amount of thallium (NaI (Tl)) who improves the light output.

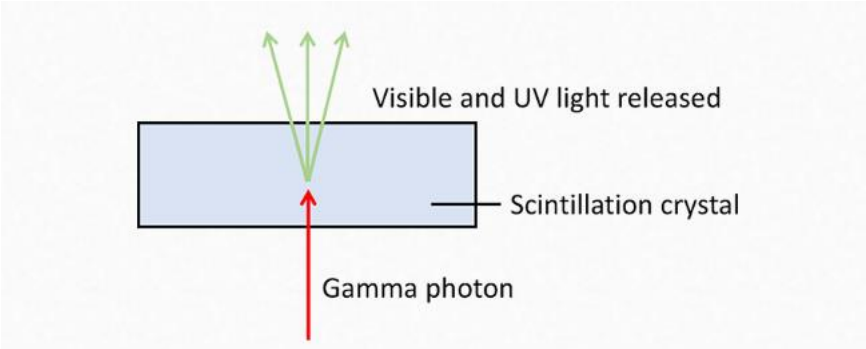


Figure 3.6. crystal

4.2.3 Photomultiplier tubes (PMT)

There is approximately 30-100 PMTs sit behind the scintillation crystal, the role of these photomultiplier is to multiply the small amount of light detected from the scintillation crystal to a large signal [11].

Figure 7 explains how the PMT transforms the light into electrical signal, taking the following steps :

- The light photons hit a photocathode at the entrance to the PMT.
- The photocathode releases electrons in proportion to the amount of light that hits it.
- The electrons are attracted to the electrodes (dynodes) which have an increasingly positive charge along the PMT. This accelerates the electrons. As they accelerate, they gain kinetic energy resulting in multiple electrons being released from the dynode for each electron that hits it. This serves to multiplied the original signal.
- The total number of electrons that hit the final anode produces the current, which forms the signal received by the pre-amplifier.

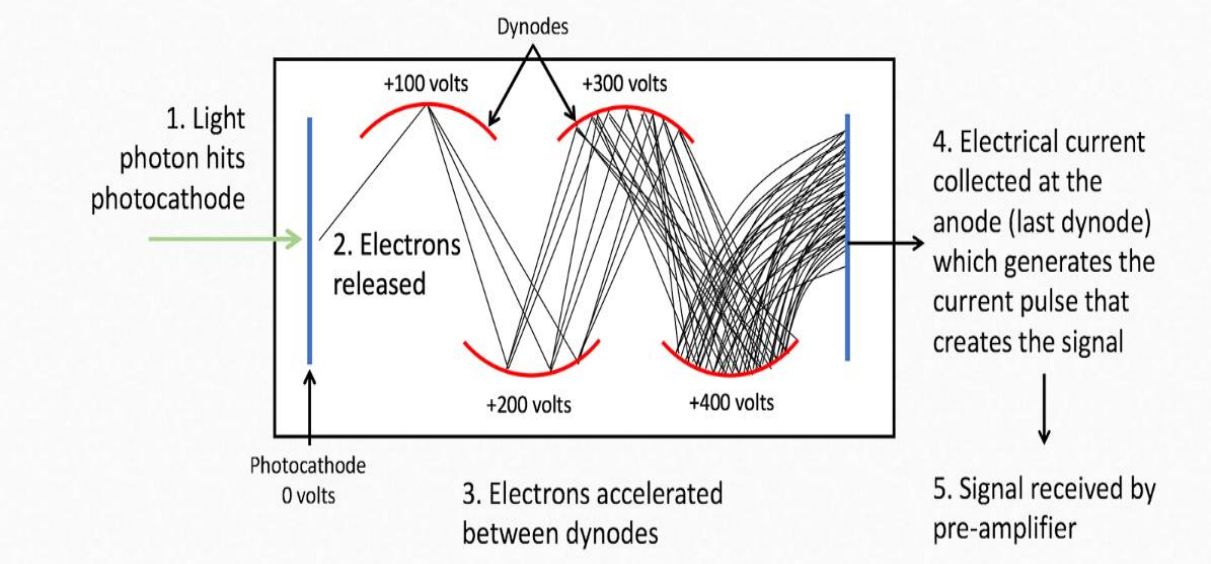


Figure 3.7. Photomultiplier tube (PMT)

5. Image formation

On the scintillation crystal, each PMT corresponds to specific coordinates mapped out onto a matrix within the image. The resulting digital image is displayed on a monitor, where each pixel corresponds to a memory location in the matrix. The brightness or color scale in the image corresponds to the count number in that location.

Display can be manipulated and optimized by:

- Smoothing to reduce noise.
- Windowing to increase contrast.
- Interpolation increases the display matrix relative to the acquisition matrix which spreads the counts and makes the pixels less apparent.
- Adding and subtracting images to extract quantified information.

Figure 8 shows an example of an image acquired through scintigraphy.

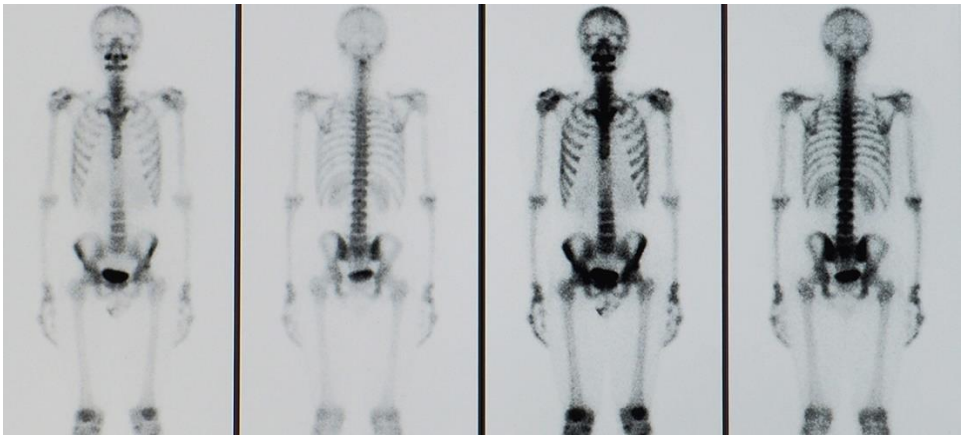


Figure 3.8. Formation of an image in scintigraphy

Scintigraphy, Single Photon Emission Computed Tomography (SPECT), and Positron Emission Tomography (PET) are all medical imaging techniques used in nuclear medicine. They share similarities but also have distinct differences.

6. Scintigraphy, SPECT and PET

Scintigraphy uses gamma cameras to detect gamma rays emitted by radiotracers (radiopharmaceuticals) within the body. Various radiotracers can be used for different organ-specific imaging. Scintigraphy typically has lower spatial resolution compared to SPECT and PET.

Scintigraphy is used for a wide range of diagnostic purposes, including bone scans, thyroid imaging, and assessing organ function.

6.1. Single photon emission computed tomography SPECT

A single photon emission computed tomography scan is an extension of scintigraphy that shows how blood flows to tissues and organs. It uses gamma cameras to acquire multiple 2D images from different angles, which are then reconstructed into 3D images.
SPECT imaging instruments based on the same principle but provide three-dimensional (tomographic) images of the distribution of radioactive tracer molecules introduced into the patient's body [11]. The most used radioisotopes in SPECT are: iodine-123, technetium-99m, Xenon-133, thallium-201 and Fluorine-18.

The difference between SPECT and PET scan is that the tracer stays in the human body instead than being absorbed by tissues, SPECT scans are cheaper and more readily available than higher resolution PET scans. Figure 9 shows an example of image obtained with a SPECT scan.

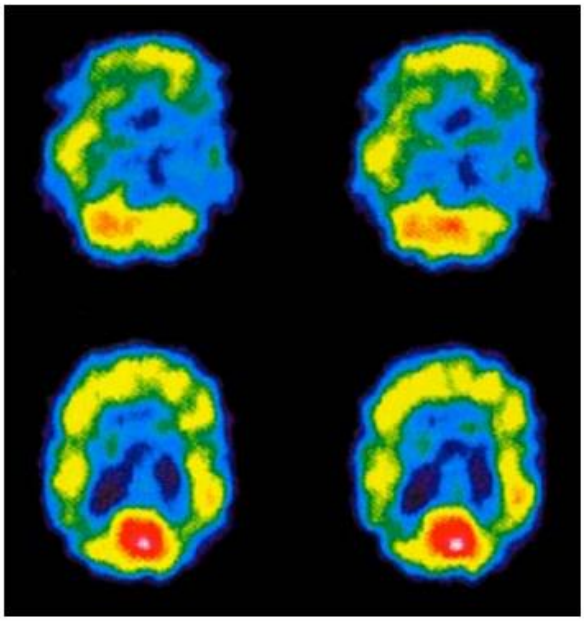


Figure 3.9. Image obtained with a SPECT scan

6.2. Positron emission tomography PET

PET scans also use radiopharmaceuticals to generate three-dimensional images, and produce a functional information about the human body. The main radiotracer commonly used in this technique in clinical practice is Fluorodeoxyglucose-18. FDG-PET scans have been shown to be more accurate in the evaluation of malignancies.

When positron is emitted, it travels a short distance and annihilates with an electron, two gamma rays are propagating and detected by the gamma camera. Distribution of this rays form the final image [13]. Figure 10 shows an example of image obtained with a PET scan.

This technology is used for diagnosis, staging and assessing response to therapy in many cancers.

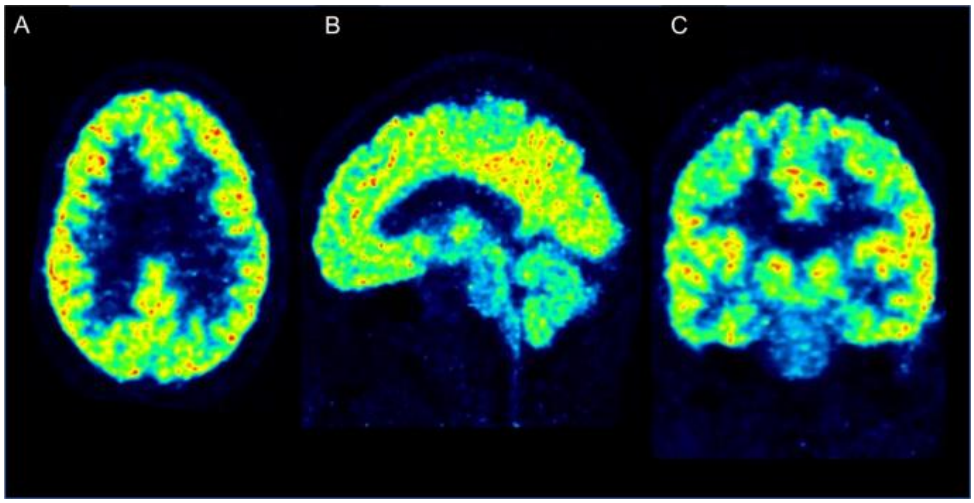


Figure 3.10. Image obtained with a PET scan

7. Scintigraphy applications

Scintigraphy is a special type of nuclear medicine procedure that can help to diagnose and assess the severity of a variety of bone diseases and conditions, among its applications:

7.1. Skeletal scintigraphy

Skeletal scintigraphy determines bone cancer, spread of cancer to other areas and the location of an abnormal bone or diagnose broken bones or inflammations. An example of skeletal scintigraphy is shown in figure 11.

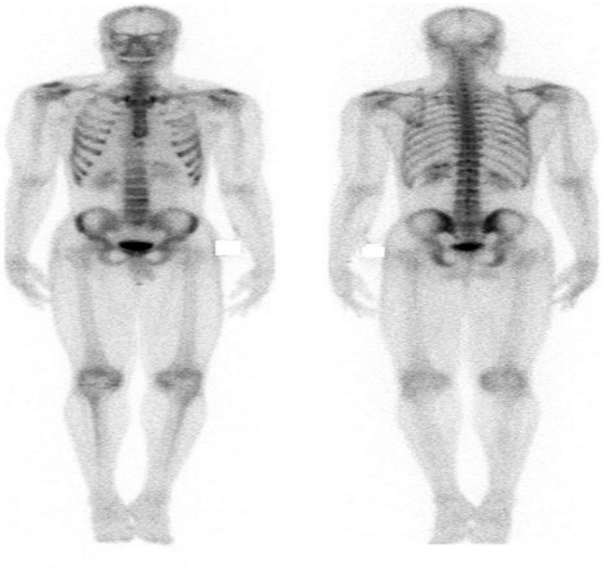


Figure 3.11. Skeletal scintigraphy

7.2. Myocardial scintigraphy

This scan informs about the condition of blood flow, vitality and function of the heart muscle. An example of myocardial scintigraphy is shown in figure 12.

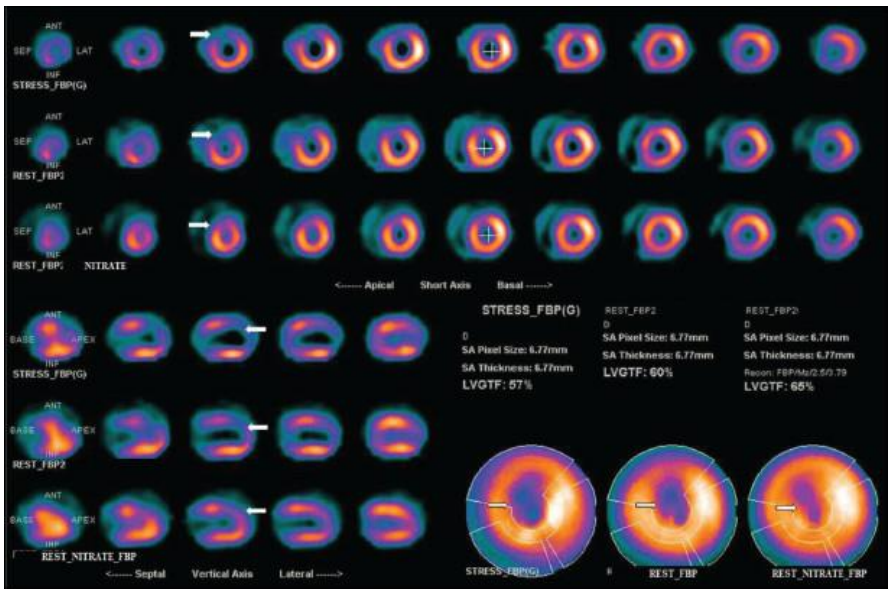


Figure 3.12. Myocardial scintigraphy

7.3. Lung scintigraphy

Figure 13 illustrates a lung scan, showing visualization of lung perfusion and ventilation.

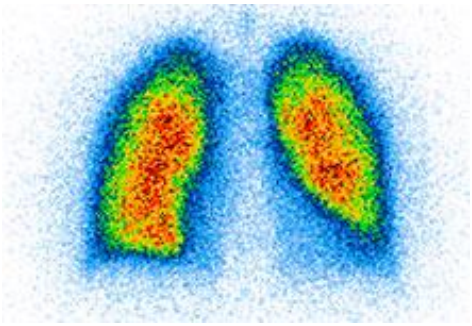


Figure 3.13. Lung scintigraphy

7.4. Lymphoscintigraphy

Lymphoscintigraphy reveals lymph vessels and lymph nodes, as depicted in Figure 14 [14].

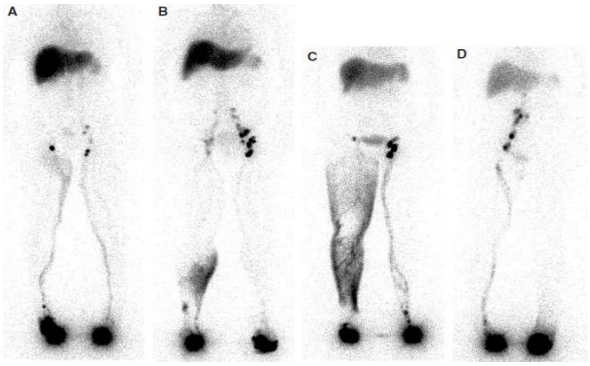


Figure 3.14. Lymphoscintigraphy

7.5. Scintigraphy of kidneys:

This type helps to diagnose renal dysfunction, figure 15 shows an example of kidney scintigraphy.

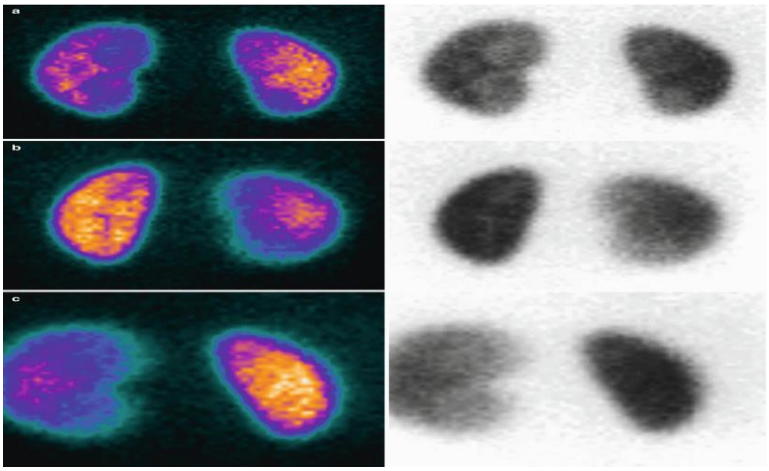


Figure 3.15. Kidney scintigraphy

7.6. Thyroid gland scintigraphy

This scan is typically used in cases of suspected hyperthyroidism, as illustrated in Figure 16.

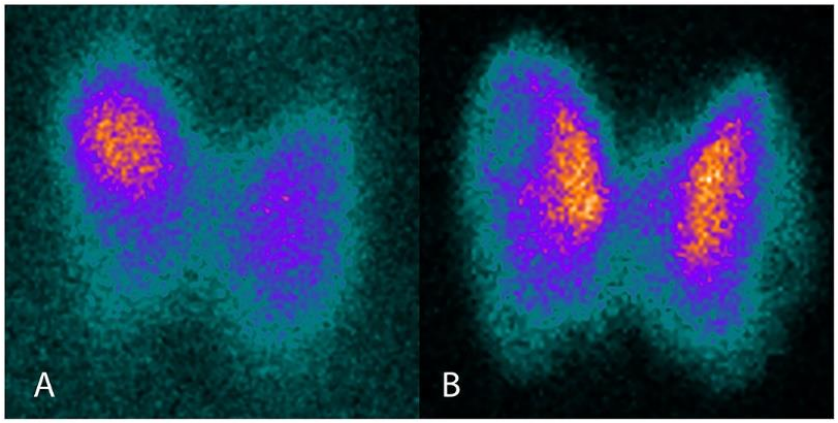


Figure 3.16. Thyroid gland scintigraphy

8. Benefits

In this section, we present some of the benefits that scintigraphy brings to medicine and healthcare:

- Nuclear medicine examinations provide unique information including details on both function and anatomic structure of the body.
- Nuclear medicine scans provide valuable information essential for making a diagnosis or determining appropriate treatment.
- A scan is less expensive.
- PET imaging has the capability to identify the early stages of a disease before it becomes visible on other imaging modalities like CT or MRI.

9. Risks

Nuclear medicine procedures expose the patient to low levels of radiation, which are considered safe for diagnostic examinations. The radiation risk associated with these procedures is extremely low when compared to the potential diagnostic benefits.

Nuclear medicine diagnostic procedures have been in use for over five decades, and there are no known long-term adverse effects associated with such low-dose radiation exposure. While rare, allergic reactions to radiopharmaceuticals may occur.

The injection of the radiotracer may cause mild discomfort, along with localized pain and redness at the injection site.

10. Conclusion

In this chapter, we have discussed scintigraphy, the gamma camera, and the steps involved in forming an interpretable image. Scintigraphy provides both functional and anatomical images, aiding doctors in diagnosing diseases related to bones, the heart, and various vital organs. This technology has significantly transformed the field of nuclear medicine and contributed to patient recovery.

These imaging techniques are widely employed due to their advantages, including reduced radiation exposure, cost-effectiveness, and the ability to detect diseases at an earlier stage compared to MRI and CT scans.

Chapter 4
Tomographic reconstruction

Authored by

Dr. Korti Amel

1. Introduction

Medical image reconstruction involves creating high-quality images from raw data acquired during imaging procedures. Medical imaging technologies, such as X-ray, computed tomography (CT), magnetic resonance imaging (MRI), positron emission tomography (PET), and ultrasound, generate raw data that need to be processed and reconstructed into meaningful images that healthcare professionals can use for diagnosis, treatment planning, and research.

The main goal of medical image reconstruction is to enhance the visibility of anatomical structures and abnormalities while minimizing artifacts and noise. The quality of the reconstructed images directly affects the accuracy of medical diagnoses and subsequent patient care.

Various techniques and algorithms are used in medical image reconstruction, depending on the imaging modality, the most popular methods are:

- **Filtred back-projection (FBP) :** FBP is a widely used technique in X-ray CT, single photon emission computed tomography (SPECT) and positron emission tomography (PET) imaging. It involves passing the raw data through a filter in the frequency domain and then back-projecting the filtered data to obtain an image.
- **Iterative reconstruction :** This approach iteratively refines an initial estimate of the image by comparing the measured data with the estimated data. It's commonly used in CT and PET imaging and can reduce noise and artifacts.

Each imaging modality presents its own challenges and considerations in terms of data acquisition, noise reduction, artifact suppression, and spatial resolution. Researchers and engineers continually work on developing and improving reconstruction algorithms to enhance the accuracy and clinical utility of medical images. This chapter gives an overview of these methods.

In the context of medical imaging, we speak about two problems : Forward and inverse problem.

2. Forward problem

The forward problem describes how the internal tissue properties (unknowns) relate to the measurements acquired by the imaging system (Figure 1) calculating the expected measurements (projections, signal intensities) based on a known image or distribution of properties.

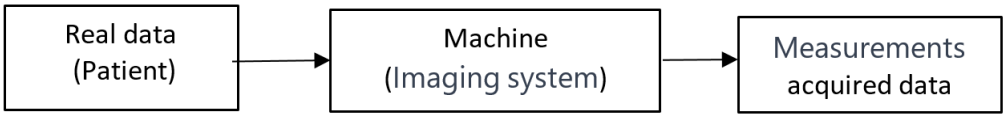


Figure 4.1. Forward problem

Forward projection is the process of adding up all of the pixel values along the measured direction. Figure 2 is an example of 2×2 image matrix demonstrates parallel beam projections from the left to the right through the image pixels.

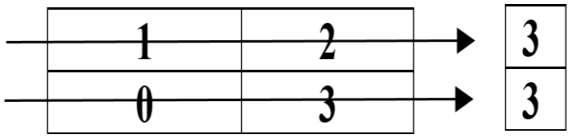


Figure 4.2. Acquired data

Parallel beam acquisition generally used for CTand PET scans, involves using a collimated beam that remains parallel as it passes through the patient and reaches a linear array of detectors on the other side. This approach provides a set of projection measurements from multiple angles, which are then used to reconstruct cross-sectional images of the patient's anatomy. Figure 3 shows a further forward projection angle of 90 degrees to the first forward projection.

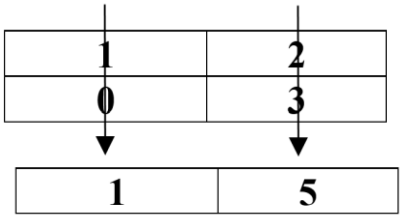


Figure 4.3. Data acquired after 90 degree rotation

2.1.In CT scan

The projection in CT scan represents the sum of X-ray attenuations along a straight path through the patient's body. These projections are acquired from multiple angles as the X-ray source and detectors rotate around the patient (see CT scan chapter for more detail).

Mathematically, the projection at angle ɵ can be represented as a line integral of the attenuation coefficient f(x,y) along the path:

$$p(u,\theta) = \int_{-\infty}^{+\infty} f(x,y)dV \qquad (1)$$

where u is the position along the detector, x and y are coordinates in the patient's body, and ɵ is the angle of projection.

Figure 4 explain CT scan principle. At each angle of rotation, a set of X-ray measurements is acquired. The rotation continues, and a series of projection images is acquired from different angles. Each projection provides information about the attenuation of X-rays along different paths through the patient's body. The measured intensities are recorded and plotted in the sinogram at corresponding angles of projection. These measurements are used to create a 2D projection image.

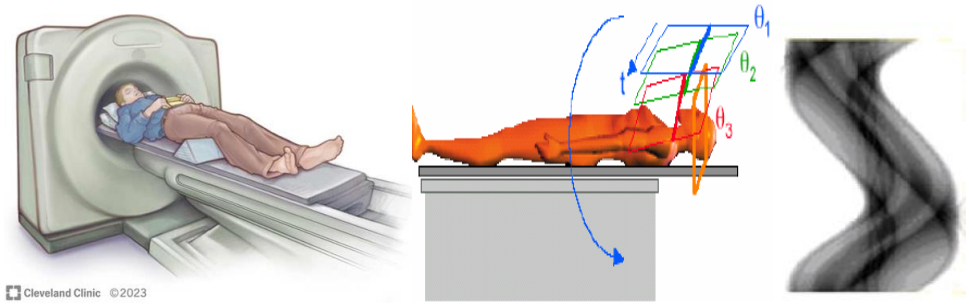


Figure 4.4. CT scan principle and organizing data into a sinogram

2.2. In PET scan

A radiopharmaceutical labeled with a positron-emitting radionuclide is injected into the patient's body. When positrons emitted by the radionuclide encounter electrons in the tissue, they undergo annihilation, resulting in the emission of two 511 keV photons in opposite directions. These photons are detected by a ring of scintillation detectors surrounding the patient (Figure 5).

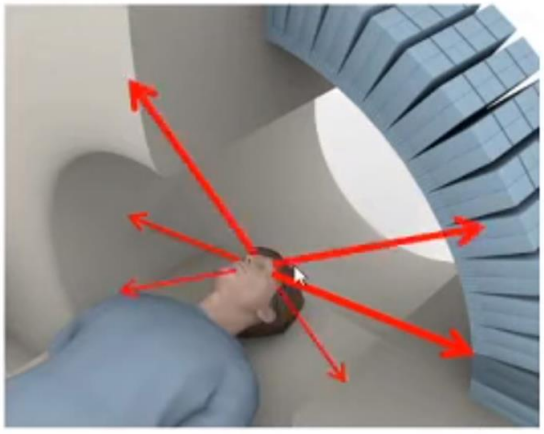
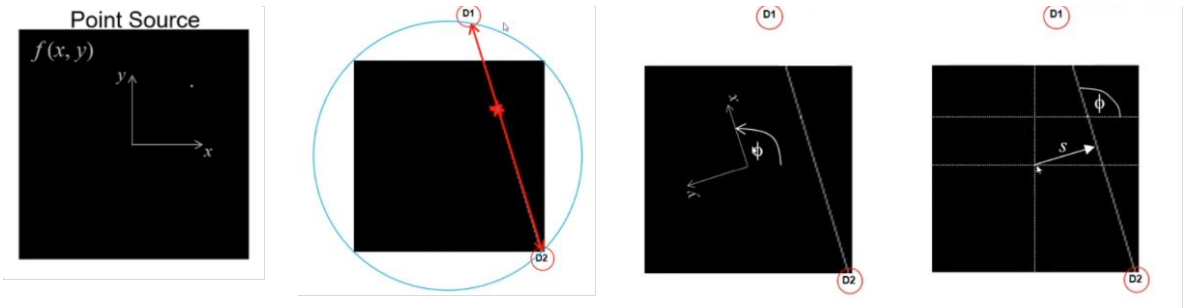


Figure 4.5. PET scanner principle

The imaginary line connecting two detectors that simultaneously detect an emitted gamma photon (in SPECT) or a pair of annihilation photons (in PET) is called the Line of Response (LOR). It represents the path along which the emission or detection event occurred.
Let's consider a single Line of Response (LOR) (Figure 6b) detected from a point source (Figure 6a). We can choose an xoy plane parallel to the LOR (Figure 6c), S vector perpondicular to the LOR and an angle ϕ between the x line and the LOR (Figure 6d). Therfore, the LOR is defined by an angle ϕ and a radial position S.
The count of coincident events for a single projection is obtained by integrating the radiotracer intensity distribution of the radiotracer activity $f(s,l)$ in the transverse plane (s,l) along the LOR specified by angle ϕ and radial position s.
Mathematically, this can be expressed as:

$$m(\phi, s) = \int_{-\infty}^{+\infty} f(s,l)dl \qquad (2)$$

Where :
$m(\phi, s)$ represents the coincident event counts along the LOR for angle ϕ and radial position s.



a b c d

Figure 4.6. Single Line of Response (LOR) [14]

The PET scanner records the coincident events and the associated LORs, the collected data is organized into a sinogram (Figure 7). Computer algorithms use the data from the sinogram to reconstruct an image of the distribution of the radiotracer in the body.

Figure 4.7. Organizing data into a sinogram in PET scan [14]

2.3.Sinogram structure

The sinogram represents several waves or sinusoids which overlap one another. The sinogram serves as the input data for image reconstruction algorithms. In this section, we present the difference between the CT scan and PET scan sinogram.

2.3.1 In CT imaging

The values in the sinogram correspond to the X-ray intensities detected at each angle and each detector element. The horizontal axis of the sinogram represents the detector channel direction (see Figure 8) and the vertical axis represents the angles of projection (θ).

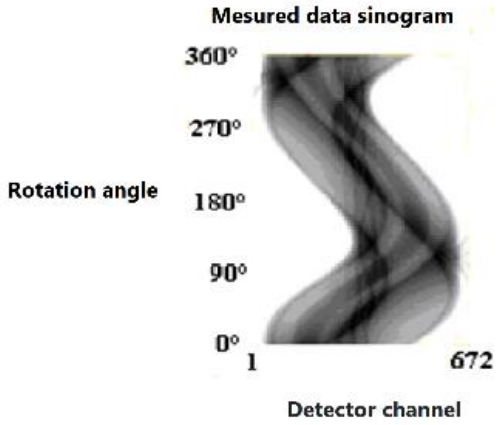


Figure 4.8. CT scan sinogram formation

2.3.2 In PET imaging

The sinogram is a representation of the measured coincidence events, which are pairs of annihilation photons detected by the PET system. Figure 9 shows an example of a PET scan sinogram formation, where (A) shows an object in a scanner and four annihilation events are detected along the LORs A, B, C and D. These LORs can be described as a function of their angular orientation and the minimal distance to the center. (B) shows a sinogram, a graph that represents the angle of a coincidence line versus its displacement from the center. The LORs illustrated in (A) are mapped to this graph. If all possible LORs that pass through point source of (A) are plotted, it maps half of a sine wave.

Each point in the sinogram corresponds to coincidences observed for one angle and one radius. The x-axis of a PET sinogram represents the angle or view of the detectors, each point on the x-axis corresponds to a specific angle at which coincidence events were detected. The y-axis of a PET sinogram is the displacement from the center, it represents the distance or "bin" along

the Line of Response (LOR) between the two detectors that detected a coincidence event. As the detectors rotate, they capture coincidence events at different distances along the LOR.

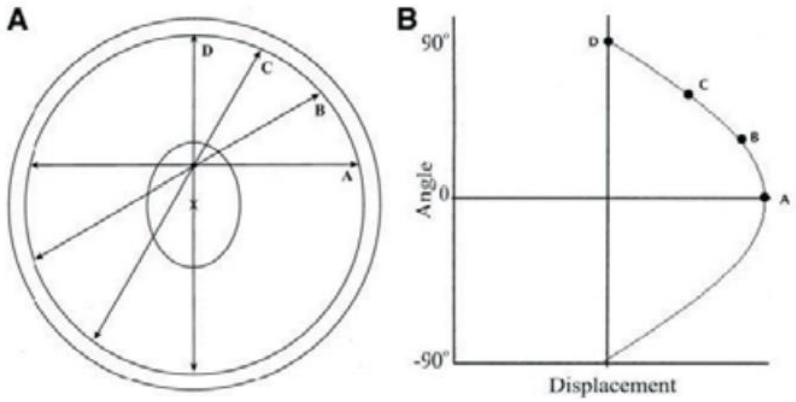


Figure 4.9. PET scan sinogram formation [15]

3. Inverse problem

An inverse problem refers to the challenge of determining the unknown parameters of a system on observed measurements. In medical imaging, image reconstruction is often framed as an inverse problem because we are trying to recover the internal distribution of tissue properties (such as density, relaxation times, or radioactivity concentrations) from measurements (projections, signal intensities) obtained by the imaging system.

Back projection is a step in the process of solving an inverse problem. The back projection process involves taking the acquired projection data and distributing the information from each projection back into the image space. For each point in the image space, the value is incremented based on the contribution of each projection passing through that point. Essentially, the information is "back projected" along the path of each X-ray, accumulating it at each point.

The backprojection operation maps from the detector back to the image, Figure 10 shows the same example of 2×2 matrix, which values are initially in the detector and the backprojection operation paints the values back into the image matrix along the direction of measurement.

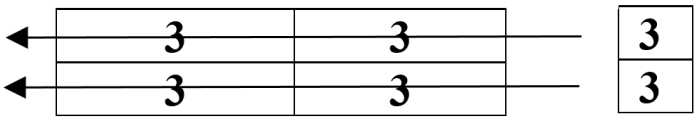


Figure 4.10. Backprojection operation

The back projection step is used to convert the estimated distribution back into an image. The back projection process spreads out the information from the estimated distribution to create an image representation.

In both PET and CT imaging, the Radon transform helps understand how X-ray attenuation or gamma photon interactions occur along different lines of projection. Projection data is collected from various angles around the patient to form projections. Analytic algorithms, like Filtered Back Projection (FBP), use the Radon transform to reconstruct images from projection data.

They use direct mathematical formulas which describes how a 2D image's intensity distribution is projected onto different angles to form projection data.
In 2D imaging, such as CT and PET, the Radon transform represents how an object's intensity distribution is integrated along lines at different angles. For a function f(x,y), the Radon transform p(u,θ) is defined as:

$$p(u,\theta) = \int_{-\infty}^{+\infty} f(x,y)dv \qquad (3)$$

Where the Figure 11 explain all parameters of the Radon transorm.

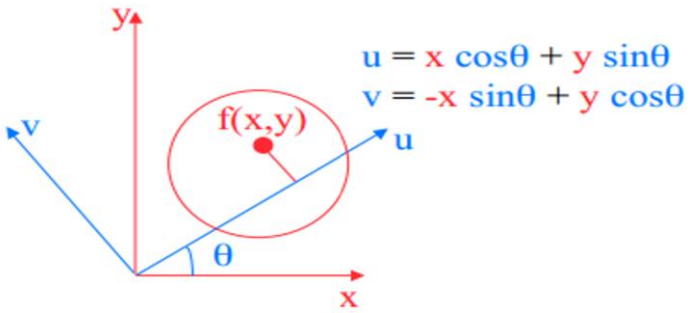


Figure 4.11. Radon transform

The inverse Radon transform is used to reconstruct an image I(x,y) from its projection data. It's a process of "back projecting" the accumulated line integrals along different lines of projection to recreate the original image.
Frequency space analysis involves examining the Fourier transform of both the projection data and the reconstructed image to understand the relationship between spatial frequencies in the data and the reconstructed image. The inverse Radon transform equation describes how the frequency content is transformed from projection data to the reconstructed image.
The Central Slice Theorem provides a bridge between the projection data and the object's Fourier domain representation. It highlights that the Fourier Transform of the projection data contains information about the object's 2D Fourier Transform.

4. Central section theorem

Central section Theorem also known as the Central Projection Theorem, states that the Fourier Transform of a projection $p(u,\theta)$ of an object $f(x,y)$ is equal to the 2D Fourier Transform of the object itself evaluated at angle ϴ.

4.1. Central section theorem in CT scan

Let $p(\upsilon,\theta)$ represents the Fourier transform of the projection data P(u,θ), υ represents the spatial frequency along the projection direction, and ϴ represents the angle of projection.

$$p(u,\theta) = \int_{-\infty}^{+\infty} f(x,y)dv \qquad (4)$$

The Fourier Transform of this projection is :

$$p(\upsilon,\theta) = \int_{-\infty}^{+\infty} p(u,\theta)e^{-i2\pi\upsilon u}du \qquad (5)$$

Replacing $p(u,\theta)$, we find :

$$p(\upsilon,\theta) = \int_{-\infty}^{+\infty}\int_{-\infty}^{+\infty} f(x,y)dv\, e^{-i2\pi\upsilon u}du \qquad (6)$$

Replacing v_u by : $v_u=v(x\cos(\Theta)+y\sin(\Theta))=x(v\cos(\Theta))+y(v\sin(\Theta))=xv_x+yv_y$ (see Figure 11).

And assuming that : $v_x=v\cos(\Theta)$ et $v_y=v\sin(\Theta)$, we find :

$$p(\upsilon,\theta) = \int_{-\infty}^{+\infty}\int_{-\infty}^{+\infty} f(x,y)\, e^{-\left(i2\pi(v_x x+v_y y)\right)}dxdy \qquad (7)$$

We can also define the 2D Fourier transform of the object $f(x,y)$:

$$F(v_x,v_y) = \int_{-\infty}^{+\infty}\int_{-\infty}^{+\infty} f(x,y)\, e^{-(i2\pi(v_x x+v_y y))}dxdy \qquad (8)$$

We find the same results :

$$P(v,\Theta)=F(v_x,v_y) \qquad (9)$$

In another way, if we choose one slice in the 2D Fourier transform of the object $f(x,y)$, we find :

$$\begin{aligned} s(v_x) = F(v_x,0) &= \int_{-\infty}^{+\infty}\int_{-\infty}^{+\infty} f(x,y)\, e^{-i2\pi(v_x x)}dxdy \qquad (10)\\ &= \int_{-\infty}^{+\infty}\left[\int_{-\infty}^{+\infty} f(x,y)\,dy\right] e^{-i2\pi(v_x x)}dx \\ &\int_{-\infty}^{+\infty} p(x) e^{-i2\pi(v_x x)}dx \end{aligned}$$

Which is just the Fourier transform of the projection $p(x)$.

Figure 12 explain that the Fourier Transform of a projection $p(u,\theta)$ of an object $f(x,y)$ represents slice in the 2D Fourier transform of the object itself evaluated at angle θ.

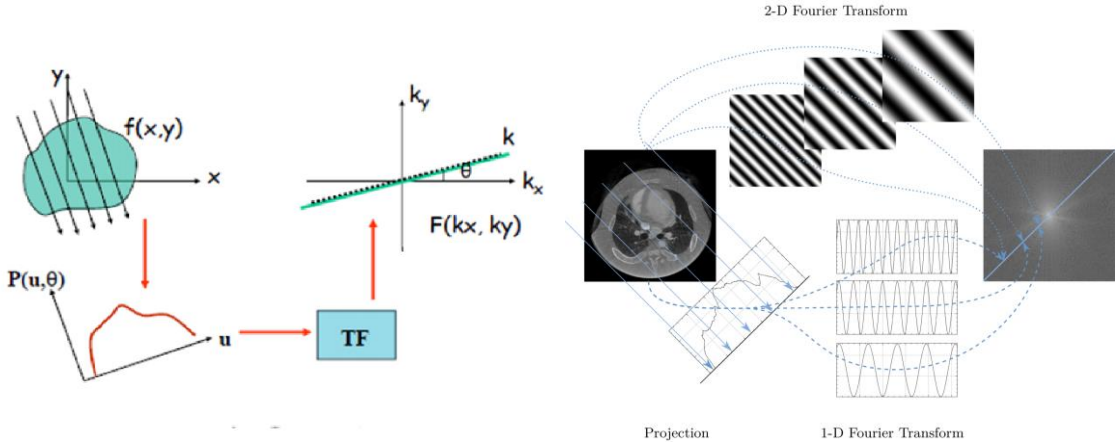


Figure 4.12. Central section theorem

The projection data in frequency space, often represented using the Fourier transform, helps analyze the frequency content of the projection data acquired at different angles.

Application example

In X ray CT, we typically have at least 512×512 matrix ; that means 512×512 pixels in a CT image. If we perform the backprojection operation for just one view the image will look like a smeared painting where the artist could only pull the brush across one time (see Figure 13).

Figure 4.13. Backprojection operation for one angle

As we add the second view to the backprojection image, we add informations. The Figure 14 shows an example of backprojection operations using 2, 4, 8, 36, 72, 144 angles respectively.

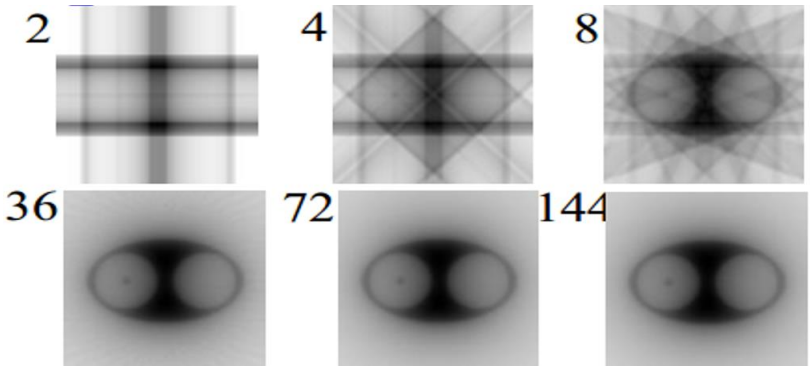


Figure 4.14. backprojection operation from 2, 4, 8, 36, 72, 144 angles respectively

After the backprojection process is completed for several projections, the result is a reconstructed image that represents the distribution of the imaged property (e.g., tissue density in CT) within the patient's body. Back projection results in blurred images and various artifacts.

4.2. Central section theorem in PET scan

Let $M_{\emptyset}(k_s)$ represents the Fourier transform of the projection data m(s,l) with fixed angle of projection ϕ, and k_s represents the spatial frequency along the projection direction.

$$\mathrm{m(s)} = \int_{-\infty}^{+\infty} f(s,l)dl \tag{11}$$

$$M_{\emptyset}(k_s) = \int_{-\infty}^{+\infty} m_{\emptyset}(s)\exp(-ik_s s)\,ds \tag{12}$$

$$M_{\emptyset}(k_s) = \int_{-\infty}^{+\infty}\int_{-\infty}^{+\infty} f(s,l)\exp(-ik_s s)\,dsdl \tag{13}$$

Like CT scan, using the central projection theorem we have :

$$f(k_s, k_l) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} f(s,l) \exp(-ik_s s - ik_l l)\, dsdl \qquad (14)$$

The Figure 15 shows an example of simple backprojection of data from 256 projection angles. Bluring is apparent in the object, and edge detail are lost.

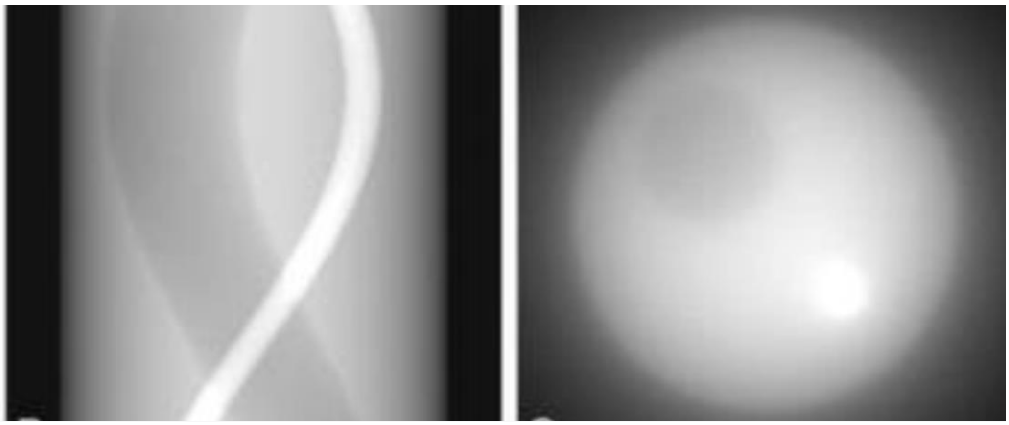


Figure 4.15. Backprojection from 256 projection angles.

In both CT scan and PET scan, back projection results in blurred images and various artifacts. To remove the blurring more advanced image reconstruction techniques have been developed such as filtering, regularization, and iterative optimization.

5. Filtered back projection (FBP)

This classic algorithm involves filtering the projection data in the frequency domain and then back projecting it to reconstruct the image. The projected profile is filtred to remove the typical starlike blurring that is caracteristic of the simple back projection technique. It's efficient and widely used, especially for parallel-beam geometry.

In the frequency domain (Figure 16), the transformed data can be visualized as a set of frequency components. Low-frequency components represent smooth variations in the image, while high-frequency components represent rapid changes or noise.

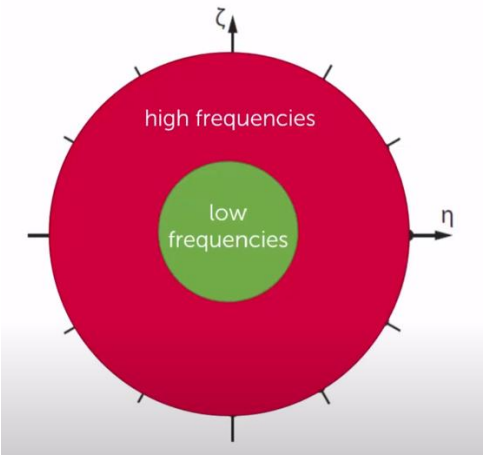


Figure 4.16. 2D frequency domain

The FBP algorithm involves two main steps: filtering and back projection.

5.1. Filtering

The first step in FBP is filtering the projection data in the frequency domain. The goal is to correct for blurring effects caused by the imaging system. The filtered projection data is represented by $F(\upsilon, \theta)$ where υ represents the spatial frequency and θ represents the angle of projection.

The filtering operation typically involves multiplying the Fourier transform of the projection data by a frequency-dependent filter function $H(\upsilon)$. This function enhances high-frequency components (fine details) and attenuates low-frequency components (blurry regions).

The filtering process can be represented as:

$$F(\upsilon, \theta) = p(\upsilon, \theta).H(\upsilon) \quad (15)$$

The choice of the filter function significantly impacts the quality of the reconstructed image. Ramp filter, also known as the Ram-Lak filter, is a basic filter used in the filtering step of the Filtered Back Projection (FBP) algorithm. The name "Ramp" comes from the linear response (the shape) of the filter in the frequency (Figure 17).

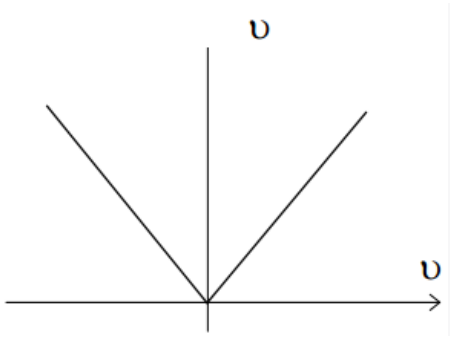


Figure 4.17. Ramp filter

The Ramp filter is defined mathematically as follows:

$$H(\upsilon) = |\upsilon| \text{ for } |\upsilon| \leq \upsilon_c \quad (16)$$

$$H(\upsilon) = 0 \text{ for } |\upsilon| > \upsilon_c \quad (17)$$

where, $H(\upsilon)$ is the frequency response of the filter, and υ represents the spatial frequency and υ_c is the cutoff frequency that determines the point at which the filter starts attenuating frequencies.

The Ramp filter operates on the Fourier transform of the projection data, it multiplies the Fourier coefficients by the frequency response of the filter. This process enhances high-frequency components while suppressing low-frequency components. The resulting filtered projection data is then used for the back projection step in FBP.

The Ramp filter enhances the visibility of fine details in the reconstructed image. High-frequency information is essential for capturing sharp edges and small structures.

While the Ramp filter is simple and widely used, it can introduce ring artifacts in the reconstructed image. These artifacts manifest as concentric rings around high-contrast structures. These artifacts are due to the abrupt transition of the filter function at the cutoff frequency. The Ramp filter does not account for noise in the projection data. In practice, noise can be amplified in the reconstructed image, especially in regions with low signal

intensity. To mitigate ring artifacts, a windowed Ramp filter is often used. This involves applying a window function to the Ramp filter to gradually taper the filter response instead of an abrupt cutoff. The common method to reduce or remove statistical noise is the application of smoothing filters. There are various approaches that can be employed (see Figure 18), each with its own characteristics and benefits, such as :

- Shepp-Logan Filter: A modified ramp filter that reduces ringing artifacts.
- Hann Filter: Balances the trade-off between resolution and noise.
- Hamming Filter: Similar to the Hann filter but with different properties.
- Butterworth Filter: Provides adjustable control over the balance between noise suppression and image resolution.

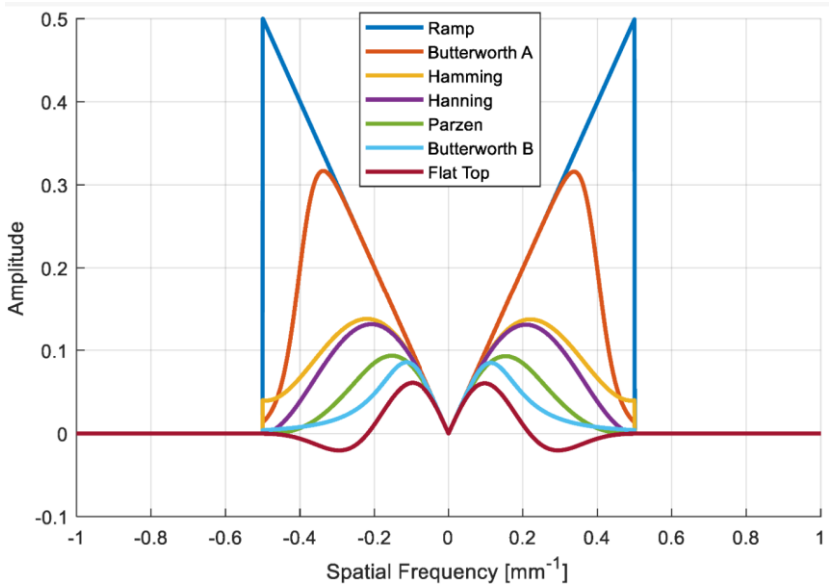


Figure 4.18. Different band limited filter shapes as a function of frequency response [16]

5.2. Back projection

In both CT and PET scan, the filtered projection data is then back projected to reconstruct the image. Back projection involves accumulating the contributions from different angles to each pixel in the image. The back projection operation is inherently a process of summation and interpolation. In CT scan, the resulting data are back-projected onto the image plane from a set of X-ray projections acquired from multiple angles around the patient's body. In PET scan, the resulting data are back-projected onto the image plane along each LOR. The back-projected data from multiple angles are summed up to reconstruct the final PET image. The reconstructed image represents the distribution of tracer activity within the imaged object.

The Figure 19 explains the Filtered backprojection process. The FBP uses a 1D projection filter (ramp filter $|\upsilon|$) along the row direction in frequency domain followed by a backprojection operation to spread the projection data back into the image.

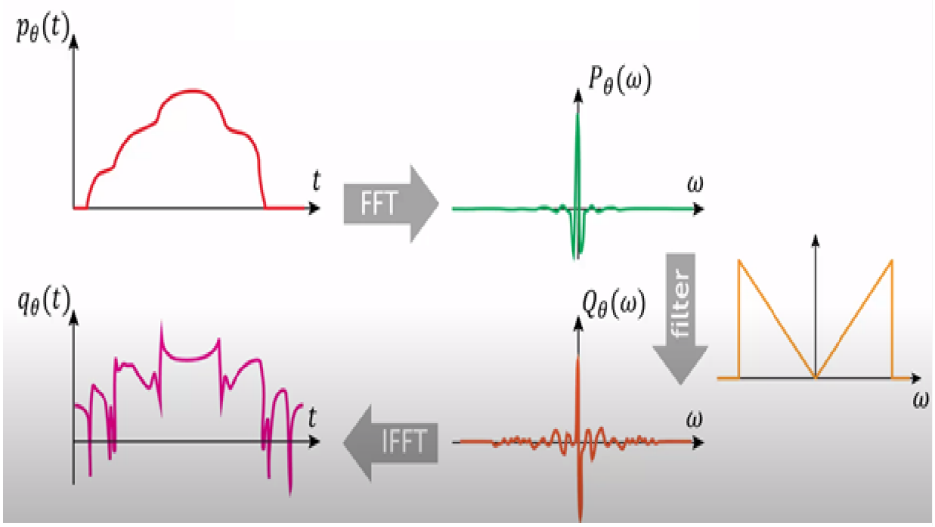


Figure 4.19. Filtered beckprojection

For a given point (x,y) in the image space, the back projection operation involves integrating the filtered projection data along all possible projection angles ɵ. Figure 20 shows a simple representation of filtered back projection

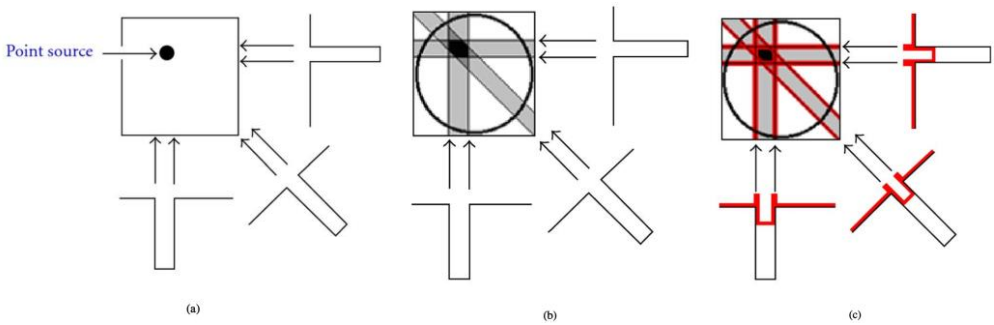


Figure 4.20. Filtered back projection. (a) Acquisition of three projections. (b) Backprojected projections. (c) Filtered backprojected projections [17]

Figure 21 shows the same image for the Figure 14 using filtred backprojection from 2, 4, 8, 36, 72, 144 angles respectively with Hann Filter .

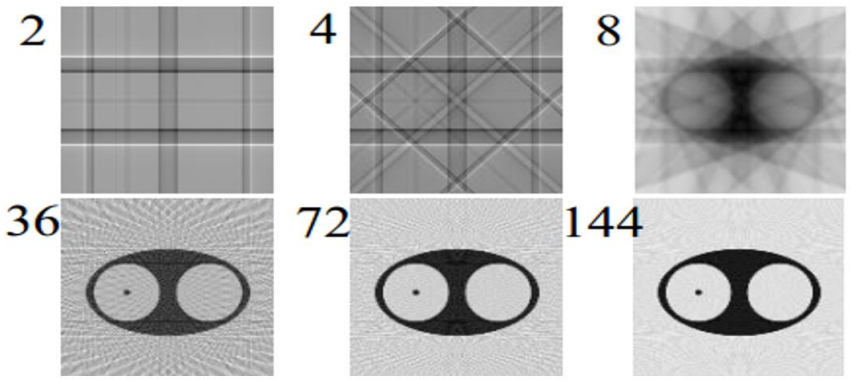


Figure 4.21. Filtred backprojection from 2, 4, 8, 36, 72, 144 projection angles respectively

The filter function is designed to shape the frequency components in a way that enhances the quality of the reconstructed image. The goal is to emphasize the relevant information and suppress noise and artifacts.

It's important to note that FBP is a fast and efficient reconstruction method but may not handle all types of artifacts and corrections as effectively as iterative methods and hybrid approaches that combine FBP with iterative techniques.

6. Iterative reconstruction

In direct methods, the matrix model relates the acquired projection data to the image being reconstructed through a series of mathematical operations. The goal is to directly invert this model to obtain the image.
Let's assume we have a 2D image matrix I and a sinogram (projection data) matrix P. Each element of the sinogram represents the line integral of the image along a specific projection angle and distance.
The basic equation for the forward projection (creating the sinogram) can be represented as:

$$P=A.I \tag{18}$$

Where A is the system matrix that represents the geometry and physics of the imaging system. For example, in X-ray CT, the system matrix A takes into account the attenuation of X-rays through different tissues in the body and the geometric setup of the CT scanner.
In direct methods like Filtered Back Projection (FBP), the reconstruction involves inverting this equation to solve for the image matrix I given the sinogram matrix P and the known system matrix A-1.
The problem of reconstructing an image from its projections can be ill-posed if there are insufficient projections or if noise is present. An ill-posed problem is a mathematical problem that lacks one or more of the following properties:

- Existence: A solution exists.
- Uniqueness: The solution is unique.
- Stability: Small changes in the input data result in small changes in the solution.

In other words, an ill-posed problem may not have a well-defined, stable solution, or the solution might be very sensitive to changes in the input data. Solving ill-posed problems can be challenging and may require additional information or assumptions to arrive at a reasonable solution.
An ill-conditioned problem refers to a situation where a small change in the input data or parameters of a mathematical problem results in a significantly large change in the solution. In other words, the problem's solution is highly sensitive to perturbations in the input.
Mathematically, the condition number of a problem quantifies its sensitivity to input changes. A high condition number indicates that the problem is ill-conditioned and small changes in input can lead to large changes in output.
These terms highlight the importance of careful analysis, regularization techniques, and numerical stability in addressing such problems.
Example :

- Noise makes problems ill conditioned (e.g. PET data)
- Incomplete data makes problems ill posed (e.g. MR : undersampling ; CT : missing angles)
- Ill posed problems (i.e. noise sensitive, or with non unique solution, or no solution) are resolved through the use of regularisation.

Least squares method is one of the methods that can be used to solve the ill posed problem in the reconstructed image. The goal is to minimise the difference between projection data matrix P and AI, so that to concel the noise n :

$$P=A.I+n \quad (19)$$

$$P-A.I=0 \quad (20)$$

To avoid negative and positive differences cancelling each other out, we seek a least square solution :

$$L(I)=\sum(P-A.I)^2 \quad (21)$$

We can write the least squares problem as

$$I_{LS} = argmin\|P-A.I\|_2^2 \quad (22)$$

By definition : $\|a\|_2 = \sqrt{\sum_1^N |a_n|^2}$ is the L2 norm
The direct methods require the system matrix A to be stored in memory as well as finding A^TA and its inverse.
In iterative reconstruction methods, the backprojection step is part of an iterative optimization process. The image estimate is refined iteratively based on a comparison between the acquired data and the estimated data.
In iterative methods, the matrix model involves iteratively updating the image estimate to match the acquired projection data. Various methods for least squares are used :

- **SART (**rows or columns access to matrix A), weighted least squares objective (similar to batch gradient descent)
- **ART**(uses row access to matrix A), weighted least squares objective (similar to stochastic gradient descent (SGD).
- **ISRA** (image space reconstruction algorithm, row or column access to matrix A) non negative least squares objective.
- **Steepest descent** (uses optimised step size)
- **Conjugate gradient** (uses optimised step size, combining gradient with previous update direction) a fast finite series expansion of the solution

The above all differ by : amount of data used in each update, relaxation parameter, weighting or not of least squares, non negative or not….
The advantage of iterative methods is that it is often sufficient to act on the rows or columns of the matrix: matrix elements can be calculated as required and are not stored in memory.

7. Conclusion

Medical image reconstruction involves the transformation of acquired raw data into meaningful and diagnostically valuable images across various imaging modalities such as CT, MRI, PET, ultrasound, and more. Image reconstruction techniques can be categorized into iterative and analytic approaches, FBP is a foundational analytic method used in CT, SPECT and PET imaging, combining filtering and back projection to create images. However, it has limitations in addressing complex phenomena. Modern imaging often employs iterative algorithms that refine reconstructions through multiple iterations, allowing for improved image quality, noise reduction, and artifact mitigation.

Chapter 5
Ultrasound scans

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1. Introduction

Ultrasound was discovered by Lazzaro Spallanzani, when he demonstrated the ability of bats to navigate through backscattering sound waves of inaudible sound in 1794. Over the past 40 years, ultrasound has evolved into an important diagnostic method. Its potential in medical imaging was recognized in the 1930s and 1940s when Theodore Dusik and his brother Friederich attempted to use ultrasound to diagnose brain tumors. But it wasn't until the 1970s that research really came to fruition.

Advances in technology have evolved ultrasound from a bulky, cumbersome device that provided suboptimal images to a sophisticated device that was portable and easy to use. Such development required connections between physics, physiology, medicine, engineering and government [18]. This chapter describes the physical characteristics and nature of ultrasound, the interaction of ultrasound with matter, followed by a description of ultrasound machines, modes of imaging, the biological effects of ultrasound. It concludes with a conclusion.

2. Physical characteristic of ultrasound

Ultrasound is defined by some physical characteristics, which are properties common to all types of waves. These characteristics are presented in details below:

2.1. Frequency f(Hz)

The frequency of sound is the number of cycles per one second (s), expressed in hertz (1 cycle per second). Frequency is an important property of ultrasonic sound images that affect sound penetration and image quality. The frequency is given by:

$$f=1/T \text{ number of period per second (Hz)} \quad (1)$$

Where f is the frequency and T is the period.

Figure 1 shows an example of frequency: in the wave **A**, the frequency is 2 cycles/second or 2 hertz and in the wave B frequency is 3 cycles/sec or 3 hertz [19].

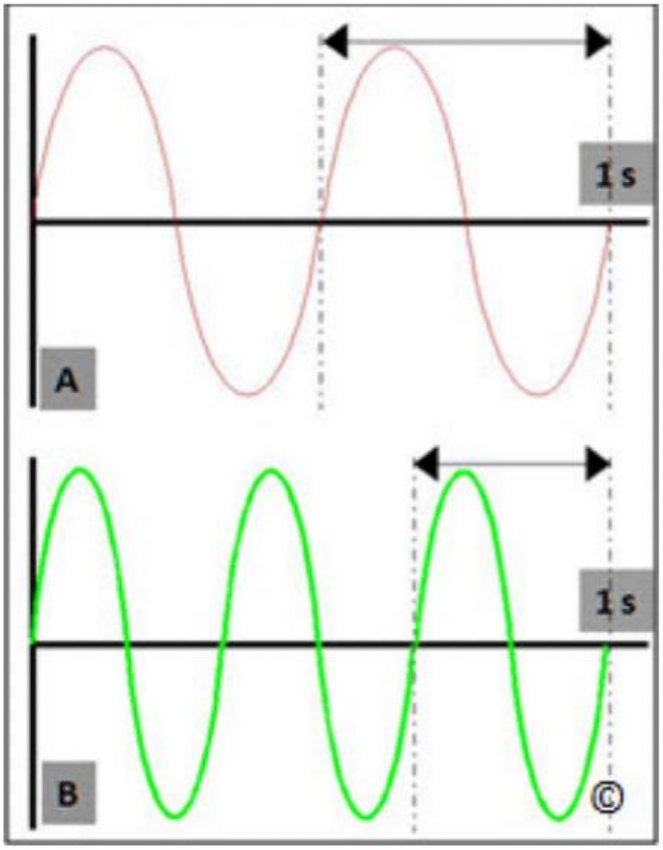

Figure 5.1. Frequency example

2.2. Amplitude A

Figure 2 shows an example of amplitude (A), defined as the difference between the maximum or the minimum of the wave and the average value. The unit of amplitude is Pascals (Pa).

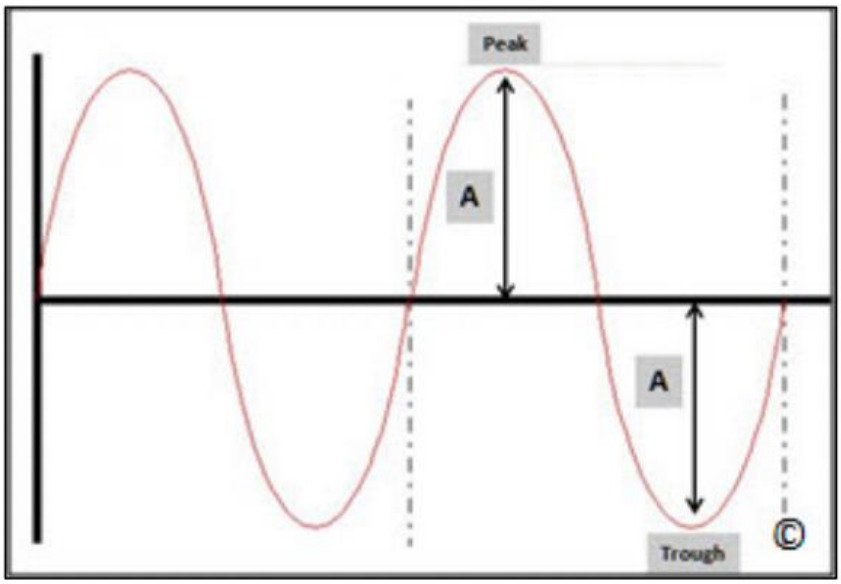

Figure 5.2. Amplitude example

2.3. Wavelength ƛ(m)

The wavelength of a sound wave is the length of the wave, it is defined as the distance of a complete cycle. It is designated by the symbol lambda (λ), and is expressed in mm. figure 3 (a to c) shows 3 sound waveforms of different wavelengths [20].

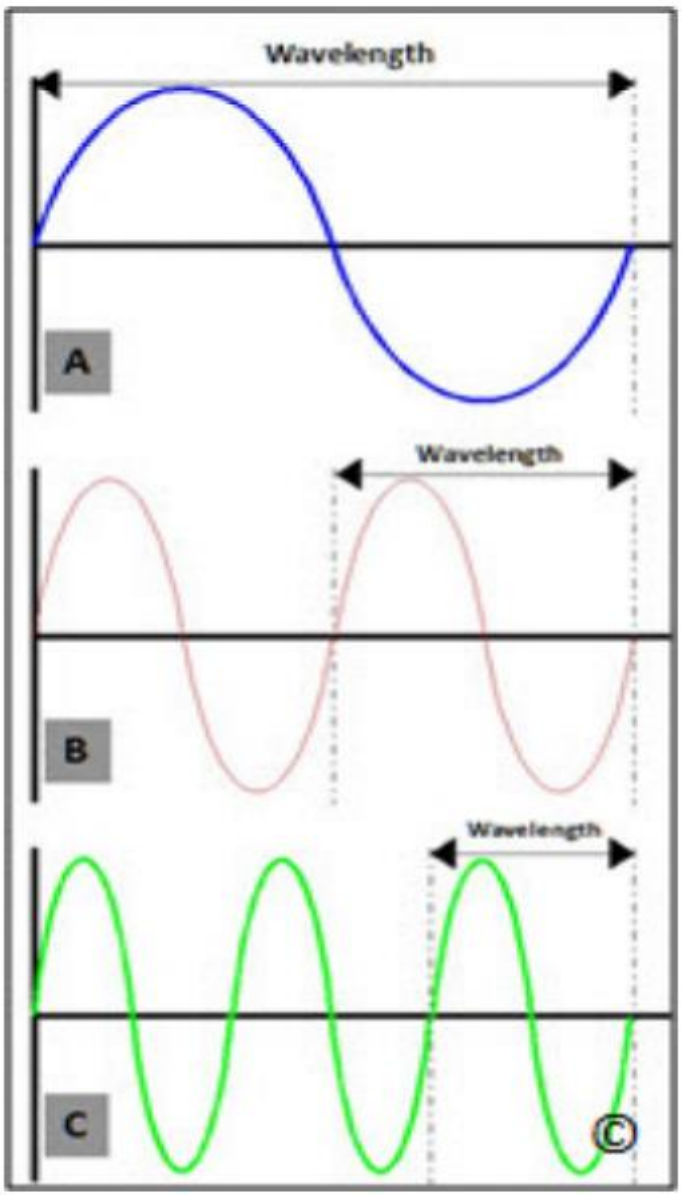

Figure 5.3. Wavelengths example

2.4. Propagation speed (m/s)

The speed of sound propagation in soft tissue is constant at 1,540 m/s. The speed of sound propagation is given by:

$$C = \lambda \, x \, f \;\; (m/s) \qquad (2)$$

Where C is the speed of sound, f is the frequency and λ is the wavelength. table 1 shows the different sound propagation values in selected biologic materials.

Table 5.1. Speed of sound in various media

Medium	Speed (m/s)
Air	330
Fat	1.450
Water	1.450
Soft tissue	1.540
Bone	3.500
metals	Up to 7.000

2.5. Period T(sec)

The period is the time required for one wavelength to pass through a certain point. Generally, a longer period indicates a lower pitch. Figure 4 shows an example of ultrasound period with some characteristics.

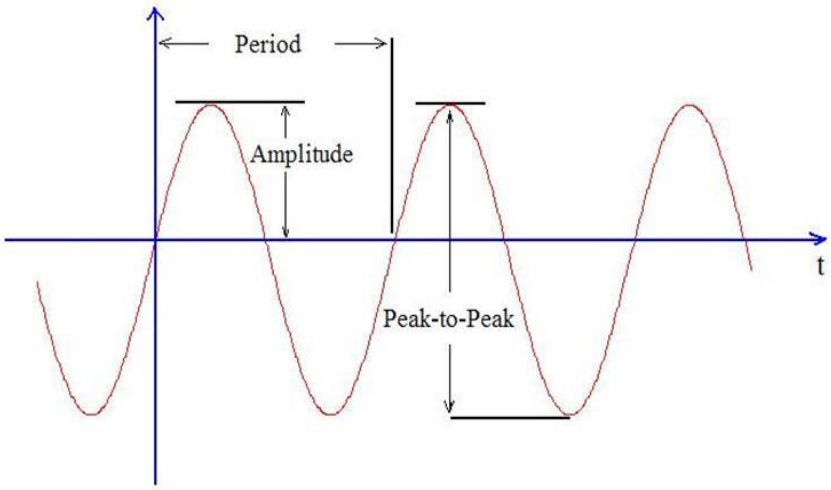

Figure 5.4. Period example

2.6. Nature of ultrasound

A vibrating source is responsible for producing a sound. Sounds are classified according to the hearing capacity of the human ear. The human ear is sensitive to frequencies around 20 KHz, which we call audible sounds. Sound frequencies below 20 Hz are inaudible to the human ear and are called infrasound; frequencies above 20 kHz are called ultrasound. Table 2 shows frequencies of different sound wave [20].

Table 5.2. Frequency spectrum of sound

Sound wave	Frequency
Ultrasound	Greater than 20 KHz
Audible sound	20 Hz to 20 KHz
Infrasound	Less than 20 Hz

3. Interaction of ultrasound with matter

A perfect ultrasound scanner produces a large number of small ultrasound beams, an incredible scanning speed and uniform energy along the entire length of the beam. The interaction of ultrasound with tissue can lead to measurement errors, artifacts and poor image quality. To avoid these errors, we need to understand the basic interactions of tissue with ultrasound, which has also led to the development of new technologies, such as automatic border detection. Figure 5 shows how ultrasound waves interact with tissue in four different ways. These interactions are as follows: [21]

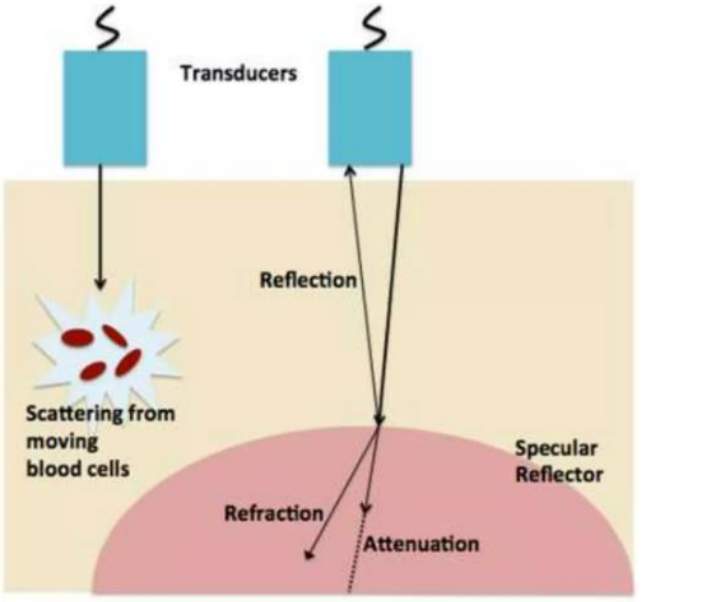

Figure 5.5. Ultrasound interaction with matter

3.1.Reflection

Reflection occurs when the ultrasonic wave is deflected towards the transducer. Reflection is sometimes affected by the following factors :

3.1.1 Incidence Angle

The angle of incidence is the angle between the sound beam and the reflecting surface, as shown in figure 6.

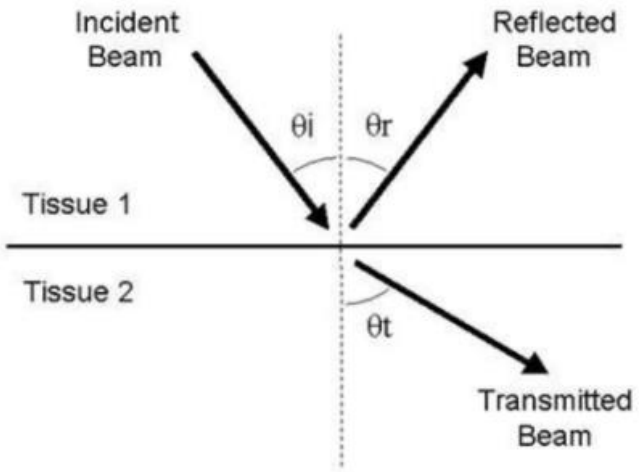

Figure 5.6. Angle of incidence

3.1.2 Acoustic impedance mismatch

The relationship between the pressure on an imaginary surface in a sound wave and the rate of particle flow across the surface is given by the acoustic impedance Z:

$$Z=p\ v \qquad (3)$$

Where P is the density and V is the velocity of sound. Table 3 shows some acoustic impedances of materials.

Table 5.3. Acoustic impedances of some materials

Material	Acoustic impedance
Air	0.0004
Fat	1.38
Water	1.54
Brain	1.58
Blood	1.61
Kidney	1.62
Liver	1.65
Muscle	1.70
Skull (bone)	7.8

3.1.3 Width of the tissue boundary

The tissue boundary width affects the reflected signal. If the tissue boundary width is less than the wavelength of the ultrasound wave, there will be no reflection of ultrasound. Otherwise, the ultrasound wave acts as a mirror or specular reflector, resulting in a significant reflection of the signal.

3.2. Refraction

Refraction occurs when the ultrasound signal is reflected from a straight path and the angle of reflection moves away from the transducer (figure 7). Refraction can improve image quality by using acoustic lenses.

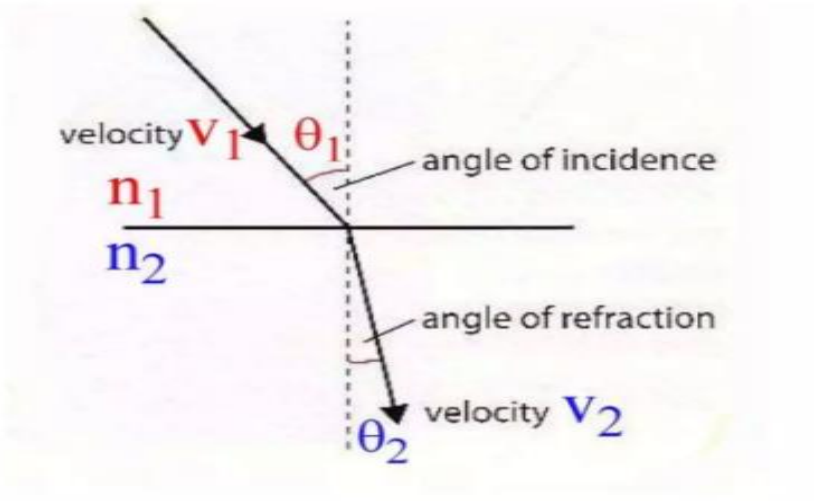

Figure 5.7. Refraction

The angle of refraction is determined by the change in the speed of sound that occurs at the boundary and is related to the angle of incidence by snell's law:

$$\frac{\sin \theta_t}{\sin \theta_i} = \frac{c2}{c1} \quad \textbf{(4)}$$

Where $\boldsymbol{\theta t}$ is the incidence angle, $\boldsymbol{\theta i}$ is the transmitted angle, **c1** is the velocity of sound for incident medium and **c2** is the velocity of sound for transmitting medium

3.3. Absorption

Absorption is the main form of attenuation. It happens when sound pass through soft tissue, the particles that transmit the waves vibrate and produce friction, a loss of sound energy and heat.

3.4. Scattering

Scattering happens when the width or lateral dimension of the tissue boundary is less than one wavelength. If a large number of small tissue boundaries occurs, the scattering can radiate in all directions.

The signal that reaches the transducer is a much weaker signal than the transmitted signal. Most scattering occurs with red blood cells, which is 20 times smaller than the ultrasound wavelength. A filter can ignore small signals from red blood cells below a threshold value. [22]

4. Ultrasound machine

Ultrasound has evolved considerably over the years, becoming more compact and delivering high-quality images. An ultrasound machine uses a widely-used imaging method: high-frequency sound waves to generate live images.

A basic ultrasound machine has the following parts (see figure 8):

• **Transducer probe:** probe that sends and receives the sound waves.

• **Central processing unit (CPU):** computer that does all of the calculations and contains the electrical power supplies for itself and the transducer probe.

• **Transducer pulse controls:** changes the amplitude, frequency and duration of the pulses emitted from the transducer probe.

• **Display:** displays the image from the ultrasound data processed by the CPU.

• **Keyboard/cursor:** inputs data and takes measurements from the display

• **Disk storage device (hard, floppy, CD):** stores the acquired images.

• **Printer:** prints the image from the displayed data.

• **Monitor:** The monitor is where the ultrasound technician views the results of the scan and it generally looks similar to a home computer monitor or television.

• **Control Panel:** The control panel is connected to the computer and allows the sonographer to alter aspects of the scan [23].

Figure 5.8. ultrasound machine

5. Probes

Ultrasound probes, called transducers, are placed on the skin and produce waves with frequencies above the threshold of human hearing. It mainly contains a crystalline element considered to be its main component .

5.1. Crystal elements

The crystal element (see figure 9) is the most important component of the transducer. It is a thin disc of piezoelectric material located near the front surface of the transducer. The crystal material may possess its piezoelectric properties naturally, induced using a combination of thermal and electrical treatment. It is important to note that the piezoelectric properties of an artificial crystal can be destroyed if the crystal is heated to high temperatures.

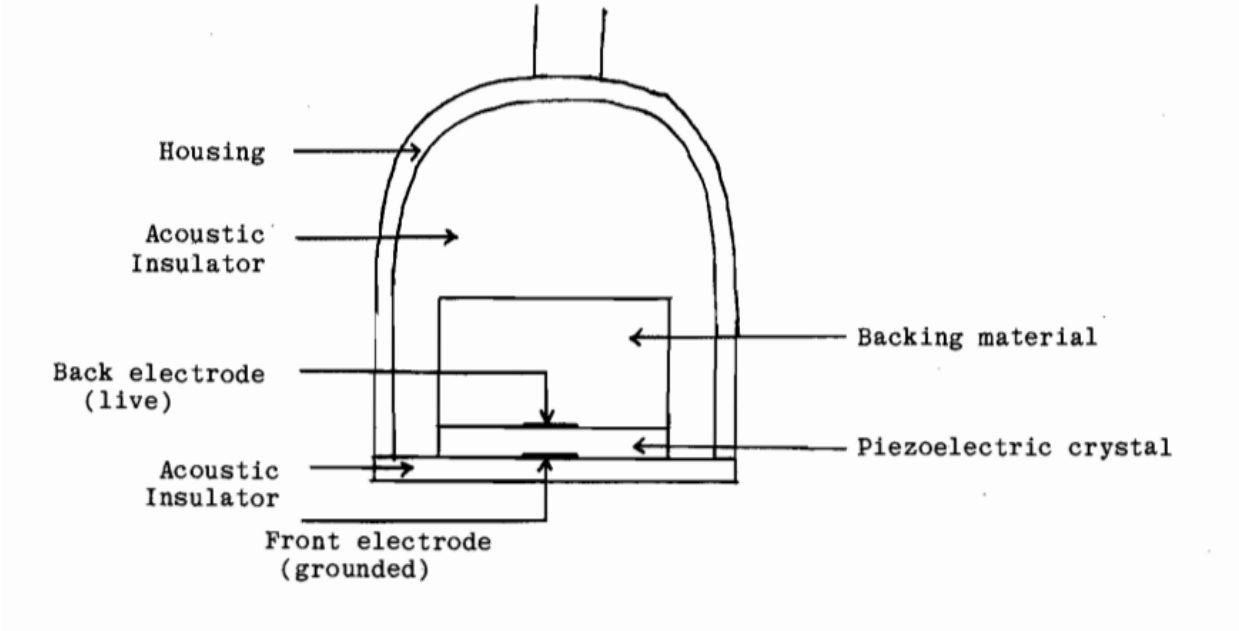

Figure 5.9. Crystal elements

There are various *types of probe* used in ultrasound machine. They are selected and used according to the type of examination.

5.2. Types of probes

Ultrasound transducers come in many shapes, sizes and specifications. This is due to the need for different specifications to maintain image quality in different parts of the body. Figure 10 shows the most commonly used ultrasound probes: 'linear and convex probes'.

Linear and convex transducers are scanned by selecting a set of active elements whose number compromises aperture. The system works by energizing the next element in the aperture and putting the last one at rest. This is done sequentially, until the entire region beneath the transducer is scanned. In linear arrays, the size of the transducer defines the size of the scanning region (~10 cm). In convex transducers, a larger scanning region is obtained by changing the shape of the transducer.

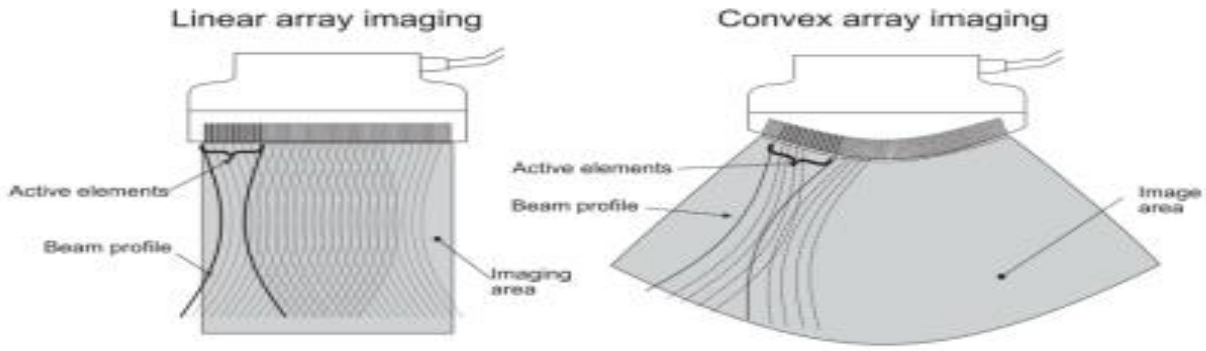

Figure 5.10. linear and convex probes

Phased array probes (see figure 11) have the same geometry as linear arrays, but are much smaller, sweeping an area much larger than the size of the aperture, with all elements used for both transmitting and receiving.

Scanning is done by steering the transmitted beam, it is used in cardiology, where there is only a small “acoustic window” between the ribs and the lungs. Scanning is performed by directing the transmitted beam; it is used in cardiology.

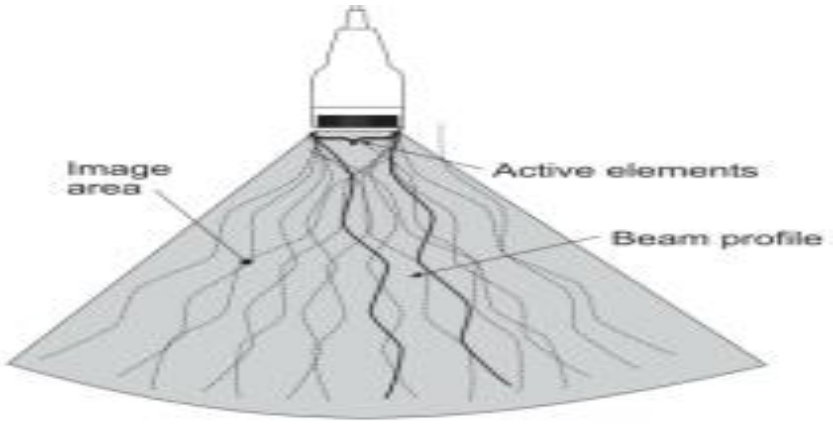

Figure 5.11. phased array probes

6. Modes of imaging

A variety of ultrasound imaging modes are available to the physician. The main ultrasound modes are listed below [24]:

6.1.A-mode (Amplitude)

It is the oldest and simplest mode of pulse echography; this technique is based on the reception of ultrasound wave and the reception of the echo in a single line of propagation. Figure 12 shows an example of an amplitude of echo signal in A-mode.

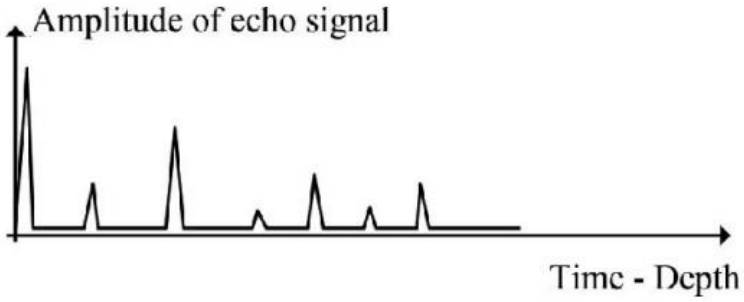

Figure 5.12. Signal in A-mode

6.2. M-mode(motion)

The M-mode or TM-mode (time motion) is used to visualize the motion of structure moving, it helps to show the evolution of a signal over the time. Figure 13 shows an acquisition with M-mode.

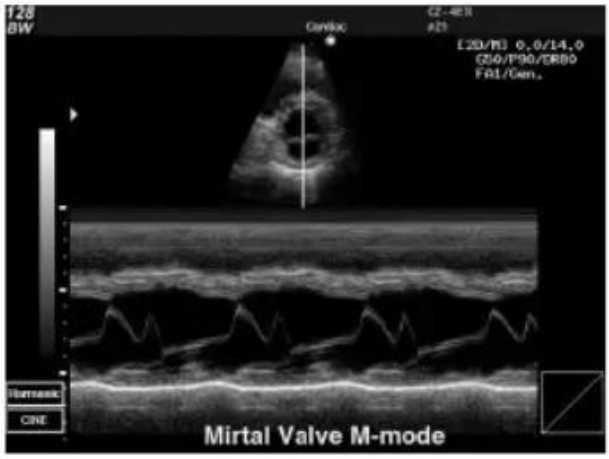

Figure 5.13. M-mode

6.3. B-mode (Brightness)

The B mode, derived from A mode, is currently the most widely used. The imaging plane is formed by the direction of wave propagation and the direction of transducer movement. Figure 14 shows an example of B-mode acquisition.

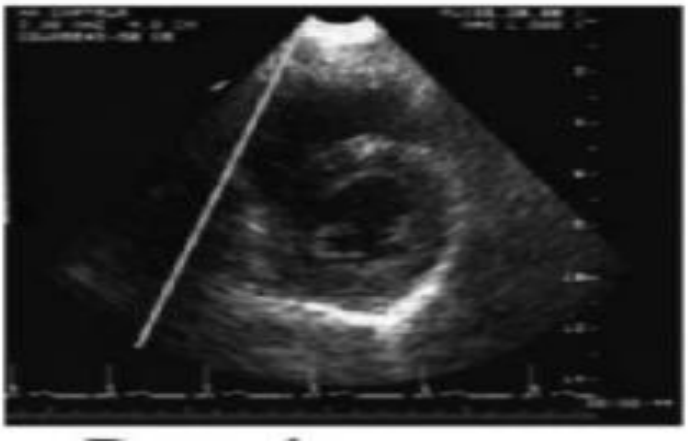

Figure 5.14. B-mode

6.4. C-mode (Constant depth)

It provides 2D images at constant depth, the plane is created by the motion of the transducer at a constant distance along the propagation path.

6.5. Continuous-wave Doppler (CWD)

It is based on the continuous emission/reception of ultrasound waves by transducer. It gives information about the distribution of the velocities in areas traversed by the beam. Figure 15 shows a signal in continuous wave doppler.

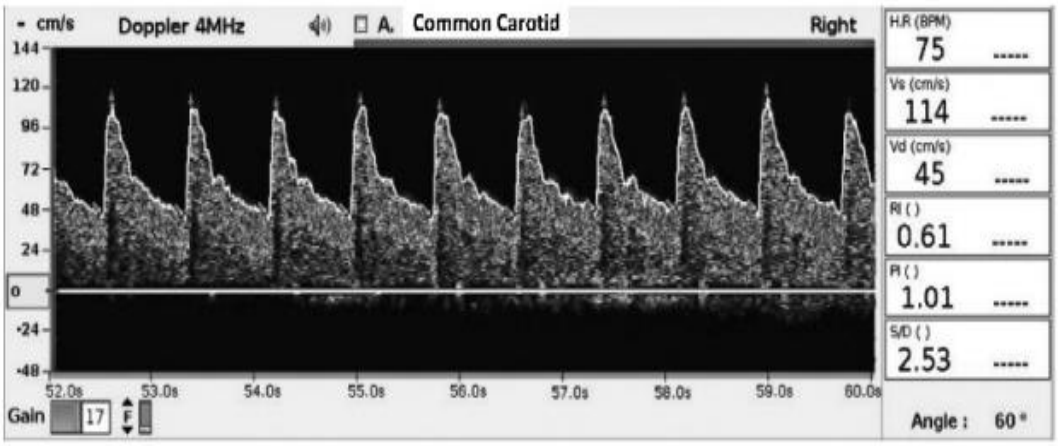

Figure 5.15. Continuous wave Doppler

6.6. Pulsed Doppler

Pulsed Doppler offer to user the ability to control the size, depth and location of the sample volume contrary to continuous wave doppler. It can also analyze data from several pulses in the same direction. Figure 16 shows the schematic representation of the location of the sample volume for pulsed Doppler imaging.

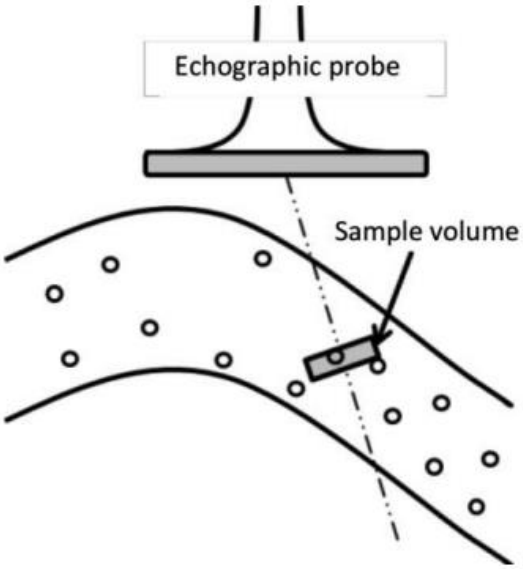

Figure 5.16. Pulsed Doppler imaging

7. Biological effects of ultrasound

Ultrasound is a form of energy emission, a sound wave alternating positive and negative pressures, so it has the potential to produce biological effects on tissues and cells that cause harm to the fetus.

There are two major mechanisms that affect embryonic and fetal tissues: thermal and mechanical. Thermal effects result from the passage of ultrasound waveforms, with transformation of acoustic energy into heat. Thermal effects depend on the type of tissue exposed, the duration of exposure, ultrasound mode, and the distance between tissue and emission source.

Mechanical effects result from the alternating pressures that are generated.

The main effect occurs because of the interaction between ultrasound waves and gas bubbles present in tissues. Sound waves generate movement in and around gas bubbles, which can affect the surrounding tissues and lead to cavitation. Ultrasonic cavitation causes mechanical damage, but it can also generate free radicals and other chemicals capable of damaging cell DNA.

8. Conclusion

Ultrasonography has become increasingly prevalent in medical practice and hospitals worldwide over the last few decades. However, the ongoing need for rapid advancements and equipment improvements in this field is essential for enhancing image quality, enabling expert diagnoses, and developing various components of ultrasound machines.

In this chapter, we have delved into the role of ultrasound, explored the mechanics of echo machines, and discussed the principles of echography. This technology plays a crucial role in assisting doctors, particularly in gynecology, for the detection and treatment of anomalies.

Chapter 6
Magnetic resonance imaging

Authored by

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1. Introduction

Magnetic Resonance Imaging (MRI) is a highly specialized medical imaging technique. MRIs are particularly well-suited for imaging non-bony or soft tissues within the body. It is often requested by doctors due to its high accuracy compared to other techniques, as it does not use damaging ionizing radiation like X-rays. MRI is commonly used for examining the spinal cord, nerves, muscles, and ligaments. It is also employed in diagnosing aneurysms and tumors and is frequently used for imaging knee and shoulder injuries. Moreover, MRI can differentiate between white matter and grey matter in the brain. Since MRI does not use X-rays or other forms of radiation, it is the preferred imaging modality when frequent imaging is required for diagnosis or therapy, especially in cases involving the brain.

In this chapter, we describe the MRI scan by discussing its various components. We provide an overview of the physical basis of NMR, MRI sequences, and MRI reconstruction. Finally, we conclude by addressing the safety considerations associated with MRI scans.

2. Definition

MRI scans (Figure 1) are non-invasive imaging technologies used to investigate the anatomy and function of the body in both health and disease, without the use of damaging ionizing radiation. They are often employed for disease detection, diagnosis, and treatment monitoring. MRI is based on sophisticated technology that excites and detects changes in protons found in the water that make up living tissues.

Figure 6.1. MRI scan

3. MRI component

MRI (Magnetic Resonance Imaging) systems consist of several key components that work together to produce detailed images of the internal structures of the body. Here are some of the main components of an MRI system:

3.1.Magnet

The letter 'M' in MRI stands for 'magnetic,' highlighting the presence of a large magnet responsible for creating a strong magnetic field (B0), which is necessary for aligning the hydrogen nuclei [25]. The characteristics of the magnet play a crucial role in determining the quality of the field, including [26]:

- Magnetic field strength and stability.
- Type of magnet used (resistive, superconducting, permanent).
- Field homogeneity.

Currently, manufacturers predominantly opt for superconducting magnets due to their superior characteristics, resulting in an extremely high magnetic field [27]. Figure 2 illustrates the different types of magnets commonly used in MRI.

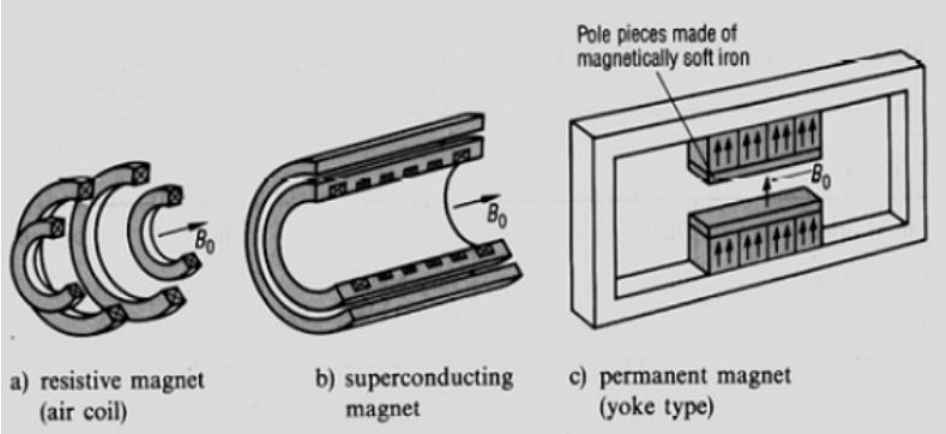
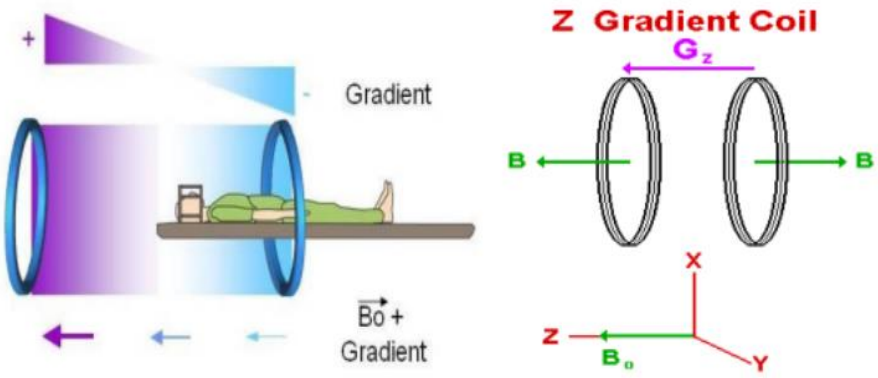


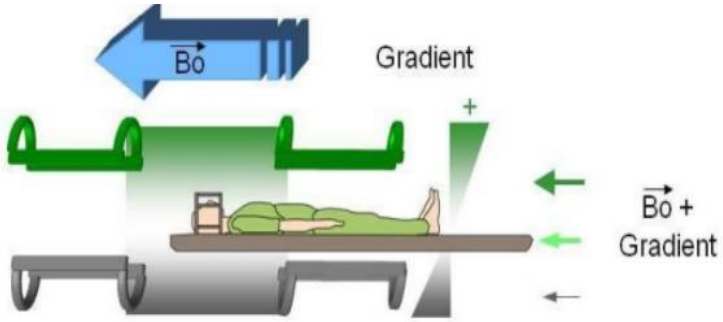
Figure 6.2. a) resistive magnet, b) superconducting magnet, c) permanent magnet

3.2.Gradient Coils

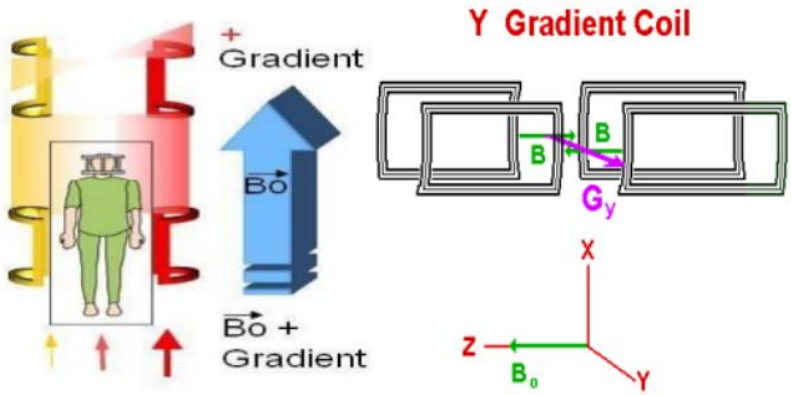
Gradient coils play a crucial role in forming the MRI image by linking distance and resonance frequency. These coils generate a gradient magnetic field (B0) that is essential for spatial signal encoding. They are responsible for creating a linear variation in the magnetic field strength along different spatial directions, namely, x, y, and z. Pairs of coils are strategically placed in each spatial direction [27]. Figure 3 illustrates the application of B0 gradient coils along the Z-axis, Y-axis, and X-axis.



(a)



(b)



(c)

Figure 6.3. Gradient Coils for: a) Z-Axis, b), X axis, and c)Y axis

3.3. Radio Frequency (RF) Coils

Radiofrequency coils, also known as antennas, serve multiple vital roles in MRI equipment. These RF coils are responsible for producing the radiofrequency magnetic field (B1) and receiving transverse magnetization. Their primary function is to transform an electrical signal emitted by a radiofrequency power transmitter into a rotating magnetic field, leading to the manipulation of nuclear magnetization. Subsequently, they convert the radiofrequency field

generated by the precession of nuclear magnetization into an electrical signal, which is then transmitted to the receiver. This signal is ultimately converted into digital data by the receiver and processed by the imaging console [28].

RF coils are typically categorized into three types:

• Transmit-receive coils: These coils transmit magnetic fields and receive signals.

• Transmission coils: These coils are responsible for generating magnetic fields.

• Receive coils: These coils detect spin relaxation signals within the subject being imaged.

Figure 4 illustrates some of the most commonly used radiofrequency antennas (coils) in MRI, including the whole-body coil, knee coil, head coil, abdominal coil, and spine coil.

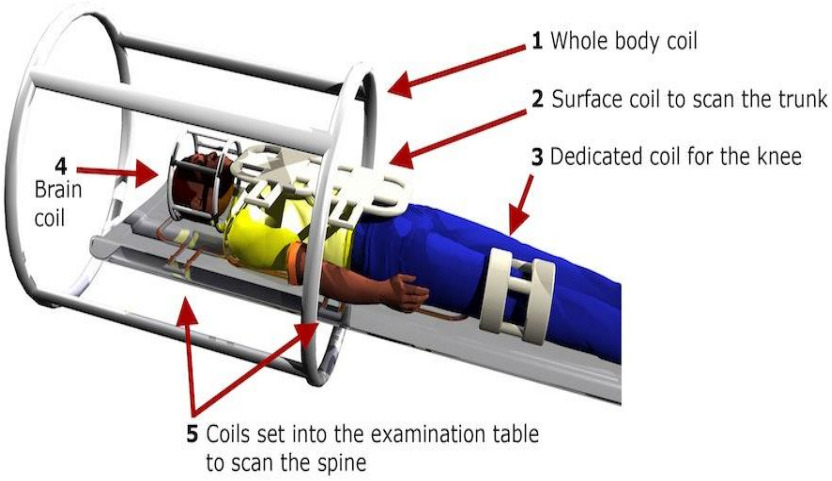


Figure 6.4. Radio Frequency coils (RF)

3.4.Shielding

The purpose of shielding in MRI is to contain the magnetic field generated by the machine and protect it from external fields that could disrupt the imaging process. There are two types of shielding configurations [29]:

3.4.1 RF shielding

This type of shielding is established within a shielded room. The walls, floor, and ceiling of the room are covered with copper plates designed to absorb external electromagnetic waves that might interfere with the MRI's magnetic field. Additionally, these copper plates prevent RF waves from escaping the room and causing interference with nearby electronics.

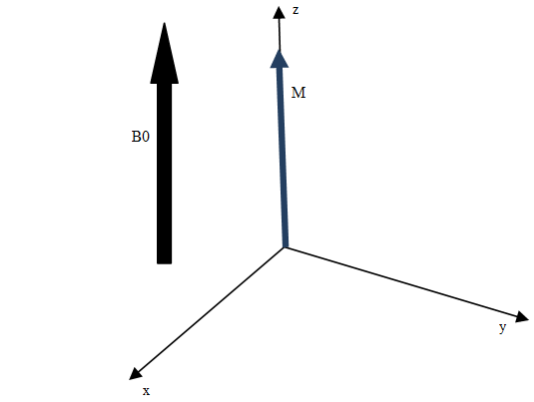
3.4.2 Magnetic Field Shielding

The primary function of magnetic field shielding is to control the extent of the magnetic field lines emanating from the MRI magnet. This includes containing the 0.5mT line within the scanning room. Depending on the MRI machine, there are two types of magnetic field shielding:

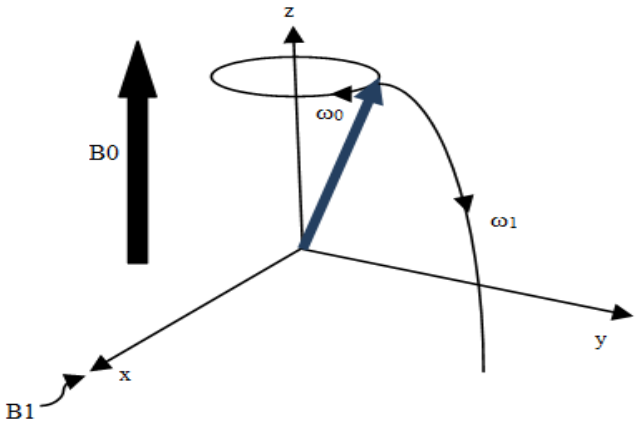
- **Passive Shielding:** This involves the use of steel or mild iron joists surrounding the magnet.
- **Active Shielding:** Active shielding employs an inverted metal winding placed at both ends of the main Bo field winding.

4. Physical basis of NMR

MRI is a medical imaging technique that uses strong magnetic fields and radio waves to generate detailed images of the internal structures of the human body.
MRI relies on the principles of nuclear magnetic resonance (NMR). In simple terms, NMR is the phenomenon where certain atomic nuclei (e.g., hydrogen nuclei in water molecules) absorb and re-emit radiofrequency energy when placed in a magnetic field. When a patient is placed in the MRI scanner, the hydrogen nuclei within their body align with the strong magnetic field (B0) (figure 5a). These nuclei have magnetic moments, which are like tiny magnets. When the radiofrequency pulse is applied, it disrupts this alignment (figure 5b), and when the pulse is turned off, the nuclei return to their aligned state, releasing energy in the form of NMR signals. Magnetization in MRI refers to the net magnetic moment of the hydrogen nuclei within a given volume of tissue.



a)



b)

Figure 6.5. a) magnetization parallel to the strong magnetic field (B0), b) precession of magnetization around field B1

In NMR, the interaction between atomic nuclei and a strong magnetic field (B0) is quantified. Only certain frequencies, known as resonance or Larmor frequencies, allow for this interaction. The specific resonance frequency depends on the type of atomic nucleus being observed and the chemical environment surrounding that nucleus.
NMR is based on the detection of magnetization arising from the spins of atomic nuclei. Each atomic nucleus with a non-zero nuclear spin acts like a tiny magnet, and its magnetic moment interacts with the external magnetic field (B0).
Magnetization from all the nuclear spins in a sample collectively forms what's referred to as macroscopic magnetization. When placed in the presence of magnetic fields B0 and B1 (a radiofrequency field), the magnetization switches to the transverse plane and follows a spherical trajectory.
The magnetization is oriented in the plane transverse to the direction of the B0 magnetic field. The precession of magnetization in this plane occurs at the Larmor frequency, which is specific to the type of nucleus and the strength of the magnetic field. When the B1 RF field (radiofrequency field) is disabled, the magnetization gradually returns to its initial value. This process is known as relaxation and involves the magnetization following a reverse trajectory.

4.1. Relaxation phenomenon

Relaxation is a phenomenon that occurs when tissue magnetization returns to its initial state, which produces an emission of energy in the form of radio waves that are recorded during an MRI. There are Two different mechanisms exist for relaxation: longitudinal relaxation, which regenerates the longitudinal component, and transverse relaxation, which decreases the transverse magnetization [30].

4.1.1 Longitudinal relaxation (spin-lattice)

The longitudinal relaxation, also known as spin-lattice relaxation, is based on a thermal energy exchange between protons and the lattice. It is measured by the relaxation time T1. This time indicates how long it takes for the longitudinal magnetization Mz to return to its original value. In Figure 6, this phenomenon is illustrated, with T1 time representing the duration required for the value of Mz to reach 63% of its initial value [31].

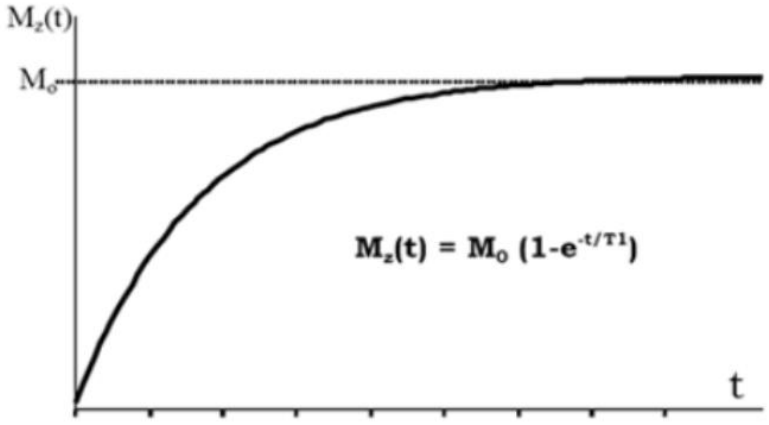


Figure 6.6. Longitudinal relaxation

4.1.2 Transverse relaxation (spin-spin)

The transverse relaxation is caused by the magnetic interaction between the spins, which causes a phase shift between them, resulting in an exponential decrease of the transverse magnetization Mxy (Figure 7).

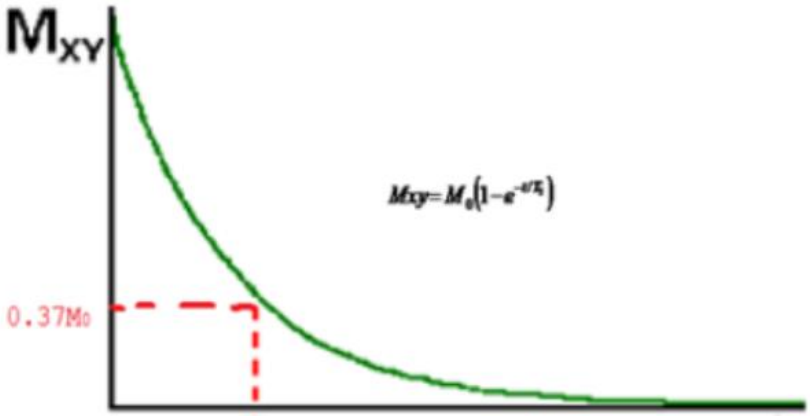


Figure 6.7. Transverse relaxation

The return of the magnetization to its equilibrium state generates a current in the receiving coil, producing a signal called the precession signal or FID (Free Induction Decay) signal in English. The FID signal is a sinusoidal wave that is damped by an exponential decay. Figure 8 presents the NMR signal generated.

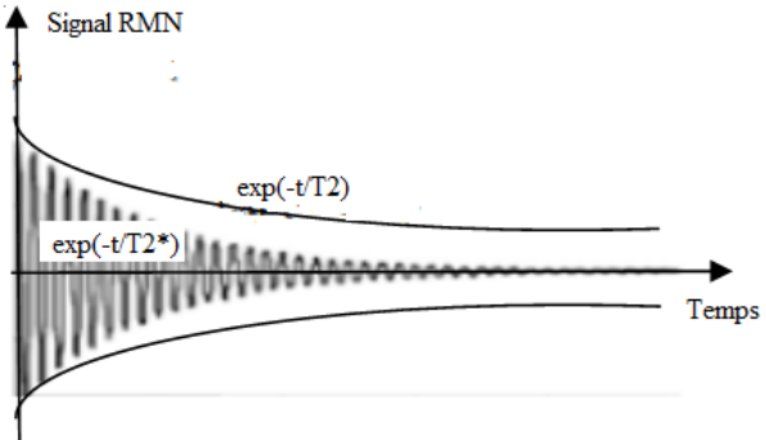


Figure 6.8. NMR signal

During the acquisition of the MR images, several sequences can be selected according to the need of the manipulators, depending on the examination.

5. MRI sequences

MRI sequences can be divided into two broad categories depending on the type echo detected: Spin Echo and gradient echo.

5.1. Spin Echo

Spin Echo sequences are fundamental in MRI because they can produce T1, T2, and proton density weighted images. This sequence involves two successive pulses: first, an excitation pulse that tilts the spins by 90°, followed by a 180° rephasing pulse applied at a time equal to TE/2 (see Figure 9). This sequence repeats at regular intervals known as the Repetition Time (TR).

Echo Time (TE): TE is the time interval between the excitation RF wave and the read time.

Repetition Time (TR): TR represents the time elapsed between two successive 90° excitation pulses. With each repetition, an MR image line is generated using different phase encodings.

The quality of the NMR image primarily depends on the choice of the TR (repetition time). It must be carefully selected to strike the best balance between achieving good resolution and minimizing scan time.

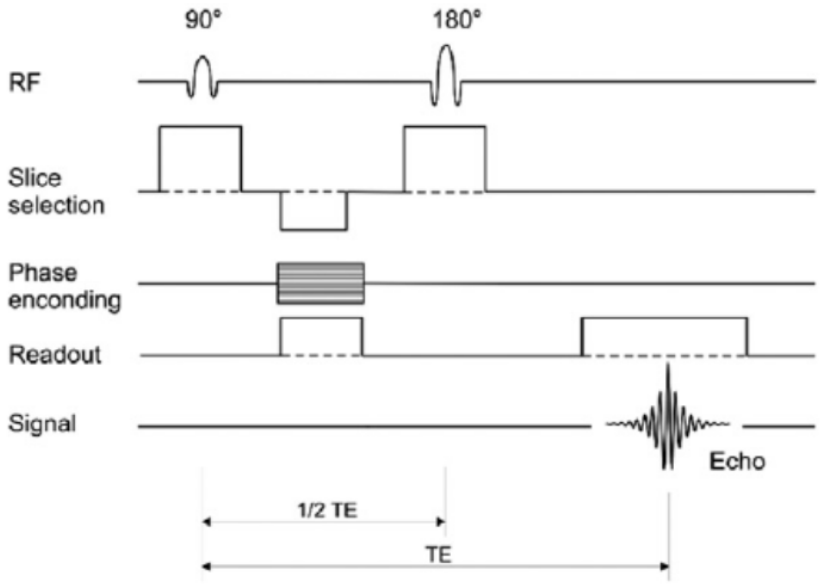


Figure 6.9. Spin echo sequence

5.2. Gradient echo

Gradient echo sequence enables rapid image acquisition compared to the spin echo sequence because it lacks the second π (180°) pulse, which restricts the flip angle of the macroscopic magnetization vector to less than 90° (as shown in Figure 10). Gradient echoes are generated by applying a bipolar reading gradient in the frequency-coded direction. The first gradient lobe accelerates the phase shift of the transverse magnetization, while the second anti-lobe rephases the spins.

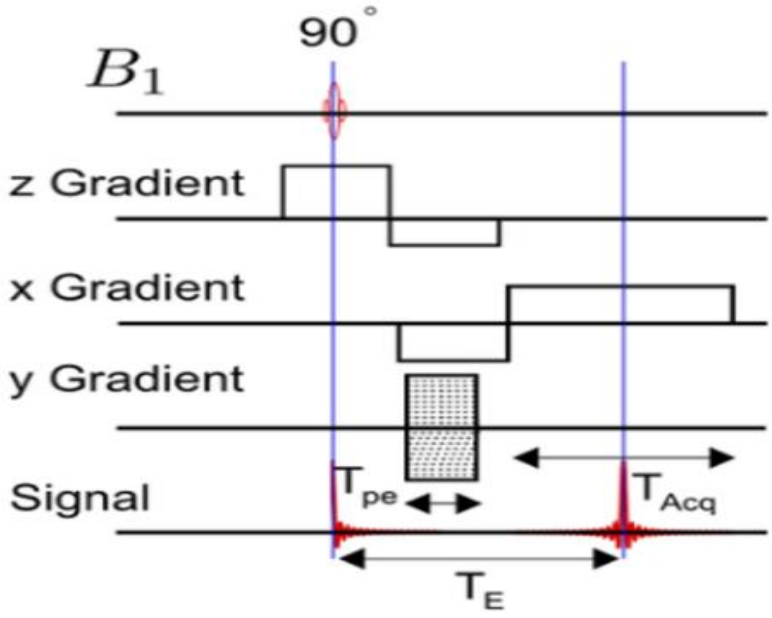


Figure 6.10. Gradient echo sequence

After grasping the working principle of MRI, we delve into one of the most crucial steps: image reconstruction.

6. Image reconstruction

In order to generate an MRI image, the echo signal must be located and spatially encoded at any point in the image. The spatial location of the signal is based on two fundamental concepts:

1- The use of a physical tool: magnetic field gradients.
2- The use of a mathematical tool: the Fourier transform.

To create a (2D) slice image in MRI, spatial encoding is employed, encompassing slice selection, phase encoding, and frequency encoding gradients. Spatial encoding necessitates the utilization of magnetic field gradients, denoted as Gx, Gy, and Gz, applied along the x, y, and z axes, respectively. Consequently, the magnetic field's intensity varies systematically along the gradient application axis.

Prior to image reconstruction, when selecting a specific slice in MRI, it becomes imperative to adjust the rotational frequency of protons to match that of a radiofrequency wave, achieving resonance. This necessitates the simultaneous application of a slice selection magnetic field gradient (GS) and excitation radiofrequency pulses in a specific direction.

6.1.Phase encoding

Phase encoding involves the application of an additional magnetic field gradient known as the Gy phase encoding gradient. This gradient serves the purpose of distinguishing signal sources along the vertical y-direction axis prior to the reading phase. Consequently, it alters the rotation frequency of spins at different positions along this gradient.

When the signal is being recorded, this phase encoding gradient is deactivated. The spins then regain their initial frequency but with varying phases (as depicted in Figure 11). Through phase encoding, the object under examination can be segmented into multiple lines along the vertical direction. Each of these lines captures an NMR signal with a distinct phase.

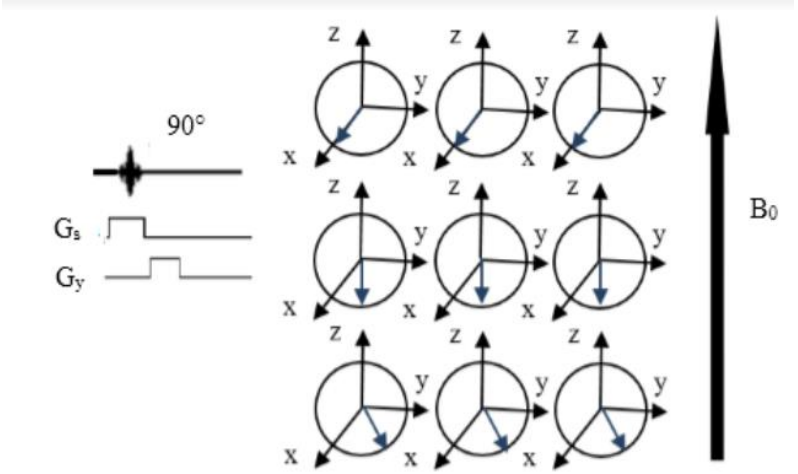


Figure 6.11. Phase encoding

6.2.Frequency encoding

The RF inputs consist of the RF pulse accompanied by magnetic field gradients, while the output represents the recorded NMR signal over time. To select a specific slice for imaging within the patient, the sequence necessitates the activation of a Gz field gradient in conjunction with the RF pulse.

Once the desired slice is selected, Gy is activated along the y-axis, and the frequency encoding gradient is engaged along the x-axis when Gy is subsequently deactivated. The NMR signal is recorded during the application of the Gx (or Gf) gradient, as illustrated in Figure 12.

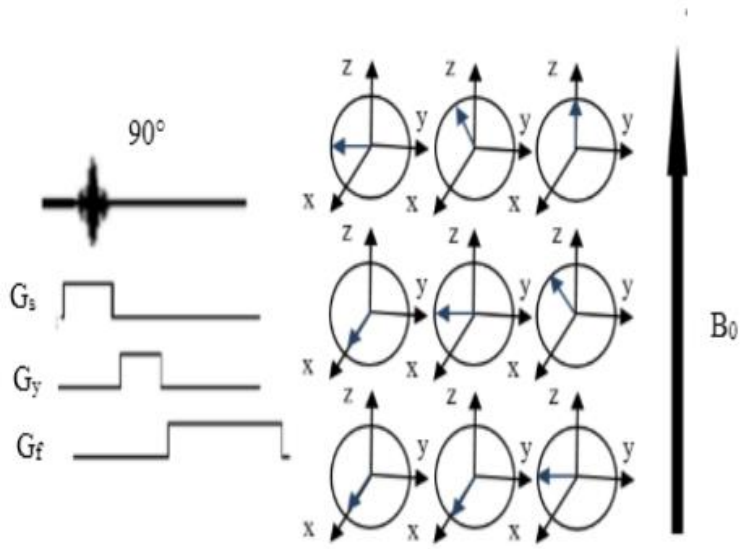


Figure 6.12 Frequency encoding

Spatial location refers to the position of different points within the imaged object. These points represent structures or tissues within the object. k-space is a representation of spatial frequencies present in an image. MRI data is collected in k-space, and image reconstruction involves converting this frequency information back into spatial coordinates to create the final MRI image, allowing us to visualize the spatial distribution of tissues and structures within the imaged object

6.3.K space

To decompose an image into two dimensions, a two-dimensional Fourier transform is applied according to the following steps:

1- The first step of the two-dimensional Fourier transform involves the application of a one-dimensional Fourier transform in the first direction.
2- The second step of the 2D Fourier transform is to apply another one-dimensional Fourier transform according to the second direction to the result obtained in the first stage.

The Fourier plane is the result of the two-dimensional Fourier transformation of an image, which is graphically represented by Figure 13, The signals recorded during a 2D MRI sequence are represented in a space called k-space, which is like a two-dimensional map describing the frequencies present in the data. The k-space is based on the mathematical principles of the Fourier transform. It is defined by two directions, kx, and ky.

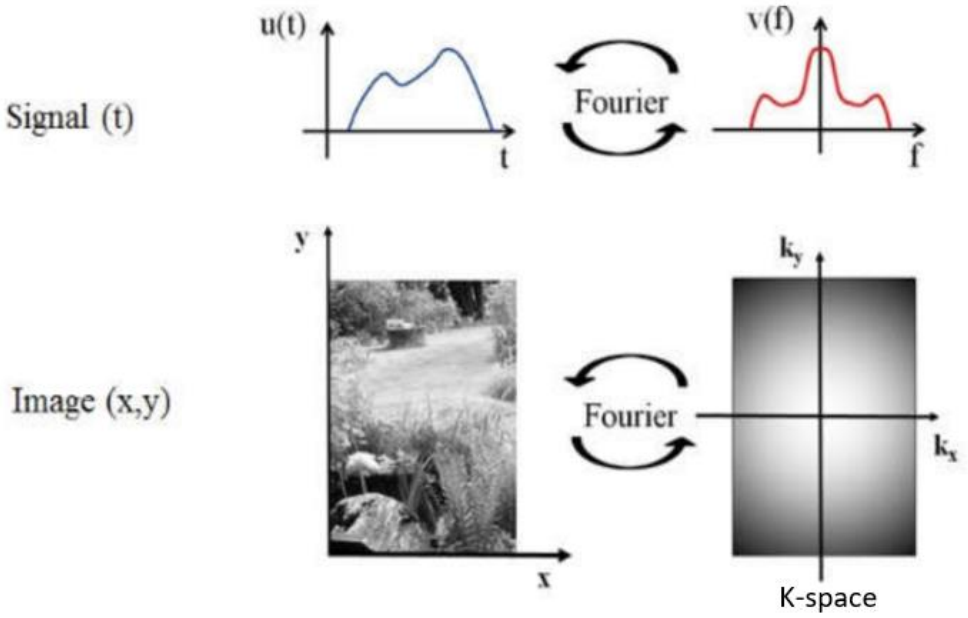


Figure 6.13. K-space and image reconstruction

7. MRI risks

Although MRI does not emit ionizing radiation like X-ray and CT imaging, it does generate a strong magnetic field. This magnetic field extends beyond the MRI machine and can exert powerful forces on objects containing iron, some steel, and other magnetizable materials. Patients should inform their physicians of any medical implants or metal objects in their body prior to an MRI scan to ensure safety.

People should not enter an MRI machine if they have certain implants, especially those containing iron. Examples of such implants include pacemakers, vagus nerve stimulators, implantable cardioverter-defibrillators, loop recorders, insulin pumps, cochlear implants, deep brain stimulators, and capsules used in capsule endoscopy. Safety precautions must be taken into consideration."

Patients undergoing MRI scans may require special ear protection due to the loud noises commonly referred to as clicking and beeping that occur during the procedure."

While there is no definitive evidence of harmful effects on the fetus, it is generally recommended that MRI scans be avoided as a precaution, especially during the first trimester of pregnancy when the organs of the fetus are forming. This includes both routine MRI and the use of contrast agents. The decision to perform an MRI during pregnancy should be carefully considered and discussed between the patient and their healthcare provider.

8. Conclusion

In the ever-evolving field of medicine, new techniques are continuously emerging to enhance medical modalities. MRI, in particular, has not only benefited from synergies with other modalities, such as PET-MRI or TEMP-MRI, but it has also seen significant progress within its own framework. These advancements often involve the application of mathematical equations to facilitate rapid and precise disease diagnosis, aiding healthcare professionals in their decision-making processes.

MRI, although renowned for its ability to provide detailed anatomical information, has historically been associated with longer scan times compared to modalities like CT scans. To address this challenge, powerful systems have been developed, aimed at minimizing scan time while preserving the valuable diagnostic information generated by a single MR scan. These techniques continue to evolve and develop, ultimately contributing to improved patient care in modern medicine

General conclusion

In closing, we have embarked on a captivating journey through the intricate world of medical imaging. Throughout this coursebook, we have delved into the foundational principles of various imaging modalities, from the penetrating power of X-rays to the gentle precision of ultrasound, from the magnetic resonance of atoms to the radioactive tracers of nuclear medicine. We have explored the physics, mathematics, and engineering that underpin these technologies, gaining a deep understanding of how they generate the images that guide clinical decisions.

It is essential to acknowledge that with great power comes great responsibility. The ethical considerations and safety concerns surrounding medical imaging cannot be understated. It is imperative that we remain vigilant in our efforts to minimize radiation exposure, protect patient privacy, and ensure the ethical use of imaging technologies in healthcare.

In closing, this course aimed to serve as your guide, providing a comprehensive understanding of medical imaging and its role in healthcare. We hope it has ignited your curiosity, expanded your knowledge, and deepened your appreciation for this dynamic and evolving field. As you continue your journey in medicine, research, or technology, may the insights gained from this course continue to inspire and inform your path.

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