

# The Physics of Radiology and Imaging



**K Thayalan**

*Foreword*  
**R Ravichandran**



# THE PHYSICS OF RADIOLOGY AND IMAGING

(Targeted to postgraduate students of medical physics and radiology,  
appearing for MSc, DMRD, MD, DNB and FRCR examinations)

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*Foreword*

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**JAYPEE BROTHERS MEDICAL PUBLISHERS (P) LTD**

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Suite 835, Philadelphia, PA 19106, USA  
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Phone: +977-9741283608  
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Website: [www.jaypeebrothers.com](http://www.jaypeebrothers.com)  
Website: [www.jaypeedigital.com](http://www.jaypeedigital.com)

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***The Physics of Radiology and Imaging***

**First Edition: 2014**

**ISBN 978-93-5152-171-6**

**Printed at**

Jaypee-Highlights Medical Publishers Inc.  
City of Knowledge, Bld. 237, Clayton  
Panama City, Panama  
Phone: +1 507-301-0496  
Fax: +1 507-301-0499  
Email: [cservice@jphmedical.com](mailto:cservice@jphmedical.com)

Jaypee Brothers Medical Publishers (P) Ltd  
17/1-B Babar Road, Block-B, Shaymali  
Mohammadpur, Dhaka-1207  
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# FOREWORD

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## Dr. Ramamoorthy Ravichandran

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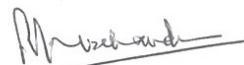
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It is a great privilege to me for writing the foreword to this book 'The Physics of Radiology and Imaging' written by Professor Dr K Thayalan, Former Professor and Head of the Department of Radiology Physics, Barnard Institute of Radiology, Madras Medical College, Chennai, India. Dr K Thayalan has vast experience in teaching Medical Physics of Radiology, Radiation Oncology and Nuclear Medicine. He authored many textbooks earlier relating to radiological physics and radiological safety; they are being considered as reference books in many institutions and universities.

There has been a long-felt need for comprehensive textbook in the field of physics applied to medical imaging, for medical postgraduates in radiology, medical physicists and technologists, and this book meets the requirement.

This book is a quintessence of basic and applied physics thoroughly explaining principles of physics of computed and digital radiography, image intensifier fluoroscopy, mammography, ultrasound, computed tomography and magnetic resonance imaging. All the chapters are well laid out with good explanations and illustrations. Fundamentals have been brought out in a unique way for easy assimilation into memory of teachers and students. Explanations on image resolution, contrast, artifacts on different modalities are very vivid and nice for good understanding of the complex details.

It was a real pleasure for me to go through the entire contents to write this foreword and readers will agree with me that all details looked for relating to physics of imaging is available in this book. I am sure that this book will be referred globally in the field of medical imaging, helping the residents of FRCR (UK) and MD postgraduates of different universities in India.



R Ravichandran

# PREFACE

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Radiation has been used in medicine since from the discovery of X-rays in 1895 by the German physicist WC Roentgen. Over the period, its application has grown enormously and it is used today in the form of radiography, fluoroscopy, mammography and computed tomography. Non-radiation tools are also competing in medicine in the form of ultrasound and magnetic resonance imaging. These tools not only provide early differential diagnosis of the disease but also improve the accuracy of clinical diagnosis. The uniqueness of the above tools is that they all work on the basis of physics principle.

Understanding the physics of the above instruments is very much essential, for those connected with radiological sciences. It helps not only the education, but also the equipment selection, its optimal use, maintenance, and safety. Hence, an attempt is made to explain the physical principle, instrumentation, function, its application and limitations in the form of a single book. Attempt is also made to incorporate nuclear imaging and radiological safety in the same book. Large numbers of figures and tables are incorporated wherever it is necessary, for better understanding of the concept. This book is intended for postgraduate students of medical physics, diagnostic radiology, Diplomate National Board (DNB), and FRCR. This is the first book of its kind from an Indian author, giving single solution for the entire range of radiology and imaging equipment.

I am very proud and happy to come out with this book, incorporating my three decades of experience in radiological/medical physics teaching. I am very happy and thankful to Dr R Ravichandran for writing the foreword to this book. I am also thankful to M/s Jaypee Brothers Medical Publishers (P) Ltd., New Delhi, for publishing this book as usual in a neat and elegant manner. Constructive comments are invited from the readers for the future betterment of the book.

I am very much thankful to my wife Tamilselvi, son Parthiban and daughter Kayal Vizhi for their support and cooperation during the book writing process.

I thank and acknowledge Dr Kamakshi Memorial Hospital, Chennai, especially the medical physics division for the support and assistance.

**K Thayalan**

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# 1

# Fundamental Concepts

## MATTER AND ENERGY

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Physics is a science dealing with nature. It is concerned with the study of two concepts, matter and energy, and how they interact with each other. Matter is one, which occupies space, and it is made up of molecules or atoms, e.g. gold, wood, water and air. Matter exists in solid, liquid, gas, liquid crystal, and plasma state. Matter can be converted from one form to another by physical or chemical means, e.g. melted ice converts from solid to liquid form by physical process and burning of wood into ash is a chemical process.

Energy is the ability to do work, it has several forms, and it can be converted from one form to another, e.g. human body converts chemical energy (food) into kinetic energy (work). Law of conservation of energy states that energy can neither be created nor destroyed, and the total energy in the universe is constant. This law holds good for all forms of energy.

In general, physicist studies the behavior of matter and energy under different physical conditions.

## MEASUREMENT AND UNITS

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To study, the matter and energy and their various properties, measurements of physical quantities, such as length, mass, and time are required. Physical quantity is measured accurately in terms of its own standard, e.g. distance is measured in meter, mass in kilogram, and time in second. Therefore, unit is a quantity adopted as a standard of measurement in terms of which similar quantities can be measured. The units which are independent of one another and having their own standard (base) are called fundamental units, e.g. kilogram, meter and second. The units, which are not having their own standard (base) and obtained

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from the fundamental units are called derived units, e.g. area-meter<sup>2</sup>, velocity-meter per second, and density-kilogram per meter<sup>3</sup>.

One meter is the distance traveled by light (Krypton-86) in 1/299,792,468 second. One kilogram is the mass of 1000 cm<sup>3</sup> of water at 4°C. The second is measured by an atomic clock and is based on the vibration of atoms of cesium.

### SI UNITS

In 1960, a new system of units called Systems International d'units (SI Units) was introduced. The SI system is superior to all other systems and more convenient in practice and is used throughout the world. There are 7 fundamental units and 2 supplementary units in the SI system as shown in the Table 1.1.

**TABLE 1.1 SI system of units**

Physical quantity	Unit	Symbol
Length	meter	m
Mass	kilogram	kg
Time	second	s
Electric current	ampere	A
Temperature	kelvin	K
Luminous intensity	candela	cd
Amount of substance	mole	mol
Plane angle	radian	rad
Solid angle	steradian	sr

### Conventions for SI Units

- i. When the unit is named after a scientist, it should not be written in a capital initial letter, e.g. newton, ampere. The symbol of the unit is expressed in capital letters, e.g. N for Newton.
- ii. The symbol of all other units should be written with small letters, e.g. 'm' for meter.
- iii. Only singular form of the unit is to be used, e.g. 500 meters is written as 500 m. No full stops or punctuation marks should be used at the end of the symbol.
- iv. Space is to be left between the numerical and symbol, e.g. 20 s and not as 20s.
- v. Mathematical indices notation should be used than slash sign (/), e.g. meters per second should be written as ms<sup>-1</sup> not m/s.

**TABLE 1.2** Prefixes used with SI units

Prefix	Symbol	Factor
tera	T	$10^{12}$
giga	G	$10^9$
mega	M	$10^6$
kilo	k	$10^3$
deci	d	$10^{-1}$
centi	c	$10^{-2}$
milli	m	$10^{-3}$
micro	$\mu$	$10^{-6}$
nano	n	$10^{-9}$
pico	p	$10^{-12}$

vi. In the temperature unit Kelvin no degree sign is used, e.g. 273 K and not as 273° K.

### Prefixes

Though the SI units are a coherent system, they are found to be either too large or low in practice, e.g. the activity of an isotope for bone scan is expressed in billions of becquerel's. Hence, prefixes are used to overcome the above difficulty, as shown in Table 1.2. These prefixes are conveniently used to describe very large or small physical quantities. In radiation physics, giga becquerel (GBq), kilovolt (kV), centi gray (cGy), milli ampere (mA), and nanometer (nm) are commonly used.

## DENSITY, MOLE, PRESSURE, AND GAS LAWS

### DENSITY

The density of a body ( $\rho$ ) is defined as the ratio of its mass (m) and volume (v) and its unit is  $\text{kg m}^{-3}$ . The density of a body is same, if it is made up of identical material. If its composition is changed, its density will vary.

$$\rho = \frac{m}{v}$$

The relative density or specific gravity of a substance is the ratio between its density with that of water.

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### MOLE

The amount of matter in a body is expressed by the number of elementary particles (atoms or molecules) it contains and its unit is mole. One mole of matter contains  $6.022 \times 10^{23}$  elementary particles, and it is known as Avogadro's number.

### PRESSURE

The total force acting on a liquid surface is called thrust. The pressure ( $p$ ) is defined as the force ( $F$ ) per unit area ( $A$ ) and its unit is  $\text{Nm}^{-2}$  or pascal (Pa). The atmospheric pressure is about  $1.01 \times 10^5$  Pa. The pressure is caused by the weight of material pressing on its surface. It may be also due to collisions of atoms or molecules of a gas within a container. The pressure of a liquid at rest is always perpendicular to the surface in contact with it. The pressure at a point within a liquid is directly proportional to the depth of the point from the free surface, density, and acceleration due gravity.

### GAS LAWS

Boyle's law states that the volume ( $V$ ) of a given mass of gas is inversely proportional to its pressure ( $P$ ), at constant temperature. Charles's law states that volume of a given mass of gas, at constant pressure, is proportional to its temperature ( $T$ ). The above two laws can be combined and stated as follows:

$$PV/T = \text{constant}$$

This is known as the perfect gas equation.

### MECHANICS

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#### VELOCITY AND ACCELERATION

Displacement ( $d$ ) is defined as the shortest distance between the initial and final positions of a body. The velocity ( $v$ ) of a moving body is the rate of change of displacement of the body in a particular direction and its unit is  $\text{ms}^{-1}$ . The magnitude of velocity is called speed, which is a scalar quantity. Velocity is a measure of how fast the matter is moving or rate of change of its position with time. It is given by the relation;

$$v = d/t, \text{ where } d \text{ is the displacement in } t \text{ seconds.}$$

Acceleration ( $a$ ) is defined as the rate of change of velocity and its unit is  $\text{ms}^{-2}$ . It is a measure of how quickly or slowly the velocity

is changing. If the velocity is constant, the acceleration is zero. It is given by the relation

$$a = (v_f - v_0)/t$$

where,  $v_0$  is the initial velocity and  $v_f$  is the final velocity, that undergone during the time interval  $t$ .

## SCALAR AND VECTOR QUANTITIES

All physical quantities can be classified into two broad categories, namely, scalar and vector quantities. Quantities that have only magnitude and no direction are called scalar quantities, e.g. length, mass, time, etc. Quantities that have magnitude as well as direction are called vector quantities, e.g. displacement, velocity, force, etc.

A vector quantity is usually represented graphically by an arrow ( $\rightarrow$ ), whose length is proportional to the magnitude of the vector. In an equation, vector quantity is represented by bold letters, e.g.  $\mathbf{F} = m\mathbf{a}$ , where, force and acceleration are vectors and mass is a scalar quantity.

## FORCE

Force is the influence that changes the state of rest or uniform motion of the body along a straight line. If a force  $F$  acts on a body of mass  $m$ , and produces an acceleration  $a$ , then  $F = m \times a$ . Hence, the force acting on the body is equal to the product of mass of the body and the acceleration produced by the force on the body.

The SI unit of force is newton and it is denoted by the letter N. One newton is the force acting on a body of mass one kilogram producing an acceleration of one  $ms^{-2}$  in its direction.

## WORK

If a force acts on a body and the point of application of the force moves, then work is said to be done by the force. If the force  $F$  moves a body through a distance  $s$  in its direction, then the work done by the force is given by  $W = F \times s$ . The displacement does not always take place in the direction of force. If the direction of displacement  $s$  is inclined to  $F$  at an angle of  $\theta$ , then the work done,

$$W = F \cos \theta \times s,$$

where,  $F \cos \theta$  is the component of force. The SI unit of work is joule (J). One Joule is the amount of work done, when the point of application

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of force of one newton acting on a body, moves it through a distance of one meter in the direction of force.

### POWER

The rate of doing work is called power. It is measured by the amount of work done in unit time. If  $W$  is the work done in time  $t$ , then power  $P = W/t$ .

The SI unit of power is joule per second ( $\text{Js}^{-1}$ ). It is also given by a special unit watt, which is equal to 1 joule per second. A larger unit of power is called kilowatt, which is equal to 1000 watt. The unit of electrical energy consumption is kilowatt-hour (kWh). One kilowatt-hour is the power consumed at the rate of 1000 watts for one hour.  $1 \text{ kWh} = 1000 \times 60 \times 60 = 36,00,000$  watt per second = 36,00,000 joules. The older unit of power is horse power (HP), and 1 HP is equal to 746 watts.

### ENERGY

The Energy of a body is its ability to do work. It is measured by the amount of work that it can perform. The SI unit of energy is joule. The electron volt (eV) is also used as unit of energy in radiation physics. There are many forms of energy, such as mechanical energy, heat energy, light energy, electrical energy, chemical energy, atomic energy, etc. There are two forms of mechanical energy, viz. potential energy and kinetic energy.

#### Potential Energy

The potential energy of a body is the energy it possesses by virtue of its position or state of strain, e.g. water stored up in a reservoir, a wound spring, compressed air, etc. For a body of mass ' $m$ ' remaining at rest at a height  $h$  above the ground, the potential energy is equal to the work done in raising the body from the ground to that height.

$$\begin{aligned}\text{The work done} &= \text{force} \times \text{displacement} \\ &= mg \times h\end{aligned}$$

$$\begin{aligned}\text{Potential energy} &= mgh \text{ joule, where 'g' is the acceleration due to} \\ &\quad \text{gravity.}\end{aligned}$$

#### Worked Example 1.1

A patient of weight 50 kg on a wheel chair has to be lifted onto a examination couch, which is 25 cm higher than wheel chair. Calculate the work done to carry out the above task ( $g = 9.81 \text{ ms}^{-2}$ ).

$$\begin{aligned}
 W &= \text{Force} \times \text{distance} \\
 &= mg \times \text{distance} \\
 &= 50 \times 9.81 \times 0.25 \\
 &= 120 \text{ J}
 \end{aligned}$$

The work done in lifting the patient on to the couch needs 120 J energy, which will increase the potential energy of the patient.

### Kinetic Energy

The kinetic energy of a body is the energy possessed by the body by virtue of its motion. Let a body of mass  $m$  moves with a velocity  $v$ , then,

$$\text{Kinetic energy} = (1/2) mv^2 \text{ joule}$$

#### *Worked Example 1.2*

A film cassette of mass 2 kg is kept in a shelf at a height of 1.5 m, possess a potential energy of 25 J. If the cassette falls on to the floor, what will be its speed?

$$\begin{aligned}
 \text{Kinetic energy} &= \frac{1}{2} \times 2 \times v^2 \\
 25 &= \frac{1}{2} \times 2 \times v^2 \\
 v &= 5 \text{ ms}^{-1}
 \end{aligned}$$

The cassette may fall on the floor with a speed of  $5 \text{ ms}^{-1}$ .

### MOMENTUM

The momentum ( $P$ ) of a moving body is the product of mass ( $m$ ) and velocity ( $v$ ) and it is given the relation:

$$P = mv$$

The momentum is a vector quantity and its direction is the same as its velocity, the unit is  $\text{kg-ms}^{-1}$ .

### TEMPERATURE AND HEAT

---

Matter is made up of atoms or molecules. These atoms and molecules are in regular movement in solids and random movement in liquids and gases. They possess potential energy as well as kinetic energy. The total energy of the molecules in the system is called as internal energy of the system. The kinetic energy is responsible for the hotness and coldness of the body.

Temperature is the measure of hotness and coldness of the body. When a body is heated, its molecules are in vigorous movement, and

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therefore have high energy, and the body is said to be in high temperature. When a body is cooled lower and lower, its kinetic energy decreases, and the body is said to be in lower temperature. Change of temperature may alter the electrical resistance, conductivity, viscosity and rate of chemical reaction of the substance, e.g. change of body temperature alter metabolism. Temperature is measured in degrees with the help of thermometers. There are three scales of temperature, namely, (i) Celsius scale, (ii) Kelvin scale, and (iii) Fahrenheit scale.

### **Celsius Scale**

In this scale, the temperature of the melting of ice is zero ( $0^{\circ}\text{C}$ ) and temperature of the boiling water is  $100^{\circ}\text{C}$ . The range between melting point and boiling point is divided into 100 intervals called degrees.

### **Kelvin Scale**

In the Kelvin scale or absolute scale of temperature, 0 degree is named as absolute zero and it is denoted as 0 K. The absolute zero is the temperature at which the molecules will have zero speed. In this scale, the temperature of melting ice is 273.15 K and the temperature of boiling water is 373.15 K. The range between the two is divided into 100 intervals. One interval is the same in both centigrade and Kelvin scale of temperature. The 0 K temperatures is equal to  $-273^{\circ}\text{C}$  in Celsius scale. At 0 K, the atomic particles are at rest and hence, it is called absolute zero. It means that the body do not have internal energy at absolute zero.

### **Fahrenheit Scale**

In this scale, the melting ice is at  $32^{\circ}\text{F}$  and boiling water is at  $212^{\circ}\text{F}$ . The entire range is divided into 180 degrees. The body temperature is about  $98.4^{\circ}\text{F}$  equal to  $37^{\circ}\text{C}$  or 310 K. The relation between Celsius and Fahrenheit scale is given by

$$\text{C}/100 = (\text{F} - 32) \div 180 \text{ or } 1.8 \text{ C} = \text{F} - 32, \text{ or } \text{C} = (\text{F} - 32) \div 1.8$$

#### *Worked Example 1.3*

Convert  $86^{\circ}\text{ F}$  into degrees of celsius

Here  $\text{F} = 86$

$$\text{C} = (\text{F} - 32) \div 1.8 = (86-32) \div 1.8 = 54 \div 1.8 = 30^{\circ}\text{C}.$$

## **HEAT**

Heat is a form of internal energy, which can be transferred from one part of the body to another. If a hot body and a cold body are placed

in close contact, the hot body will transfer some of its heat energy to the cold body until the temperature of the two become equal. The difference in temperature creates temperature gradient. There are three methods of heat transfer, namely, conduction, convection and radiation.

### **Conduction**

It is the process in which heat energy is transferred by collisions between neighboring atoms, without the visible motion of the particles. Conduction takes place in solids, liquids and gases. Let us consider a rod of length L and area A and temperature  $\theta_1$  and  $\theta_2$  at their ends. The rate of flow of heat ( $dQ/dt$ ) is directly proportional to cross-sectional area (A), temperature gradient  $(\theta_1 - \theta_2)/L$  and thermal conductivity (k) of the material. The thermal conductivity of a material is its inherent ability to conduct thermal energy and it is expressed in  $\text{Wm}^{-1}\text{K}^{-1}$ . The relation for thermal conductivity is given by

$$dQ/dt = kA (\theta_1 - \theta_2)/L$$

The thermal conductivity of various materials are listed in Table 1.3. Metals in general are good conductors of heat, e.g. silver, copper, etc. Nonmetals are bad conductors of heat, e.g. glass, rubber, wood, etc.

### **Convection**

It is the process in which heat energy is transferred by the actual motion of the particles of the body. Heat in liquid causes the fluid to expand and making it less dense and starts rising. The cold, dense fluid molecules move to their place from other area. Convection takes place in liquids

**TABLE 1.3 Thermal conductivity of various materials**

Material	Specific heat capacity, $\text{Jkg}^{-1}\text{K}^{-1}$	Thermal conductivity, $\text{Wm}^{-1}\text{K}^{-1}$ at $20^\circ\text{C}$
Aluminum	910	237
Tungsten	136	178
Molybdenum	246	140
Graphite	711	130
Copper	386	401
Rhenium	138	48
Water	4200	0.59
Glass	67	0.9–1.3

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and gases, e.g. trade winds, land and sea breezes. Convection current in air remove heat from X-ray tube housing to the atmosphere. Oil and then water circulation remove heat from large X-ray systems like CT scan. Convection forms the basis for domestic heating system and air-conditioning. Convection may be caused by natural or forced circulation

### **Radiation**

It is the process by which heat energy is transmitted from one place to another without the aid of any material medium. When a body has internal energy, its atoms and molecules vibrate and emits electromagnetic radiation, which can transport energy across a vacuum, e.g. heat reaches the earth from the sun. A black body and matt surface will radiate and absorb energy efficiently, while white and glossy surface will not. Stefan's law states that the rate of heat energy emission ( $dQ/dt$ ) is directly proportional to the area of the emitting surface (A) and the fourth power of its temperature (T)

$$dQ/dt \propto \sigma AT^4$$

where,  $\sigma$  is the Stefan–Boltzmann constant =  $5.670 \times 10^{-8}$  W m<sup>-2</sup>K<sup>-4</sup>

The SI unit of heat is joule. However, the special unit calorie is still in use. One calorie is the amount of heat which will raise the temperature of one gram of water by one degree Celsius. 1 calorie = 4.2 joules.

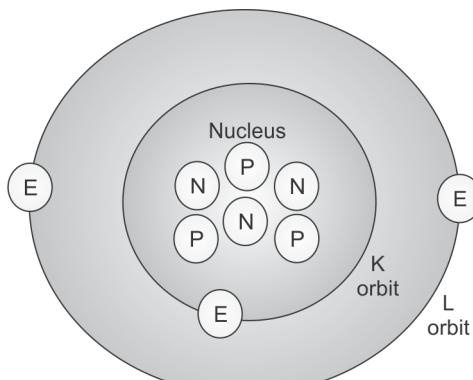
### **HEAT CAPACITY**

The heat capacity of a material is the heat required to raise its temperature by 1 K. It is independent of material size or shape and expressed in JK<sup>-1</sup>. The heat required to raise temperature of a 1 kg material by 1K is called specific heat capacity, and it is expressed in Jkg<sup>-1</sup>K<sup>-1</sup>.

### **ATOMIC STRUCTURE**

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All matter is composed of elements and compounds. Elements are the simplest chemical entity, which cannot be broken further, e.g. hydrogen, carbon. Two or three elements form a compound, e.g. water. The smallest particle of an element is the atom, which forms the fundamental unit of matter. The atoms are very small and its diameter is of the order of  $10^{-10}$  m. Every atom posses a central core called a nucleus, which is positively charged. The diameter of the nucleus is of the order of  $10^{-14}$  m (Fig. 1.1).



E = Electron, P = Proton, N = Neutron

**FIG. 1.1:** Atomic structure

The nucleus consists of two particles called protons and neutrons and collectively known as nucleons. The protons are positively charged and the neutron has no charge. The space around the nucleus consists of another important particle, called electron. The electrons are negatively charged particle, and they circulate around the nucleus at varying distances, similar to planets rotation around the sun. The number of electrons in an atom is equal to the number of protons and hence, atom is said to be neutral.

There are two types of forces exist in the nucleus. The electrostatic repulsive force, exist between particles of similar charge. The strong forces (attractive) resulting from the exchange of pions among all nucleons, hold the nucleus together. These two forces act in opposite directions. The nucleus has energy level and the lowest energy state is called the ground state. Nuclei with energy excess of the ground state are said to be in an excited state. Excited states that exists  $> 10^{-12}$  s are referred to as meta stable or isomeric states.

### ATOMIC NUMBER AND MASS NUMBER

In 1913, HGJ Mosley stated that the atomic number of an atom is the number of protons in the nucleus. It is also equal to the number of electrons of the atom, which is represented by Z. The mass number of an atom is the total number of protons and neutrons in the nucleus and it is denoted by A. An element (X) is symbolically described as  ${}_z^A X$ . The subscript gives the atomic number Z while superscript gives the mass number A. Some of the important elements, their symbol, atomic number and mass number are given in Table 1.4.

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**TABLE 1.4** Symbol, atomic number, and mass number of few elements

Element	Symbol	Atomic No. (Z)	Mass No. (A)
Hydrogen	H	1	1, 2, 3
Aluminum	Al	13	27
Cobalt	Co	27	59, 60
Copper	Cu	29	63, 65
Tin	Sn	50	116, 118, 120
Iodine	I	53	125, 127, 131
Cesium	Cs	55	133, 134, 137
Barium	Ba	56	137, 138
Tungsten	W	74	182, 183, 184, 186
Lead	Pb	82	206, 207, 208
Radium	Ra	88	224, 226, 228

### EFFECTIVE ATOMIC NUMBER

The effective atomic number ( $Z_{\text{eff}}$ ) is meant for a compound or mixture, which has more than one element.  $Z_{\text{eff}}$  is the atomic number of an element with which photons interact the same way as with the given composite material. Mayneord has defined the effective atomic number as follows:

$$Z_{\text{eff}} = (a_1 Z_1^{2.94} + a_2 Z_2^{2.94} + \dots + a_n Z_n^{2.94})^{1/2.94}$$

where,  $a_1, a_2, \dots, a_n$  are the fractional contribution of each element to the total number of electrons in the mixture. The density and effective atomic number of few compounds are given in Table 1.5.

### ISOTOPES

The atoms composed of nuclei with the same number of protons but different number of neutrons is called isotopes. In other words, isotopes have the same atomic numbers and different mass numbers, e.g. hydrogen have 3 isotopes, namely:

${}_1^1\text{H}$  have 1 proton (Hydrogen),

${}_1^2\text{H}$  have 1 proton and 1 neutron (Deuterium)

${}_1^3\text{H}$  have 1 proton and 2 neutrons (Tritium).

Isotopes of an element have the same chemical properties but have different physical properties. Isotopes capable of performing radioactivity are called radio-isotopes and their nucleus is said to be unstable. Nuclides

**TABLE 1.5** Density and effective atomic number of few compounds

Material	Effective atomic number ( $Z_{\text{eff}}$ )	Density ( $\rho$ ), $\text{kgm}^{-3} \times 10^{-3}$
Air	7.78	1.205
Muscle	7.64	1.04
Water	7.5	1.0
Bone	12.3–14	1.65
Fat	6.46	0.916
PMMA	6.56	1.18
Polystyrene	5.74	1.044
LiF	8.31	2.675

having the same mass numbers but different number of protons are called isobars. Nuclides having same number of neutrons but different number of protons are called isotones. An isomer is the excited state of a nucleus, and it will have same number of proton and neutron.

## ELECTRON SHELLS

In 1921, Burry and Bohr independently gave a scheme for the arrangement of electrons in an atom. According to this scheme, the orbits in the atom are named as shells and denoted as K, L, M, N, etc., from the nucleus. The following are the rules of their scheme: The maximum number of electrons in each shell can be obtained from the formula  $2n^2$  where  $n = 1, 2, 3, 4$ , etc. In the case of K shell,  $n = 1$ , the number of electrons in the K shell =  $2 \times 1^2 = 2$ . In the case of L shell,  $n = 2$ , the number of electrons in the L shell =  $2 \times 2^2 = 8$  and so on. Each shell is provided with subshells, which are denoted as s, p, d, f, etc. The K shell ( $n = 1$ ) has one subshell, namely, 1s. The L-shell ( $n = 2$ ), has two subshells, namely, 2s and 2p and so on. One electron in the s subshell of K shell is denoted as 1s<sup>1</sup>, while 2 electrons in the same subshell is denoted as 1s<sup>2</sup>.

The outermost orbit is called valence shell, which is responsible for chemical, thermal, optical and electrical properties of the element. No valence shell has more than 8 electrons, e.g. metals have one, two or three valence electrons. The elements are arranged in the periodic table based on the similarities of chemical properties of different elements. As we go across the periodic table the atomic number of the atom increases. The number of electron also increases in the same step.

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### **QUANTUM NUMBER**

The energy level of an electron or position in an atom is described by quantum numbers as follows:

- i. The principle quantum number ( $n$ ) defines the main energy level or shell of an orbiting electron. For K shell,  $n = 1$ ; for L shell,  $n = 2$  and so on.
- ii. The azimuthal quantum number ( $l$ ) describes the angular momentum of the orbiting electrons. It can have values 0, 1, 2, 3....  $n-1$ , e.g. M shell principal quantum number is 3 and its azimuthal quantum numbers are  $3 - 1 = 2$ , which are 0, 1 or 2.
- iii. The magnetic quantum number ( $m$ ) describes the spatial orientation of the plane of the orbiting electron and it can have values from  $-l$  to  $+l$ . When  $l = 1$ ,  $m$  can have  $-1, 0, +1$  values.
- iv. The spin quantum number ( $s$ ) describes direction of spin of the electron and it can value  $+1/2$  (spin up) or  $-1/2$  (spin down).

### **IONIZATION**

Removal of one or more electrons from a neutral atom is called ionization. After ionization, the remainder of the atom is left with positive charge and is known as positive ion. The positive atom and the removed electrons form one ion pair.

### **BINDING ENERGY**

The binding energy of an electron in an atom is the energy required to remove the electron completely from the atom against the attractive force of the positive nucleus. The magnitude of the binding energy depends on the atomic number and the shell from which the electron is being removed. It is greater for elements of higher atomic number and greatest for the K shell (inner most shell).

Binding energies are negative because they represent amounts of energy that must be supplied to remove electrons from atoms. Electron shells are often described in terms of the binding energy of electrons occupying the shells, e.g. the binding energy of hydrogen K shell is  $-13.5 \text{ eV}$  and  $-3.4 \text{ eV}$  for L shell. The K-shell binding energies of various elements are given in Table 1.6.

### **EXCITATION**

In an atom, if energy is supplied, the electrons can be moved from the inner orbit to the outer orbit. Now, the atom will have more energy than its

**TABLE 1.6** Atomic number (Z) and binding energies ( $E_k$ ) of few elements

Element	Z	$E_k$ , keV
Aluminum	13	1.6
Calcium	20	4
Molybdenum	42	20
Iodine	53	33
Barium	56	37
Gadolinium	64	50
Tungsten	74	70
Lead	82	88

normal state. It is said to be in an excited state and the process is known as excitation. For example, to move an electron from K to L shell of the hydrogen atom, the energy required is  $(-3.4 \text{ eV}) - (-13.5 \text{ eV}) = 10.1 \text{ eV}$ .

### ELECTRON VOLT

The electron volt (eV) is the unit of energy in radiation physics, where it deals with microscopic objects. One electron volt is the kinetic energy imparted to an electron accelerated across a potential difference of one volt. In practice, we use kiloelectron volt (keV) and million electron volt (MeV) and

$$1 \text{ eV} = 1.6 \times 10^{-19} \text{ J} = 1.6 \times 10^{-12} \text{ erg} = 4.4 \times 10^{-26} \text{ kWh}$$

The electron volt describes potential as well as kinetic energy. The binding energy of an electron in an atom is a form of potential energy and it is expressed in keV.

### ELECTROMAGNETIC RADIATION

An electric charge is surrounded by an electric field and if the charge moves, a magnetic field is produced. When the charge undergoes an acceleration or deceleration, the magnetic and the electric fields of the charge will vary. The combined variation of the electric and magnetic fields results in loss of energy. The charge radiates this energy in a form known as electromagnetic radiation. The electromagnetic radiation moves in the form of sinusoidal waves (Fig. 1.2). The nature of the electromagnetic radiation (X-rays, ultraviolet, etc.) depends on the way in which the electric charges are disturbed. Electromagnetic radiations

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are transverse waves that transfer energy away from the electric charge. Electromagnetic radiations may be absorbed or scattered in a medium, resulting in loss of energy.

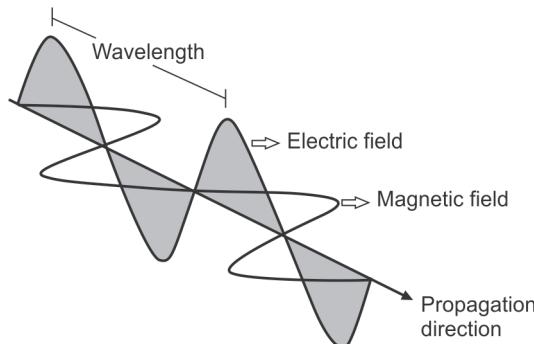


FIG. 1.2: Electromagnetic wave

### WAVE CHARACTERISTICS

The electromagnetic wave have wavelength ( $\lambda$ ), frequency ( $v$ ), and velocity ( $c$ ). The distance between two consecutive positive peaks is known as wavelength. The number of cycles of the wave which pass a fixed point per second is known as the frequency of the wave. The velocity of the wave is the distance traveled per second by the wave. The relation between wavelength, frequency, and velocity of the electromagnetic wave is

$$c = v\lambda$$

All electromagnetic waves, travel at the same velocity in a given medium and its velocity in vacuum is about  $2.998 \times 10^8 \text{ ms}^{-1}$ . The wavelength of X-rays and gamma rays are in nanometers (nm).

### PARTICLE CHARACTERISTICS

Though electromagnetic radiations have the properties of waves, they also behave like particle during interaction with matter. The actual amount of energy ( $E$ ) carried by a photon is given the equation  $E = hv$ , where,  $h$  is the Planck's constant  $= 6.63 \times 10^{-34} \text{ J}$ . Substituting the value of  $v = c/\lambda$  in the above equation, the energy

$$E (\text{keV}) = hc/\lambda = 1.24/\lambda$$

where,  $\lambda$  is in nanometer (nm). It is seen that the energy of the photon is inversely proportional to its wavelength and as the wavelength decreases, the energy increases.

## MASS ENERGY EQUIVALENCE

Einstein's theory of relativity states that mass and energy are equivalent and are interchangeable. In any reaction, the sum of the mass and energy must be conserved. Einstein showed that the speed of some nuclear processes approach the speed of light. At these speeds, mass and energy are equivalent.

$$E = mc^2$$

where,  $E$  represents the energy equivalent to mass ' $m$ ' at rest and ' $c$ ' is the speed of light in a vacuum. For example, the energy equivalent of an electron of mass  $9.109 \times 10^{-31}$  kg is

$$\begin{aligned} E &= 9.109 \times 10^{-31} \text{ kg} \times (2.998 \times 10^8 \text{ m/s})^2 \\ &= 0.511 \text{ MeV} \end{aligned}$$

## ELECTROMAGNETIC SPECTRUM

Electromagnetic spectrum includes radiowaves, microwaves, infrared, visible light, ultraviolet, X-rays, gamma rays and cosmic rays (Fig.1.3). All of them travel at a velocity ' $c$ ' in a vacuum. The wavelength and photon energy of the whole range of electromagnetic radiation are summarized in Table 1.7.

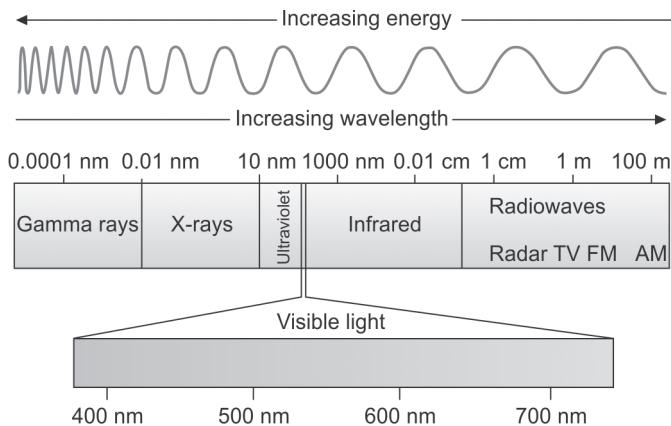
## IONIZING RADIATION AND NON-IONIZING RADIATION

Ionization is a process of removal of electron from neutral atom. The radiation which does ionization in a medium, by removal of electron is called ionizing radiation, e.g. UV, X-rays, and gamma rays have sufficient energy to do ionization. As a result, ionized atoms and molecules or ion-pairs are produced. This forms the basis for biological effects of radiation. Radiation that do not have sufficient energy to produce ionization are called non-ionizing radiation, e.g. visible light, infrared, radiowaves, and TV broadcasts, etc.

## FLUORESCENCE

When electromagnetic radiation falls on a phosphor, visible or ultraviolet light is emitted from the phosphor and it is called as luminescence. The electromagnetic radiation raises the valence electrons to the conduction band, which return to the valence band to fill up the holes. As electron falls through the luminescence centers, they emit the surplus energy in the form of flashes of light, called luminescence. If the luminescence is instantaneous, within  $10^{-8}$  s, it is called fluorescence.

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**FIG. 1.3:** Electromagnetic spectrum

**TABLE 1.7** Electromagnetic radiation spectrum

Radiation	Wavelength	Frequency	Energy
Radiowaves	1000 – 0.1 m	0.3 – 3000 MHz	0.001 – 10 $\mu$ eV
Microwaves	100 – 1 mm	3 – 300 GHz	10 – 1000 $\mu$ eV
Infrared	100 – 1 $\mu$ m	3 – 300 THz	10 – 1000 meV
Visible light	700 – 400 nm	430 – 750 THz	1.8 – 3 eV
Ultraviolet	400 – 10 nm	750 – 30000 THz	1.8 – 100 eV
X- and gamma rays	1 nm – 0.1 pm	$3 \times 10^5$ – $3 \times 10^9$ THz	1 keV – 10 MeV

The energy of light emitted depends on the difference in energy across the luminescence centers. It is always less than the energy which originally stimulated the fluorescence, e.g. a phosphor exposed to ultraviolet may emit visible light. Fluorescent phosphors, such as thallium activated sodium iodide (NaI:TI, gamma camera), terbium activated gadolinium oxysulfide (intensifying screen) and sodium activated cesium iodide (image intensifier) are used in diagnostic radiology.

If the emission of light is delayed beyond  $10^{-8}$  s, it is called phosphorescence. When the valence electrons are stimulated, they get trapped in the conduction band. They acquire energy from the atom (internal energy) and return to the valence band by emitting luminescence. It is a random process, which takes time to accomplish. The emission of light decays exponentially with a time constant, that depends upon the temperature of the phosphor.

## INVERSE SQUARE LAW

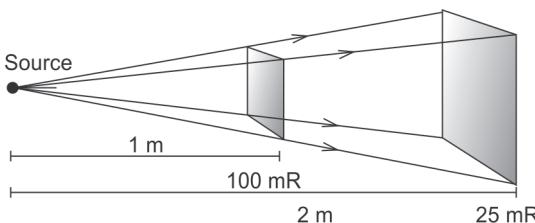
The intensity of electromagnetic radiation is inversely proportional to the square of the distance from its source. Let us consider a point source 's', emitting radiation at constant rate. The radiation spread over the inner surface of an imaginary sphere of radius  $d$  with surface area  $4\pi d^2$ . Then the radiation intensity at a point 'd' is given by the relation

$$I \propto 1/d^2$$

The inverse square law is based on the following assumptions:

- i. The source of radiation is a point source.
- ii. The radiation travels in straight lines.
- iii. The radiation is emitted equally in all directions.
- iv. The energy is radiated at a constant rate.
- v. No radiation energy is lost on its way from the source to the point of measurement.

Let 100 mR be the radiation exposure at 1 m for a point source (Fig. 1.4). The radiation exposure at 2 m is found to be 25 mR, by inverse square law. Hence, if distance is doubled, the radiation is reduced by a factor of 4. Keeping higher distance always reduce radiation exposure.



**FIG.1.4:** Inverse square law

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## RADIOLOGICAL MATHEMATICS

### LOGARITHMS

The logarithm of a decimal number is the exponent to which the base must be raised to produce the number. For example, the logarithm of 1000 to base 10 is 3, because 1000 is 10 to the power 3:  $1000 = 10^3 = 10 \times 10 \times 10$ . More generally, if  $x = b^y$ , then  $y$  is the logarithm of  $x$  to base  $b$ , and is written as  $\log_b(x)$ , so  $\log_{10}(1000) = 3$ . There are three types of logarithms, namely, common logarithm ( $\log_{10}$ ), natural logarithm ( $\log_e$ ), and binary logarithms ( $\log_2$ ), where ' $e$ ' = 2.71828, e.g.  $\log_{10}2 = 0.301$ , the base 10 must be raised to power of 0.301,  $10^{0.301} = 2$ .

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Similarly,  $\log_e 2 = 0.693$ , the base 'e' must be raised to power of 0.693,  $e^{0.693} = 2$ .

The measurements of optical density and sound intensity are expressed in logarithm to base 10. Radioactive decay, and X-ray attenuation uses logarithm of base 'e', which is denoted by  $\ln_e$  (natural logarithm). Logarithmic scales reduce wide-ranging quantities to smaller scopes. Logarithm is useful to describe many radiation events such as X-ray absorption, radioactive decay, etc.

### GRAPHS

Graph gives the relationship between physical quantities, plotted as series of points or lines with reference to the set of axis. A Cartesian graph has two axis, namely, 'x' axis called abscissa and 'y' axis called ordinate. The x axis contains independent variable (time, distance) and the 'y' axis contains dependent variable (velocity, exposure).

If a physical quantity 'y' varies with 'x' in a proportional way, then a linear plot can be drawn. It is straight line graph obeying the equation

$$y = mx + c$$

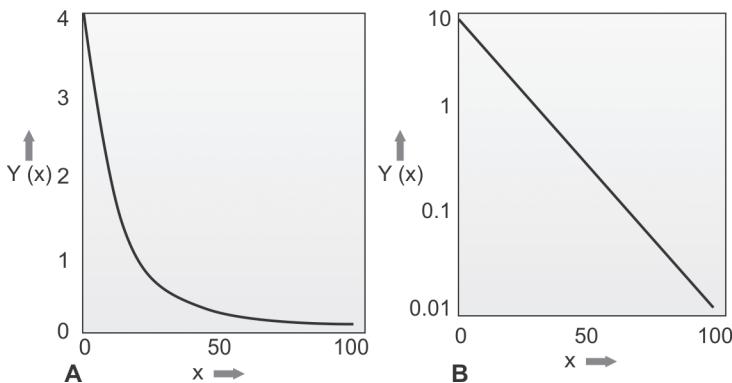
where, m is the slope of the line and c is the intersection with the y axis.

Logarithmic functions such as  $e^x$  and  $e^{-x}$  can also be plotted as curve, where a rapid increase or rapid decrease may be seen. A semi-log graph is a way of visualizing such data that are changing with an exponential relationship. One axis is plotted on a logarithmic scale and the other in linear scale. On a semi-log graph the spacing of the scale on the y-axis is proportional to the logarithm of the number, not the number itself. It is equivalent to converting the Y values to their log, and plotting the data on linear (lin-lin) scales. The term log-lin is used to describe a semi-log plot with a logarithmic scale on the y-axis, and a linear scale on the x-axis (Fig.1.5).

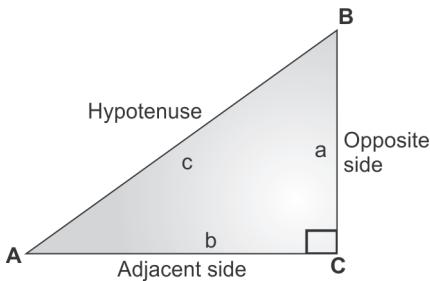
This kind of plot is useful when one of the variables being plotted covers a large range of values and the other has only a restricted range. The advantage being that it can bring out features in the data that would not easily be seen if both variables had been plotted linearly. Semi-log plot requires only few measurements of the exponential function.

### TRIGONOMETRY

Trigonometry is a mathematics which deals with triangles and the relation between angle and sides (Fig.1.6). If one angle of a triangle is



**FIG. 1.5:** (A) In a lin-lin (linear) graph; (B) Log-lin (semi-log) graph



**FIG. 1.6:** Relation between angles and sides in a trigonometry

90 degrees and the other angle is known, then the third angle can be obtained easily. Since the sum of the angles is 180 degrees, the two acute angles therefore added up to 90 degrees, they are said to be complementary angles. The shape of a triangle is determined by the angles. Once the angles are known, the ratios of the sides can be determined, regardless of the overall size of the triangle. If the length of one of the sides is known, the other two can be determined. These ratios are given by the following trigonometric functions of the known angle A, where a, b and c refer to the lengths of the sides in the accompanying figure:

Sine function (sin), defined as the ratio of the opposite side to the hypotenuse.

$$\sin A = \frac{\text{Opposite side}}{\text{Hypotenuse}} = a/c$$

Cosine function (cos), defined as the ratio of the adjacent side to the hypotenuse.

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$$\cos A = \frac{\text{Adjacent side}}{\text{Hypotenuse}} = b/c$$

Tangent function ( $\tan$ ), defined as the ratio of the opposite side to the adjacent leg.

$$\tan A = \frac{\text{Opposite side}}{\text{Adjacent side}} = a/b$$

The hypotenuse is the side opposite to the 90 degrees angle in a right triangle; it is the longest side of the triangle, and one of the two sides adjacent to angle A. The adjacent leg is the other side that is adjacent to angle A. The opposite side is the side that is opposite to angle A. The terms perpendicular and base are sometimes used for the opposite and adjacent sides, respectively.

### STATISTICS

**Source of errors:** There are three types of errors in measurements, namely, systemic error, random error and blunder. Systemic error occurs when measurements differ from the correct values in a systemic fashion. Random error is caused by random fluctuations in the measurement process itself. The processes by which radiation is emitted and by which radiation interacts with matter are random in nature. Therefore, all radiation measurements are subject to random error. The counting statistics helps us to judge the validity of measurements.

**Accuracy and precision:** If a measurement is close to the correct value, it is said to be accurate. If measurements are reproducible, they are said to be precise. Precision does not imply accuracy. If a set of measurements differ from the correct value in a systematic fashion, the data are said to be biased.

#### **Mean, median and standard deviation**

The mean is the arithmetic average of a group of data. The mean ( $x$ ) of a set of measurements is defined as

$$x = \frac{x_1 + x_2 + x_3 + \dots + x_N}{N}$$

where, N is the number of measurements. The median is measure of the central tendency and is the value that separates the data in half and defines the 50%. It is the middle most measurement, if the number of measurements is odd. It is the average of the two middle

most measurements, if the number measurements are even. For example, the median of the five measurements 5, 8, 9, 12 and 14 is 9.

The variance ( $\sigma^2$ ) and standard deviation ( $\sigma$ ) are measures of the variability of a set of measurements. The standard deviation is used to describe the spread of a data set and is the square root of the average of the square of all the sample deviations. The variance is determined from a set of measurements as follows

$$\sigma^2 = \frac{(x_1 - x)^2 + (x_2 - x)^2 + \dots + (x_N - x)^2}{N - 1}$$

where,  $N$  is the total number of measurements and  $x$  is the sample mean. The standard deviation is the square root of the variance,

$$s = \sqrt{\sigma^2}$$

When samples are taken from a large population, there is uncertainty between the sample mean and the actual population mean. This is measured by the standard error, given by the relation

$$\text{Standard error} = \sigma/\sqrt{N}$$

The coefficient of variation (CV) is a measure of spread within the samples, given in percentage. It is given by the relation

$$CV = (\sigma/x)100$$

where,  $\sigma/x$  is the fractional error in the measurements.

# 2

# Electricity, Electronics and Magnetism

## ELECTRIC CHARGE

---

The term electric is derived from the Greek word electron. Electric bodies said to possess electric charge ( $q$ ) and it is a basic property of any matter. There are two types of charges, namely, (i) positive charge and (ii) negative charge. Two like charges repel each other and two unlike charges attract each other. The unit of charge is coulomb. One coulomb (C) is defined as the quantity of charge which when placed at a distance of 1 meter in air or vacuum from an equal and similar charge experiences a repulsive force of  $9 \times 10^{-9}$  N. The amount of charge in an electron is equal to  $1.6021 \times 10^{-19}$  coulombs.

The charges can neither be created nor be destroyed, and the total amount of charge in the system does not change. While calculating the total charge in a system, the signs of the charges should be taken into account.

## ELECTRICAL FORCE AND FIELD

The force between two charged particles is directly proportional to the product of the magnitude of the charges and inversely proportional to the square of the distance ( $r$ ) between them. If  $F$  is the force between two charges  $q_1$  and  $q_2$ , separated by a distance  $r$  then,

$$F = \frac{q_1 q_2}{4\pi \epsilon r^2}$$

where, ' $\epsilon$ ' is the absolute permittivity of the medium. The permittivity of free space is  $8.85 \times 10^{-12} \text{ C}^2 \text{ N}^{-1} \text{ m}^{-2}$ . The force between the particles may be a attractive force or repulsive force.

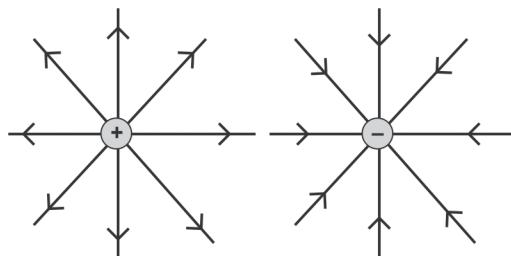
The space surrounding an electric charge in which another charge experiences a force is called an electric field. The electrical field strength

(E) at a point is the force experienced by a unit positive charge kept at that point and it is given by

$$E = F/q \text{ newton/coulomb}$$

Electric field is a vector quantity and has a magnitude and direction. An electrical field is represented by electrical lines of forces (Fig. 2.1). They start from positive charge and terminate on negative charge. An electrical dipole possess positive and negative charge distribution, but net charge is zero. When placed in an electric field, the dipole tends to align with the field, because of the torque exerted by the field. When a charged particle is placed in an electrical field, it experiences a force ( $qE$ ) and acceleration ( $a$ ) =  $qE/m$ , where,  $m$  is the mass and ' $q$ ' is the charge of the particle.

Electric induction is the phenomenon in which positive and negative charges are accumulated or separated in a substance, when a charged body is brought nearer to it.



**FIG. 2.1:** Electric lines of force

## ELECTRICAL POTENTIAL

The electric potential ( $V$ ) at a point in an electric field is the work done ( $W$ ) in taking a unit positive charge ( $q$ ) from infinity to that point, i.e.

$$V = W/q$$

Positive charges flow from a point of higher potential to point of lower potential and negative charges flow in the reverse direction. The unit of potential is volt and one volt is equal to 1 joule per coulomb. The potential is a scalar quantity and the potential of earth is taken as zero. In practice, kilovolt (kV) and megavolt (MV) are used as units,  $1 \text{ kV} = 1000 \text{ volts}$  and  $1 \text{ MV} = 10^6 \text{ volts}$ .

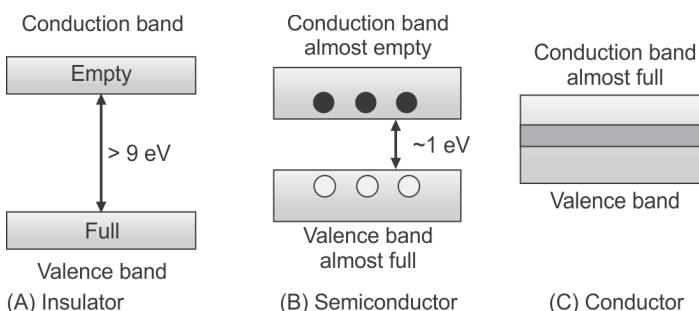
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### **CONDUCTORS, INSULATORS AND SEMICONDUCTORS**

Substances in which electric charge moves freely are known as conductors. Substances, which do not allow charge to move freely through them, are known as insulators or dielectrics. The term insulator or conductor is only a relative term and nobody is perfectly insulating or conducting. Substances, which are having their conductivity intermediate between conductor and insulator, are known as semiconductors.

As per the band theory of conduction, matter is made up of three energy levels, namely, filled band, valence band, and conduction band. Valance band is the highest energy band whose electrons are tied up to individual atoms. It corresponds to the valence shell of a single isolated atom. The filled bands are below the valance band, and they do not contribute to electrical conduction. Hence, it is normally not included in the energy band diagrams.

The conduction band is above the valance band and the electrons are not tied to particular atoms. Hence, it offers free electrons for electrical conduction. The gap between the valance and conduction band is called forbidden gap, which is responsible for the conduction properties of materials. Based on the forbidden gap width, materials may be classified as conductor, insulator and semiconductors (Fig. 2.2).



**FIG. 2.2:** Insulator, semiconductor, and conductor

### **CONDUCTOR**

In conductors, the highest electron energy levels are partially filled and hence its electrons are free to move. There is no forbidden gap between valance and the conduction band, hence electrons move easily from valance band to conduction band. Metals, such as copper, silver and aluminium are good electrical conductors.

## **INSULATOR**

In insulators, the forbidden gap is large,  $> 9$  eV and the electrons unable to flow to the conduction band. Hence, the conduction band is empty and no flow of electric current, e.g. oil, glass, rubber and plastic. At very high temperature, few electrons may move from the valance band, but the material undergo breakdown. This breakdown depends on the applied voltage and the thickness of the material. Hence, X-ray cables are made up of higher thickness of insulation material.

## **SEMICONDUCTOR**

In semiconductors, the width of the forbidden band is 1 eV, e.g. germanium and silicon. At low temperatures, there is no electron flow from valance to conduction band due to lack of sufficient energy and they behave like an insulator. However, at room temperature, they utilize the internal energy of the system and gain  $> 1$  eV energy. This is sufficient to offer electron flow from valance to conduction band, but in a limited way. As the temperature increases, number of electron also increases, resulting higher conductivity. As the electron leaves the valance band, holes are created, which also act like charge carriers. This type of conduction that takes place in a pure semiconductor is called intrinsic conduction.

The conducting property of a semiconductor can be modified by adding impurities to it, which is called doping. By doing so it is possible to create additional energy levels in the forbidden band, resulting higher conductivity. This type of conductivity made out of doping is called extrinsic conductors. There are two types of extrinsic conductors, namely, N-type and P-type (Fig. 2.3). In N-type, there are extra-energy levels, which helps the electrons to move from the valance band to conduction band. In P-type semiconductor, the extra-energy level helps the holes to move, and offer higher electrical conduction.

### **N-type Semiconductor**

When a pentavalent impurity such as phosphorous or arsenic is added to a pure silicon in the ratio  $1:10^6$ , N-type semiconductor is formed. Four out of five valence electrons of the impurity phosphorous atom form covalent bonds with neighboring silicon atoms. The fifth electron is not associated with any covalent bond and it is free, responsible for conduction. In this type, the majority charge carriers are electrons and the minority charge carriers are holes. Since the impurity donates one electron to the conduction band, it is called donor impurity.

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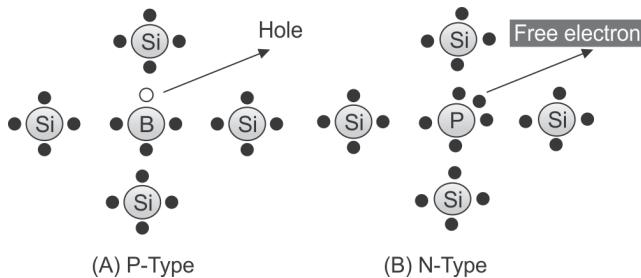


FIG. 2.3: P and N types semiconductor

### P-type Semiconductor

When a trivalent impurity such as boron is added to a pure silicon (Si), P-type semiconductor is formed. The three valence electrons of boron atom form covalent bands with the three neighboring silicon atoms. The fourth electron of the Si atom is unable to form a covalent bond with the boron atom. Hence, a vacancy is available in the fourth covalent bond. This vacancy is called hole (positive charge) which can accept electrons from other atoms. The majority charge carriers are holes and the minority charge carriers are electrons. Since, there is a hole in the impurity, it is called acceptor impurity.

### SEMICONDUCTOR DIODE

A semiconductor (solid state) diode consists of a P-type and a N-type semiconductors which are joined together (Fig. 2.4). Such a arrangement is called the P-N junction diode. When a P-N junction is formed, the holes diffuse from P region and electrons diffuse from N region due to thermal energy. As a result, the holes and electrons combine with each other and neutralize near the junction. After a short interval of time, a potential barrier is setup near the junction with immobile negative and positive ions which stops further diffusion. The above potential barrier which is created, when a P-N junction is formed is called internal

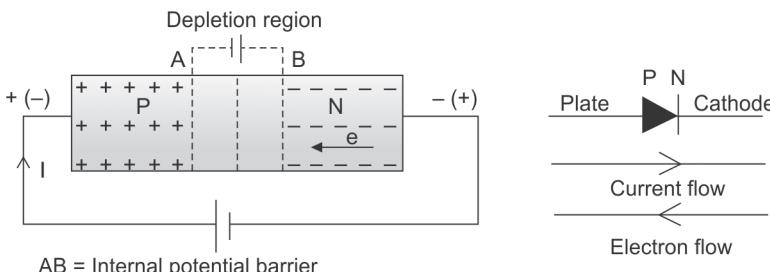


FIG. 2.4: Junction diode and current flow

potential barrier or depletion layer. The width of this barrier region is about  $10^{-6}$  to  $10^{-8}$  m.

A battery is connected to the terminals of the diode. When P is positive and N is negative, the diode is said to be forward biased. Now, the holes in the P region are repelled from the positive terminal of the battery and moves towards the junction. Similarly, the electron move towards the junction. These ions penetrate the depletion region, there by reducing the internal potential barrier. There is a continuous flow of electrons through the junction from N to P region, which will constitute a current. The current flow is in the order of mill amperes.

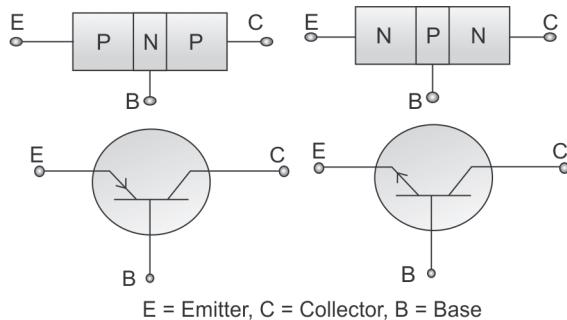
When P is negative and N is positive, the diode is said to be reverse biased. Since the battery terminals attract both holes and electrons, the internal potential barrier is increased. Hence, there is no flow of electrons across the junction and there is no current flow. Only the minority carriers cross the junction constituting very low reverse saturation current. This current is of the order of microamperes. Thus, the PN junction diode allows the electron flow only when P is positive. This property is used for the conversion of AC into DC, which is called rectification.

## TRANSISTORS

A transistor is formed by three semiconductor materials, which are sandwiched together. Schematic symbols for PNP and NPN transistors are shown in Figure 2.5. There are three regions in a transistor and are called emitter, base and collector. The emitter, base and collector are provided with terminals which are labeled as E, B and C. In the schematic symbols, the arrow head is always at the emitter. The arrow head indicates the conventional current direction flow. The junction between emitter and base is called emitter base (EB) junction. The junction between collector and base is called collector base (CB) junction. Hence, a transistor basically consists of two junctions manufactured back to back in a single piece of a semiconductor.

The emitter forms the left hand side of the transistor and its main function is to supply majority charge carriers to the base. The base forms the middle section of the transistor and it is very thin. The collector forms the right hand side of the transistor and its main function is to collect majority charge carriers through the base. In most transistors, the collector region is made physically larger than emitter region, because it has to dissipate much greater power.

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**FIG. 2.5:** PNP, NPN transistor block diagram and symbol

### Principle of Operation

In a PNP transistor the following biasing is required: (i) emitter-base junction is always forward biased and (ii) collector-base junction is reverse biased. The forward bias causes the majority carriers (holes) to diffuse from emitter into base, as a result emitter current ( $I_E$ ) flows. Once the emitter injected holes reach the base, there is a recombination of holes and electrons. Since the base is thin, only few holes recombine with electrons. The other holes reach the collector, causing a collector current,  $I_C$ . If  $I_B$  is the base current due to flow of holes out of base, then applying Kirchhoff's law, that total emitter current equals to the sum of the collector and base current, i.e.  $I_E = I_B + I_C$ . This is true for all type of transistors, irrespective of type and configurations.

Transistor circuit connections are made either by common base, common emitter or common collector types. The common base circuit configuration enables the transistor to function as power amplifier. In this circuit, the collector current is higher than the base current and collector voltage is higher than the emitter voltage. Similarly, a common emitter circuit enables the transistor to function as current amplifier. The gain of transistor is a ratio between the collector current and base current and this is in the order of 100.

### Transistor Applications

Transistors enable a small current to control the flow of a larger current and have applications in switching and amplification, etc. Large number of transistors along with resistors and capacitors are incorporated in a single silicon chip, known as large scale integration (LSI) and very large scale integration (VLSI) circuits that have multiple applications in medicine and industry.

## CAPACITANCE

The property of a conductor to store electric charge is known as capacitance. It is defined as the ratio between the charge and its potential. If  $Q$  is the charge in a conductor of potential  $V$  then, the capacitance  $C$  is given by,

$$C = Q/V$$

Capacitance also refers the amount of charge that can be transferred per unit change in its potential. The unit of capacitance is farad ( $F$ ) and one farad is the capacitance of a capacitor, which requires one coulomb electric charge to raise its potential by one volt. In practice, microfarad and picofarad are used as capacitance units.

$$1 \text{ farad} = 1 \text{ coulomb/volt}$$

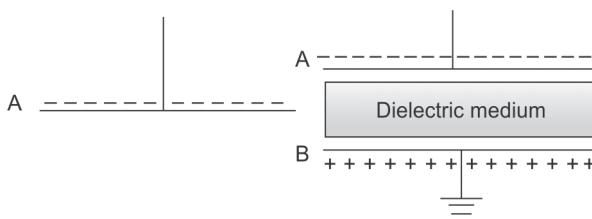
$$1 \text{ microfarad } (\mu\text{F}) = 10^{-6} \text{ farads}$$

$$1 \text{ picofarad} = 10^{-12} \text{ farads.}$$

## CAPACITOR

A capacitor is a device, which increases the capacitance of a conductor. It usually consists of two conductors, one is charged and the other is earthed. The space between the plates is filled with some insulating material called dielectric. To understand the principle, consider a conductor A as shown in Figure 2.6. When it is negatively charged, there is a rise in negative potential, and its capacitance is very small. When a second similar conductor B is brought very nearer to the first, positive charges are induced in B by electric induction. This positive charge decreases the negative potential on A. Hence, for the same charge  $Q$ , the potential  $V$  has fallen. Since  $C = Q/V$ , the capacitance of the first conductor increases.

The capacitance of a capacitor depends on (i) area of overlap between two plates (ii) distance between the plates and (iii) nature of dielectric medium. The capacitors are in different types, the most commonly used one is the parallel plate capacitor.



**FIG. 2.6:** Principle of capacitor

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Capacitors are used (i) to store electric charges, (ii) to measure potential difference and small currents, (iii) to reduce voltage fluctuations, generating oscillations, for providing time delay in various electric circuits, and (iv) to obtain required electric field.

When capacitors are connected in parallel, the total capacitance in the circuit is equal to the sum of the individual capacitance. If three capacitances,  $C_1$ ,  $C_2$  and  $C_3$  are connected in parallel, then the total capacitance  $C$  is given by the relation:

$$C = C_1 + C_2 + C_3$$

If the capacitances are connected in series, then the total capacitance is given by the relation:

$$1/C = (1/C_1) + (1/C_2) + (1/C_3)$$

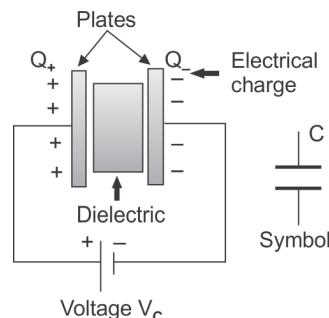
### PARALLEL PLATE CAPACITOR

The parallel plate capacitor consists of two parallel conductors (electrodes) of area  $A$  and separated by a distance  $d$  (Fig. 2.7). A thin layer of dielectric material is sandwiched between the electrodes. One plate is charged positively and the other is negatively charged.

The capacitance of a parallel plate capacitor is proportional to the area of the plates and inversely proportional to their separation distance, and is given by the relation:

$$C = k (A/d)$$

where,  $k$  is a constant called permittivity and it is equal to  $8.84 \times 10^{-12} \text{ Fm}^{-1}$  in free space. Generally, the conductive plates of a capacitor are separated by air or some kind of insulating material or gel rather than the vacuum of free space.



**FIG. 2.7:** Parallel plate capacitor

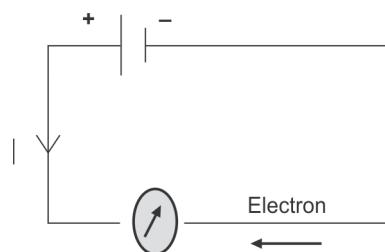
### ELECTRICAL CURRENT

The flow of electric charge in a conductor is called an electric current. It is equal to the quantity of charge passing a given point in one second. Charge may flow through solid, liquid and gas or vacuum. The unit of current is called ampere (A). The electric current through a wire

is called one ampere, if one coulomb of charge flows through the wire in one second. It is found that one ampere current consists of  $6.281 \times 10^{18}$  electrons/s. In practice, milliampere (mA) and microampere ( $\mu\text{A}$ ) are used as units,  $1 \text{ mA} = 10^{-3}\text{A}$ , and  $1 \mu\text{A} = 10^{-6} \text{ A}$ .

## DIRECTION OF CURRENT

Initially, physicists thought that the current is due to the flow of something from positive end to the negative end. This is only an imaginary flow, which is now known as conventional current. After the discovery of electron, it is found, that the electrons are responsible for current and it flows from the negative terminal to the positive terminal. However, the direction of current is given by the conventional current, which is always opposite to the electron flow (Fig. 2.8).



**FIG. 2.8:** Direction of current

## OHM'S LAW

The Ohm's law states that, a steady current flowing through a metallic conductor is proportional to the potential difference between its ends, provided the temperature remains constant. If  $I$  is the current in ampere and  $V$  is the potential difference in volts then,  $I \propto V$ , at constant temperature, i.e.  $I = V/R$ , where,  $R$  is a constant known as resistance of the conductor. Ohm's law is applicable only for metallic conductors.

## RESISTANCE

Resistance is the property of a conductor by which it opposes the flow of electric current. It is defined as the ratio of the potential difference applied across a conductor to the current flowing through it:

$$\text{i.e., } R = V/I.$$

The device which offers resistance to the flow of current is called resister. In a conductor, the atoms are vibrating and the electrons move randomly. When a voltage is applied, the electrons move towards the positive terminal. During the process, they collide with vibrating atoms, resulting resistance. The unit of resistance is ohm ( $\Omega$ ). One ohm is the resistance of a conductor through which a steady current of one

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ampere passes, when a potential difference of one volt exists across it. In practice, kilo ohm ( $k\Omega$ ) and mega ohm ( $M\Omega$ ) are used as units and  $1\ k\Omega = 1000\ \Omega$ , and  $1\ M\Omega = 10^6\ \Omega$ .

When resistances are connected in series, the total resistance in the circuit is equal to the sum of the individual resistances. If three resistances,  $R_1$ ,  $R_2$  and  $R_3$  are connected in series, then the total resistance  $R$  is given by the relation:

$$R = R_1 + R_2 + R_3$$

If the resistances are connected in parallel, then the total resistance is given by the relation:

$$\frac{1}{R} = \left(\frac{1}{R_1}\right) + \left(\frac{1}{R_2}\right) + \left(\frac{1}{R_3}\right)$$

### SPECIFIC RESISTANCE

The resistance of a resistor at a given temperature depends upon the material and its dimension. The resistance ( $R$ ) is directly proportional to the length ( $L$ ) and inversely proportional to the area ( $A$ ) of cross section of resistor.

$$R \propto \frac{L}{A}, \quad R = \frac{\rho L}{A}$$

where,  $\rho$  is a constant, called the specific resistance or resistivity and the unit is ohm-meter. The specific resistance is given by:

$$\rho = RA/L$$

This relation reveals that the resistance of a thick wire would be lesser than that of a thin wire. The resistance is greater if the length is greater. The reciprocal of the resistivity ( $1/\rho$ ) is called the conductivity ( $\sigma$ ) and its unit is  $(\text{ohm-meter})^{-1}$ .

### SUPERCONDUCTIVITY

At very low temperatures, the resistivity of some materials (metals, compounds or alloys) becomes zero. Materials in such state are called superconductors. In general conductors, become superconducting at certain temperatures called super conducting transition temperature ( $T_c$ ). Current in a ring shaped super conducting material has been observed to flow for years in the absence of a potential difference, with no measurable decrease.

Superconductors are mainly used for generating very strong magnetic fields, since it has ability to withstand large magnetic fields and carry

large currents. A material in the super conducting state requires no power to carry large currents. Since the resistance of the material in the super conducting state is zero, no thermal losses are associated with passage of large currents.

Metal alloys like niobium-titanium become a superconductor below 10 K. It is used in magnetic resonance imaging equipment, but requires liquid helium as a coolant system. Ceramic metal oxide compounds become super conductors at higher temperatures, e.g. yttrium barium copper oxide at 93 K and thallium-calcium-bismuth copper oxide at 125 K. Since they operate at higher temperatures, liquid nitrogen can be used as effective coolant, which is cheaper than helium.

## **ELECTRICAL POWER**

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The electrical power ( $P$ ) is the rate at which energy is expended and it is equal to the product of potential difference ( $V$ ) and current ( $I$ ) in a circuit, i.e.  $P = VI$ . The unit of electrical power is watt, which is equal to one joule per second ( $J s^{-1}$ ). In practice, kilowatt and kilowatt hour (kWh) are used as units of electrical power and one kilowatt hour is equal to  $3.6 \times 10^6$  J.

## **HEATING EFFECT OF AN ELECTRIC CURRENT**

When electric current flows through a conductor having a resistance, certain amount of electrical energy is converted into heat energy. This heat energy will raise the temperature of the conductor. The above heat is produced by the free electrons as they move through the conductor. On their way, they collide frequently with atoms and give some of their kinetic energy to the atoms. The atoms which gains kinetic energy, generate heat in the conductor.

**Joule's law of heating:** The heat ( $H$ ) developed in a current carrying conductor is directly proportional to (i) the square of the current ( $I$ ) passing through the conductor, (ii) resistance ( $R$ ) of the conductor and (iii) time ( $t$ ) of flow of current

$$\text{i.e., } H = I^2Rt \text{ joule}$$

If the current is doubled, the heat generated is four times higher. This concept is applied in fuse wires, as the current goes to higher value, the heat generated in the circuit is sufficient to melt the fuse wire. The melting point of the fuse material is very critical, for material selection.

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### MAGNETISM

Magnetism is a fundamental property of a matter and it is produced by motion of electrical charges. As we are aware that electrons are in random motion in materials. The atoms and molecules that have paired electrons cancel their magnetic fields and the net magnetic field is zero. Whereas, the atoms with unpaired spinning of electrons in different shells, present net magnetic field. A magnet possess, two poles namely, north pole and south pole (Fig. 2.9). The term pole refers the end of the magnet in which the entire magnetism appears to be concentrated. The pole that points north under the influence of earth's magnetic field is called north pole and the other is called south pole. As in the case of electric charge, like poles repel and unlike poles attract each other. When a magnet is broken into two, each part became a magnet with north and south pole. The simplest magnet at the atomic level is the magnetic dipole.



**FIG. 2.9:** Magnetic dipole

The force of attraction ( $F$ ) between two magnetic poles is given by the relation:

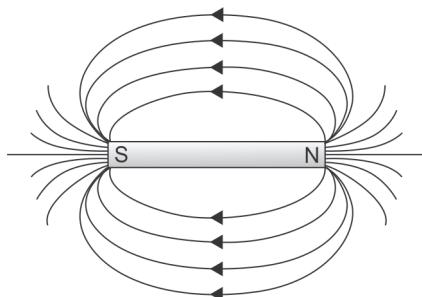
$$F = \frac{m_1 m_2}{4\pi\mu r^2}$$

where,  $m_1$  and  $m_2$  are the pole strength of two magnetic poles which are separated by distance  $r$ ,  $\mu$  is the absolute permeability of the medium, expressed in henry per meter ( $Hm^{-1}$ ). The permeability of free space is  $1.26 \times 10^{-6} Hm^{-1}$ . The unit of pole strength is weber (Wb).

### MAGNETIC FIELD AND FLUX DENSITY

The space around a magnetic pole in which another pole experiences a force is called the magnetic field. Magnetic fields can be visualized by magnetic lines of forces, which are imaginary lines (Fig. 2.10). A line of force can be defined as the path taken by an independent north pole moving from the north pole of magnet to south pole. The total number of lines of force used to express pole strength is called the magnetic flux. The term magnetic flux density is defined as the magnetic flux per unit area and its unit is  $Wbm^{-2}$ . The SI unit of magnetic flux is tesla (T) and one tesla =  $1 Wbm^{-2}$ . The older unit of pole strength

is gauss (G) and one tesla is equal 10,000 gauss. The earth's magnetic field strength varies from 0.5 to 1.0 G.



**FIG. 2.10:** Magnet and magnetic lines of force

## MAGNETIC INDUCTION

If a material is placed in a magnetic field, magnetism may be induced in that material by the magnetizing force. The atoms of the material tend to align with the direction of magnetizing force and induce a magnetic flux within the material. If  $H$  is the magnetizing force that induces a magnetic flux  $B$  in the material, then,

$$B = \mu H$$

where,  $\mu$  is the permeability of the medium =  $\mu_r \times \mu_0$ , where  $\mu_r$  is the relative permeability and  $\mu_0$  is the permeability in vacuum.

## MAGNETIC PROPERTIES

The magnetic properties of materials are determined by the atomic and molecular structures in relation with electron behavior. Magnetic susceptibility is one of the property, which describes the extent to which the material becomes magnetized, when placed in a magnetic field. Based on susceptibility, the material can be classified as diamagnetic, paramagnetic and ferromagnetic substances.

### ***Diamagnetic Substances***

They have negative susceptibility, e.g. calcium, water, and organic materials. In these materials, the orbiting electrons do not lie in a given plane and the net magnetic field is so small to be measured. When they are placed in an external magnetic field, the electron motions are altered, by the induced electromotive force. As a result, a reverse magnetic field is created, which will oppose the applied magnetic field. Thus, diamagnetic materials tend to reduce the applied magnetic field.

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### Paramagnetic Substances

They have slightly positive susceptibility, which tend to move from weaker to stronger parts of the magnetizing field. They will enhance the local magnetic field, but do not have measurable self magnetization, e.g. molecular oxygen ( $O_2$ ), blood degradation products, and gadolinium contrast agents. Paramagnetic materials have unpaired electrons (odd number), which have a magnetic field. When placed in an external magnetic field, the electron magnetic field align themselves with the applied field, enhancing the applied field.

### Ferromagnetic Substances

They are those, which are attracted by magnets and also be magnetized, e.g. iron, cobalt and nickel. The susceptibility of these materials is very high. Ferromagnetic materials are basically transition elements; in which electron fill the outer orbital shells, before inner shells are completely filled. When the spins are in random motion, usual cancellation of field do not take place, resulting in higher magnetic moment. When they are placed in an external magnetic field, the magnetic dipoles non-randomly aligned with applied field. Thus, ferromagnetic materials enhance the applied magnetic field.

## MAGNETIC EFFECT OF AN ELECTRIC CURRENT

When current is passed through a straight wire, a magnetic field is produced. This magnetic field is in cylindrical form and the lines of force form concentric circle about the conductor. A compass needle placed very close to the conductor gets deflected. If the direction of the current is reversed, the compass needle also gets deflected in the reverse direction (Fig. 2.11A). This shows that the current carrying conductor produces magnetic field, which exert a force on the compass needle. This is an example for conversion of electrical energy into mechanical energy. This effect is called the motor effect.

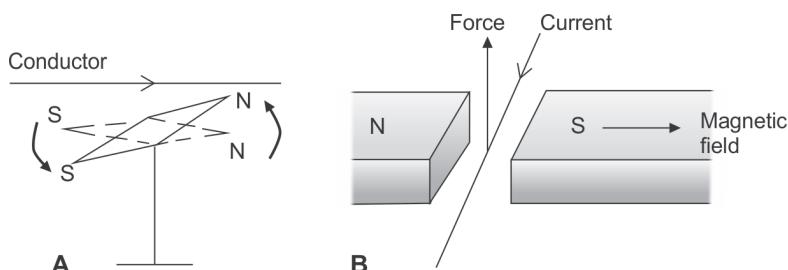


FIG. 2.11: (A) Compass needle and (B) Motor effect

The motor effect can also be obtained by the following method. Consider a conductor, which lies at right angles to the magnetic field. The magnetic field is produced by the fixed pole pieces as shown in Figure 2.11B. If current is passed through the conductor, it produces a magnetic field around the conductor. This magnetic field interacts with the original magnetic field and a force is developed. As a result, the conductor moves in a direction which is perpendicular to both the magnetic field and current. The direction of movement (force) of the conductor is determined by the Fleming's left hand rule.

### MAGNETIC FIELD DUE TO A COIL AND SOLENOID

When current is passed through a coil, each electron produces anticlockwise magnetic lines of force about itself. Within the coil, the lines of force is in one direction; whereas outside the coil, it is in the opposite direction. The lines of force is similar to that produced by a bar magnet. Thus, a coil-carrying current, give effects the same way as that of a bar magnet.

A solenoid consists of several coils joined together, usually made by winding insulated copper wire around soft-iron

core (Fig. 2.12). The solenoid may be considered as equivalent to a large number of circular loops placed in contact with each other. When current is passed through the solenoid, magnetic field is produced and the solenoid behaves just like a bar magnet.

The magnetic field of the solenoid also magnetizes the soft iron. The magnetic field of the iron core adds with that of the solenoid. As a result, many hundred times stronger magnetic field is obtained. This arrangement is called an electromagnet. One of the advantages of the electromagnet over the permanent magnet is that the former can be switched on and off. This principle is used in relay which is called the contactor in X-ray circuits.

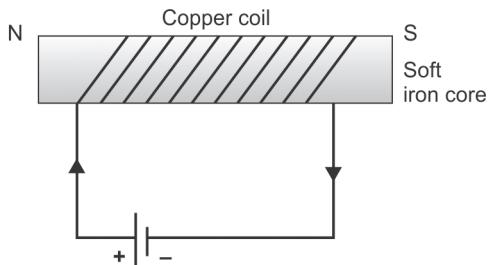


FIG. 2.12: Current-carrying solenoid or electromagnet

### ELECTROMAGNETIC INDUCTION

In 1831, Michael Faraday showed the production of current from a magnetic field. According to Faraday, current can be produced in a

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closed conductor, whenever, there is a change in the magnetic flux passing through the conductor. The current exists only so long as the change takes place. The current produced in the conductor in this way is called an induced current. The electromotive force (emf), that produces the current is called an induced emf. The whole phenomenon is known as electromagnetic induction.

### FARADAY'S EXPERIMENTS

Consider a coil connected to a galvanometer (Fig. 2.13). If a bar magnet is moved into the coil, the galvanometer will show a deflection in one direction. If the magnet is withdrawn from the coil, the galvanometer will show the deflection in the opposite direction. This shows the production of induced emf in the coil, whenever, the magnet is moved. Reversal of the movement results in a reversal of the emf. The reversal of the emf can also be produced by reversing the magnet or by reversing the direction of winding of the coil. If the magnet is moved more quickly relative to the coil, a larger emf will be induced. Similarly, a stronger magnet will induce a larger emf. If the experiment is tried with a coil of more number of turns, the induced emf will be larger.

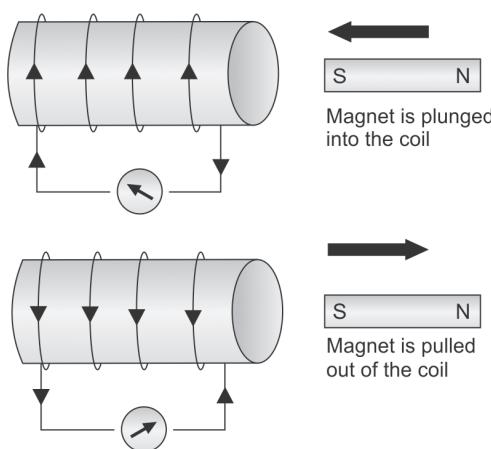


FIG. 2.13: Electromagnetic induction

In the above experiment, when the magnet is moved, the magnetic flux linked with the coil changes. Hence, an emf is induced in the coil. If the magnet is stationary, there is no change in magnetic flux, and hence no induced emf. Instead of moving a magnet, one can also use current to perform electromagnetic induction. If the current passing through a closed circuit is switched on or off, the magnetic flux will change. As a result, current will be induced in the neighboring circuit.

## LAWS OF ELECTROMAGNETIC INDUCTION

- i. A change of magnetic flux linked with a conductor induces an electromagnetic force (emf) in the conductor.
- ii. The magnitude of the induced emf is proportional to the rate of change of magnetic flux and to the area of the circuit.
- iii. The induced emf is in such a direction that it always opposes the change of magnetic flux, which induced the emf.

The first two laws are called Faraday's law, while the third is called as Lenz's law. Let  $\phi$  is the magnetic flux linked with a coil of N turns, and the induced emf ( $e$ ) is given by

$$e \propto -N(d\phi/dt)$$

where,  $d\phi/dt$  is the rate of change of flux linked with the coil. The negative sign represents the Lenz's law. The direction of the induced emf may be predicted by using Fleming's right hand rule. To apply this rule, one's right hand is arranged so that the thumb, the forefinger, and the middle finger are at right angles to each other. If the thumb denotes the motion of the conductor, the forefinger denotes the magnetic field, then the middle finger will denote the direction of the induced emf.

## SELF INDUCTION

In a single coil or solenoid, if there is a change in the magnetic flux, an emf will be induced in the same coil. This phenomenon is known as self induction (Fig. 2.14). Consider a coil of N turns, carrying a current of  $I$  and  $\phi$  be the magnetic flux linked with the coil. Then  $\phi \propto I$ , and the induced emf,  $e$  is given by:

$$\begin{aligned} e &\propto -N(dI/dt) \text{ or} \\ e &= -L N (dI/dt) \end{aligned}$$

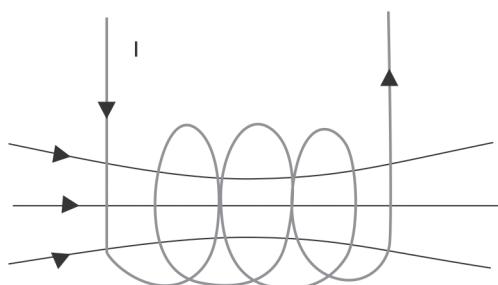


FIG. 2.14: Self induction

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where, L is a constant called coefficient of self inductance. The unit of self induction is henry. One henry is the inductance of a circuit in which an emf of 1 volt is induced, when the current in it changes at the rate of 1 ampere per second.

### MUTUAL INDUCTION

When two coils are placed very close to each other, the magnetic flux change in one coil induces an emf in other coil. This phenomenon is known as mutual induction (Fig. 2.15).

Consider two coils P and S placed close to each other. If a current is passed through P, a change in magnetic flux takes place. This will induce an emf in the coil P, which, in turn induces an emf in the coil S. Let  $\phi_1$  be the flux in the coil P due to current  $I_1$  flowing in it and  $\phi_2$  be the flux induced in the coil S due to the  $\phi_1$  in the coil P, then,

$$\phi_2 \propto I_1 \text{ and } \phi_2 = MI_1$$

where, M is a constant, called coefficient of mutual inductance for the two coils. If  $e_1$  and  $e_2$  are the emf's in P and S then,

$$e_2 \propto -N(d\phi_2/dt) \text{ or}$$
$$e_2 = -MN(dI_1/dt)$$

Hence, the coefficient of mutual induction of the given pair of coil is equal to the emf induced in one coil, when the current through the other coil changes at the rate of one ampere per second. This principle is applied in the transformer. The unit of mutual induction is also henry.

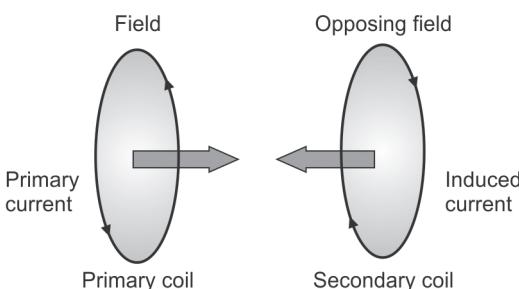


FIG. 2.15: Mutual induction with two coils

### ALTERNATING CURRENT

Current can be classified into direct current and alternating current. A constant current which is flowing in only one direction is called the

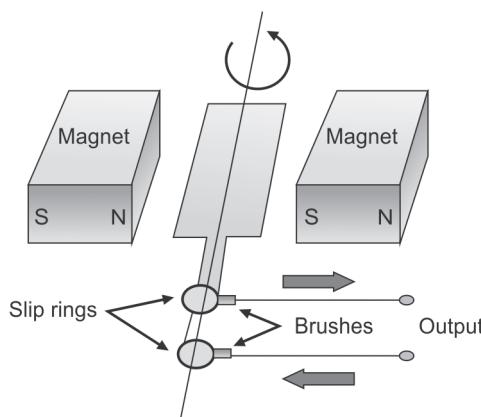
direct current. Cells or batteries are used to produce direct current in a small quantity. In the cells or batteries, chemical energy is converted into electrical energy. However, the production of large quantity of direct current is very costly.

A varying current, which reverses direction periodically, is called alternating current. This is generated in power stations, making use of the phenomenon electromagnetic induction. The machine used in the production of alternating current is called AC generator or alternator. Alternating currents are widely used because it is less costly.

### AC GENERATOR

The AC generator makes use of the principle of electromagnetic induction. It consists of four main parts, namely, armature, magnet, slip rings and brushes (Fig. 2.16). It is designed in such a way, so that a conductor cuts the magnetic lines of force continuously. A permanent magnet produces a parallel magnetic field between pole pieces N and S in the case of low power systems. Electromagnet is used in the case of high power generators.

The armature rotates between the poles of a strong magnet and this cut the magnetic lines of force from north pole to south pole. The soft iron core serves two purposes: (i) it serves to support the coils and (ii) increases the magnetic field, by substituting air with iron. The armature is mounted on a soft, which is driven continuously by means of a pulley, which is rotated by a engine through the belt.



**FIG. 2.16:** Alternating current generator

The two slip rings are metal rings to which the ends of the armature coil are connected. These rings are fixed to the soft, which rotates

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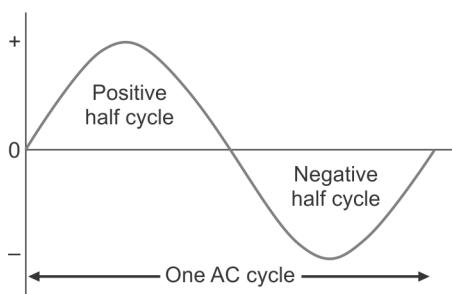
the armature coil. The slip rings also rotate along with the armature. There are two flexible metal plates or carbon brushes, which are fixed and constantly touch the revolving slip rings. With the help of these brushes current is passed out to the external circuits from the armature.

As the armature rotates at a constant speed, a varying emf will be induced across the ends of the coil. The induced emf will be at a maximum, when the magnetic lines of force cut the coil windings at right angles ( $90^\circ$  and  $270^\circ$  position). The emf will be zero when the magnetic field is parallel to the coil windings ( $0^\circ$  and  $180^\circ$ ). Thus, the induced current changes its direction after every half rotation.

The emf values can be plotted against the position of the armature in a graph. It can be seen that the emf starts at zero, reaches a positive maximum, or peak, drops to zero, goes to negative peak then return again to zero. The whole series of operations is called one cycle (Fig. 2.17). This emf is called an alternating emf and the current it would produce is called the alternating current. This emf is also called as sinusoidal voltage because; the emf is proportional to the sine of the angle of rotation. The induced emf at any instant time  $t$  is given by

$$e = E_0 \sin \omega t$$

The corresponding alternating current is given by  $i = I_0 \sin \omega t$ , where,  $E_0$  and  $I_0$  denote the peak value of the emf's and the current respectively,  $\theta = \omega t = 2\pi v$  and  $v$  is the frequency of the alternating current. The term,  $\omega t$  is the phase angle.



**FIG. 2.17:** Alternating current waveform

The frequency of the alternating voltage is the number of cycles made in one second. If the armature rotates 50 times per second, the frequency will be 50 cycles per second (hertz, Hz). The phase of an AC voltage or current is the fraction of the cycle that has elapsed,

since the starting of the voltage or current waveform. It is expressed in terms of angle of armature rotation.

### Peak and RMS Values

Since the alternating voltage is changing continuously, it is usually referred by instantaneous values. The instantaneous values of 'e' and 'I' merely give the value at any specified instant in time. But the instantaneous values are of limited application in normal use. The peak values  $E_0$  and  $I_0$  are the maximum instantaneous values. Since the peak value occur only for two instants in each cycle, its application is also limited.

The most useful measure of alternating quantity is root mean square value (rms). The rms value of an alternating current is that value of direct current which has the same average heating effect as the alternating current. The relation between the peak and rms values are as follows:

$$\text{rms voltage } (E_{\text{eff}}) = \text{Peak voltage } (E_0)/\sqrt{2}$$

$$= E_0 \times 0.707$$

$$\text{rms current } (I_{\text{eff}}) = \text{Peak current } (I_0)/\sqrt{2}$$

$$= I_0 \times 0.707$$

whenever the voltage and current values are given without any mention, it refers to rms value only.

AC ammeters and voltmeters are calibrated to give effective values of current and voltages respectively. When it is said that the electric main line is 220 V, 5 A, it means that the effective value of voltage and current are 220 V and 5 A respectively. A voltage rating of 220 V means that the incoming sinusoidal voltage has peak value of  $220 \div 0.707 = 310$  V. A fuse having a current rating of 5 ampere indicates that the current through the fuse wire can have a peak value of  $5 \div 0.707 = 7.07$  amperes.

### Power in AC Circuit

In AC circuit, the emf and the current vary continuously and also there is a phase difference ( $\theta$ ) between them. To estimate power at any instant, the power for a complete cycle is calculated. The average power of the circuit containing only resistor is given by,

$$P_{\text{av}} = E_{\text{eff}} \times I_{\text{eff}}$$

The average power for a circuit containing a resistance (R) and inductance (L), over a full cycle of AC is given by,  $P_{\text{av}} = E_{\text{eff}} \times I_{\text{eff}} \cos \theta$ , where,  $\cos \theta$  is known as the power factor of the AC circuit.

$$\text{The power factor, } \cos \theta = R/\sqrt{R^2 + (L\omega)^2}$$

where,  $\omega = 2\pi f$  and f is the frequency of the alternating current.

### THREE PHASE ALTERNATING CURRENT

Every country requires wide variety of AC power supply, to meet its demands. This can be achieved by having more than one phase of AC supply. Instead of single coil, three coils are used, to design a three phase AC generator. In this three symmetrical windings, separated by an angle of  $120^\circ$ , are mounted on a soft. Though the coils are mounted on the same soft, but connected to three separate pairs of slip rings. If the soft is rotated, each coil in turn moves through the magnetic field and induce three alternating voltages. Since the coils are inclined at  $120^\circ$  to one another and each having its pair of slip rings, the emf and the currents in the three coils will differ in phase by  $120^\circ$ . Such a generator is called a three phase generator and the current produced by it is three phase AC (Fig. 2.18).

Three phase supply provides more constant voltage to the circuit. When X-ray units are provided with three phase supply, it gives short exposure times. However, it is very expensive, difficult to install due to bulky hardware.

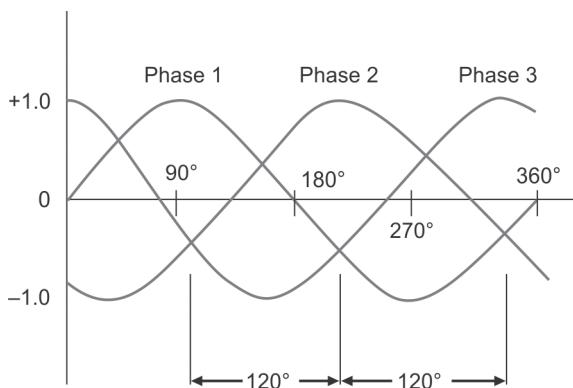
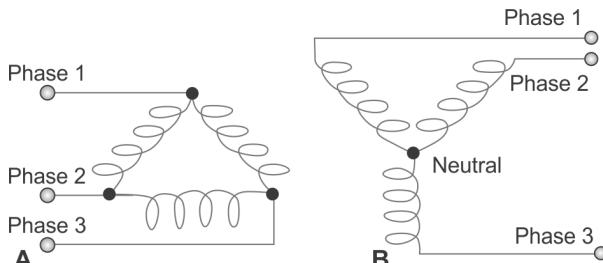


FIG. 2.18: Three phase AC waveforms

### Three-phase Supply Connections

The three phase supply is connected in two ways namely, star or wye connection and delta connection (Fig. 2.19). In the star connection, one end of each of the three windings is connected to a common center point, which is earthed (neutral). The other three terminals are used for connections, which are called line conductors. If equal load is given in all the three lines, the current flowing towards the center

in one of the windings is equal to the sum of the currents flowing away from the center in the other two. In practice, the electrical loads placed on the three phases may not be equal. Therefore, the fourth line, neutral acts as a common return path for current flowing from any of the other winding.



**FIG. 2.19:** (A) Delta and (B) Wye connection circuit

In delta circuit, the coils are connected like a triangle. Since triangle is similar to the Greek capital letter delta, this is called as Delta connection. The star connection has certain advantages over the delta connection as follows: (i) it is cheaper to produce and less stress and liability to breakdown, (ii) provide two different voltages, and (iii) reduces the number of conductors required for electrical transmission.

### CHOKE COIL

A choke coil consists of a large number of turns of insulated conducting wires wound over a soft iron laminated core. Its self inductance ( $L$ ) is very high and it offers a very high resistance to the flow of AC, thereby reducing the value of the current. It plays the same role in AC circuits as resistance in DC circuits. For many purposes, it is required to reduce the current in a given circuit, without wastage of energy, when the voltage is constant. If the resistance of the choke is  $R$  and the inductance is  $L$ , then the power factor

$$\cos \theta = R / \sqrt{R^2 + (L\omega)^2}$$

If a pure inductance having no resistance is used ( $R = 0$ ), then the power factor is zero and the power absorbed by the inductance is zero and the impedance offered is equal to  $L\omega$ . Even though there is no wastage of energy, the current is reduced due to the inductive reactance  $L\omega$ . The only wastage of energy is due to hysteresis loss in the iron core but it is very less compared to  $I^2 R$  losses.

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Choke coils are used both in low frequency (audiofrequency) currents and in high frequency (radiofrequency) currents. A typical radio frequency choke coil consists of one or more coils mounted on an insulating rod. A long single layer solenoid is also sometimes used as choke coil. Choke coils are used in radio sets, mercury lamps and sodium vapor lamps in AC circuits.

### **EDDY CURRENT**

When a conducting metal is placed in a varying magnetic field, current will be induced in the metal. This current will flow in closed paths inside the material. This current is called the eddy current or Foucault current. The production of eddy current leads to loss of energy due to heating.

Eddy currents tend to flow at right angles to the direction of field. Eddy currents cannot be completely eliminated but can be minimized by using metallic conductor in the form of thin sheets, called lamina insulated from one another. The core of dynamos, transformers and electric motors are laminated, to reduce the eddy current formation.

Eddy current has variety of applications: (i) In galvanometer, it is used to stop the motion of the oscillating coil, (ii) eddy current brakes are used in stopping electric trains, (iii) in induction furnace, it is used to produce heat, in order to melt and separate a particular metal from the ore, and (iv) the speedometer of the car involves eddy current principle.

### **ELECTRIC POWER TRANSMISSION**

In the transmission of electricity, transformers play an important role. Power plants are often situated at far off places, electricity must be transmitted over long distances, and there is always some power loss in the transmission lines. This loss can be minimized if the power is transmitted at high voltage.

Suppose an electric power  $P$  has to be delivered at a potential difference  $V$  by supply lines of total resistance  $R$ , then

$$\text{Current } I = P/V \text{ and}$$

$$\text{Power loss} = I^2 R = (P/V)^2 R$$

It is clearly seen here that, higher the potential difference, smaller will be the power loss. It is for this reason, the power is usually transmitted at very high voltage. By using a thick wire, resistance can also be minimized.

A typical power generator gives an output of 1000 kW at 6600 V. In practice, this voltage is stepped up to 132000 V before transmission. The cable used for transmitting power over long distances, are suspended by the large porcelain insulators from large steel structures called pylons. The main transmission lines from power station form part of a common system called the grid. Power from all the power stations in the region is fed into the grid, which covers a large region of the country. This forms a common pool from which power can be drawn. This allows an efficient power distribution and acts as a safeguard for ensuring a minimum power supply to the consumers in the event of failure of power generation at some station.

From the grid, the power is fed to the cities at 33000 V by using a step down transformer. At the substation, the supply is again stepped down to 6600 V. Power is supplied to the big factories at this voltage and they step down it according to their needs. For ordinary domestic consumers, the voltage is reduced to 220 V by using step-down transformers.

# 3

# Physics of X-rays

## **DISCOVERY OF X-RAYS**

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X-rays were discovered by WC Roentgen, the German physicist in 1895 when he was investigating the conduction of electricity through gases at low pressure in glass tubes. He noticed that the positive electrodes in the tubes gave off invisible rays which made fluorescent screens (Barium platinocyanide screen kept near the tube) to glow and fogged photographic plates. The rays were highly penetrating, they passed through black paper and even thicker objects. They were not deflected in magnetic field. Therefore, Roentgen concluded that they were not charged particles. As their nature was not known he called them X-rays; later, they were shown to be electromagnetic radiation of very short wavelength. Roentgen received the first Nobel prize in physics in 1901 for his discovery.

## **PROPERTIES OF X-RAYS**

1. X-rays are electromagnetic radiation of shorter wavelength (few nm).
2. They travel in straight line with a velocity equal to light.
3. X-rays are not influenced by electric and magnetic fields.
4. X-rays penetrate through substances that are opaque to visible light.
5. X-rays produce fluorescence in materials like calcium tungstate, and cesium iodide, etc.
6. X-rays affect the photographic film and form latent image.
7. X-rays produce ionization and excitation in the substances through which they pass.
8. X-rays produce chemical changes in substances through which they pass.
9. X-rays produce biological effects in living organisms. The cells can be either damaged or killed due to X-ray exposure.

## PRODUCTION OF X-RAYS

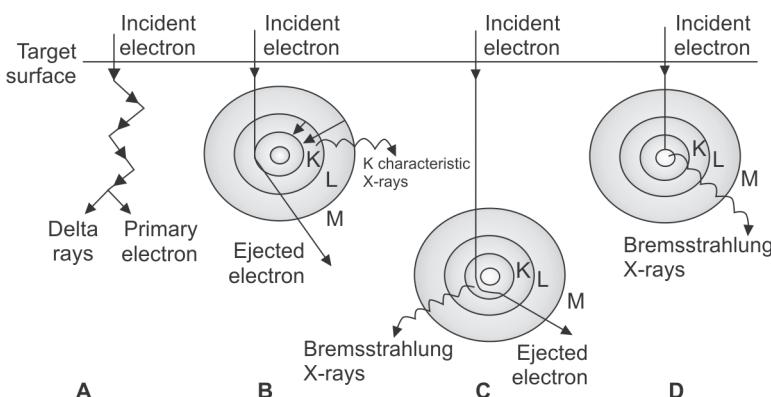
X-rays are produced when fast moving electrons are stopped by means of a target material. The moving electrons possess kinetic energy. When the electron is suddenly stopped, its kinetic energy is converted into heat and X-rays. This conversion is taking place in the target material. Therefore, the interaction of electron with the target is the basis for X-ray production.

### ELECTRON INTERACTION WITH THE TARGET

When the electron arrives at the target, it interacts in four ways as follows (Fig. 3.1).

The electron interaction involves ionizational collisions (i) and radiative collisions (ii), (iii) and (iv).

- i. Ionization of target atoms: The fast moving electron enters the surface layer of the target and undergoes collisions. In this process, the incident electron transfers sufficient energy and removes an electron from the atom. This involves small energy transfer, resulting in ionization of target atoms. The incident electron may undergo number of such collisions and each time its direction gets altered. A 100 keV electron may encounter 1000 of such interactions, before coming to rest and most of its energy appears as heat in the target. The displaced electron, known as a secondary electron, may have sufficient energy and produce further ionization of target atoms. They are few in number and produce their own track, known as delta rays.



**FIG. 3.1:** Interaction of electron with target atoms: (A) Ionization of target atoms, (B) Characteristic X-rays, (C) Interaction with nuclear field, (D) Interaction with nucleus

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- ii. Characteristic X-rays: This is an interaction between the incident electron and the electron in the K shell. In this process, the incident electron directly hit the K shell, transfers sufficient energy and removes the K shell electron. The vacancy in the K shell is filled by an electron moving inwards from the outer shell. During this transition, the difference in binding energies of the two shells is given out as X-ray photon. This photon is known as the characteristic X-ray. The ejected electron may produce further interaction in other target atoms.
- iii. Interaction with nuclear field: The incident electron occasionally reaches nearer to nucleus of an atom in the target. Since the electron is a negative particle, it is attracted by the positive nucleus. It is made to orbit partially around the nucleus, decelerates and goes out with reduced energy. The loss of energy appears in the form of X-ray photons, known as Bremsstrahlung. The energy of the X-ray photon depends on the degree to which the electron is decelerated by the nuclear attraction. The photon energy can take any value from zero to a maximum. This process is unlikely at low energies, but dominant at high energies.
- iv. The electron may hit the nucleus directly and is stopped completely in a single collision. The entire electron energy appears as bremsstrahlung radiation. This type of interaction is very rare, but capable of giving high energy X-rays.

In general, the interaction of B, C and D are very rare in the diagnostic range of energies, leading to lesser amount of X-ray production. The ionizational collision dominate ( $> 99\%$ ) the interaction process and produce heat. Thus, a X-ray tube is inefficient in the conversion of electron energy into X-rays.

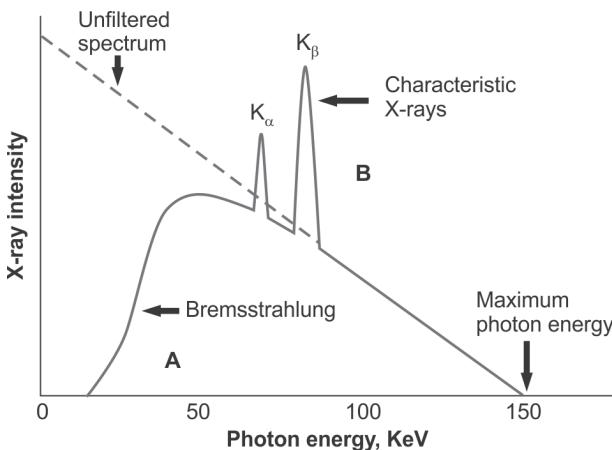
### X-RAY SPECTRA

X-ray photons produced by an X-ray tube are heterogeneous in energy. There are two types of X-ray spectrum, namely, (a) bremsstrahlung or continuous spectrum and (b) characteristic spectrum. A bremsstrahlung spectrum consists of X-ray photons of all energies up to maximum in a continuous fashion, which is also known as white radiation, because of its similarity to white light. A characteristic spectrum consists of X-ray photons of few energy, which is also called as line spectrum. The position of the characteristic radiation depends upon the atomic number of the target.

The intensity of the X-rays can be plotted against photon energy in a graph (Fig. 3.2). The area under the curve is proportional to the total number of photons emitted. The highest X-ray energy is determined by the peak voltage ( $kV_p$ ) applied in the X-ray tube. The characteristic spectrum is superimposed on the continuous spectrum. An unfiltered beam spectrum (theoretical) will be a straight line and mathematically given by Kramer's equation

$$I_E = KZ (E_m - E)$$

where,  $I_E$  is the intensity of photons with energy  $E$ ,  $Z$  is the atomic number of the target,  $E_m$  is the maximum photon energy and  $K$  is a constant. The unfiltered X-ray spectrum looks like a ramp.



**FIG. 3.2:** X-ray spectrum, (A) Bremsstrahlung and (B) Characteristic spectrum

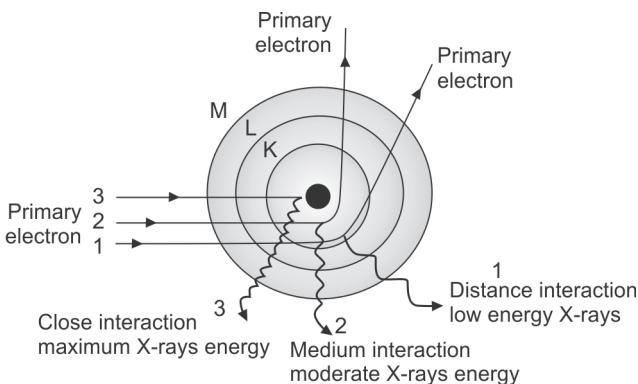
In practice, the X-ray beam is a filtered beam, due to inherent and added filtration. The filtration hardens the beam, by absorbing the low energy X-rays up to 10 keV, which is evident by the bremsstrahlung spectrum. The number of photons increases initially, with photon energy and later decrease linearly up to maximal photon energy. The X-ray spectrum is influenced by applied voltage, target material, tube current, exposure time, bremsstrahlung process and filtration. To specify the quality of X-rays, a rule of thumb is used, which states that the effective energy is about 1/3 to 1/2 of maximum X-ray energy.

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### BREMSSTRAHLUNG

The bremsstrahlung is a German word meaning braking radiation. It is a process of radiative collision between the electron and a nucleus in the target (Fig. 3.3). The electron while passing near the nucleus may suffer a sudden deflection and acceleration by the action of coulomb forces of attraction. As a result, the electrons may lose their kinetic energy, in the form of bremsstrahlung X-rays. The electron may have one or more such interactions and this may result in partial or complete loss of energy.

The amount of bremsstrahlung production is determined by the distance between the bombarding electron and the nucleus. At very large distance, the columbic force is weak, only low energy X-rays are created, but this process has higher probability to occur. When the electron is very close to the nucleus, columbic force is strong, electron lose more kinetic energy, resulting production of high energy X-rays. But this process has lower probability to occur. When the electron is in the middle, the electron interaction is moderate and the X-ray energy is also moderate. If the electron hit the nucleus directly, it lose all its kinetic energy, but the probability of this type of interaction is very low (5%). To conclude low energy X-rays are produced in greater abundance compared to high energy X-rays.



**FIG. 3.3:** Production of bremsstrahlung radiation

Thus, the bremsstrahlung radiation will have all possible energy from zero to maximum. The maximum energy is determined by the maximum kinetic energy of the incident electron. Also the direction of emission of bremsstrahlung photons depends on the energy of the incident electron. At electron energies below 100 keV, X-rays are emitted equally in all

directions. As the kinetic energy of the electron increases, the direction of X-ray emission becomes increasingly forward. In diagnostic radiology, it is technically advantages to obtain the X-ray beam on the same side of the target, i.e., at 90° with respect to the electron beam direction.

In diagnostic X-ray tubes, thicker targets are used, to stop the entire electron beam. Hence, X-rays are produced in all directions around the target. Those X-rays that are produced in the forward direction will be absorbed by the target itself. The term efficiency of X-ray production is defined as the ratio of output energy emitted as X-rays to the input energy deposited by the electron. It can be shown that

$$\text{efficiency} = 9 \times 10^{-10} \times Z \times V$$

where,  $Z$  is the atomic number of the target material,  $V$  is the tube voltage in volts. Thus, the bremsstrahlung X-ray production increases with accelerating voltage and atomic number of the target material. Alternatively, the X-ray production efficiency may be expressed in terms of radiative and collisional losses as follows:

$$\frac{\text{Radiative energy loss}}{\text{Collisional energy loss}} = \frac{E_k \times Z}{820,000}$$

where,  $E_k$  is the kinetic energy of the incident electron. Radiation loss is due to bremsstrahlung production, whereas collisional loss is due to excitation and ionization. If an electron with 100 keV energy interacts with a tungsten target ( $Z = 74$ ), then the above ratio =  $(100 \times 72)/820,000 = 0.9\%$ . Thus, the efficiency of tungsten target is found to be less than 1% and the rest of the input energy,  $> 99\%$  appears as heat. Higher the photon energy, greater the X-ray production efficiency, and lesser the heat production. The X-ray production efficiency is more than 50% for a 6 MV electron, resulting lesser production of heat.

## CHARACTERISTIC X-RAYS

Electron incident on a target may produce characteristic X-rays. An electron with kinetic energy  $E_0$  may interact with the atoms of the target, by ejecting an orbital electron from the K shell. Now, there is a vacancy in the K shell and the atom is said to be ionized. The original electron will have energy  $E_0 - E$ , where,  $E$  is the energy given to the orbital electron. The outer orbital electrons (from M or L) will fall down to fill the vacancy in the K shell (Fig. 3.4). In doing so, the difference in binding energy of the two shells is radiated as X-ray photons, which

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is called the characteristic radiation. This will have only discrete energies. Since the binding energy difference is unique to an atom, the X-rays emitted are characteristic of that element. If the transition involved an electron from L shell to K shell of a tungsten target, then the emitted photon will have energy:

$$\begin{aligned} h\nu &= E_K - E_L \\ &= 69.5 \text{ keV} - 10.2 \text{ keV} \\ &= 59.3 \text{ keV} \end{aligned}$$

where,  $E_K$  and  $E_L$  are the binding energies of the K and L-shells of tungsten atom respectively. The K-shell characteristic X-ray energies are slightly lower than the K-shell binding energy. Electron transition may be from adjacent and non-adjacent shells.

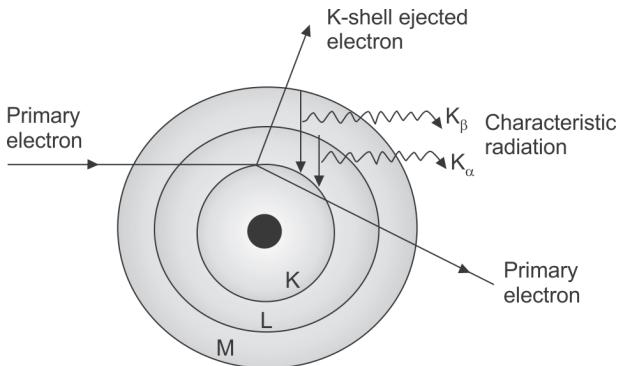


FIG. 3.4: Production of characteristic X-rays

Energy transitions are designated by the shell capturing the electron with a subscript of  $\alpha$  or  $\beta$ . The subscript  $\alpha$  refers to an adjacent shell transition, e.g.  $L \rightarrow K$  transition is denoted as  $K_\alpha$  X-rays. The subscript  $\beta$  refers to a non-adjacent shell transition, e.g.  $M \rightarrow K$  transition is denoted as  $K_\beta$  X-rays. Thus, transition results in fine energy splitting of the characteristic X-rays, due to subshells of the given orbit. Only K characteristic X-rays are very important in diagnostic radiology ( $K_{\alpha 1}$ ,  $K_{\alpha 2}$  and  $K_{\beta 1}$  transitions). The K-shell characteristic X-rays for various target atoms are given in Table 3.1.

The characteristic X-rays other than K-shell transitions are not important, since they are entirely attenuated by the tube window and filters. K characteristic X-rays are emitted only, if the incident electrons have energies greater than the binding energy of the K-shell electron. Hence, the kilovoltage applied must be greater than 69.5 keV for tungsten,

**TABLE 3.1** K-shell characteristic X-rays (keV) and target material

Transition	Tungsten, Z = 74	Molybdenum, Z = 42	Rhodium, Z = 45
K <sub>α1</sub>	59.32	17.48	20.22
K <sub>α2</sub>	57.98	17.37	20.07
K <sub>β1</sub>	67.24	19.61	22.72

20 keV for molybdenum and 23.2 keV for rhodium targets respectively, which is called as threshold energy. As the energy of the incident electron increases above the threshold energy, the % of characteristic X-rays also increases. The 100 kV<sub>p</sub> X-rays spectrum consists of about only 10% characteristic X-rays.

## X-RAY TUBE DESIGN

The production of X-ray needs the following: (i) electron source (cathode), (ii) target to stop the electrons (anode), (iii) high voltage supply to accelerate electrons, (iv) vacuum and (v) tube insert (glass envelope). Electron can be produced either by ionization in gas or by thermionic emission. The electron source acts as a cathode and the target acts as an anode. The high voltage is applied in between the cathode and anode. This voltage accelerates the electrons to a higher velocity; as a result the electron will possess high kinetic energy. When the electrons are stopped by a target, the electron kinetic energy is converted into X-ray energy and thus X-rays are produced. The equipment having all the above requirements is called an X-ray tube.

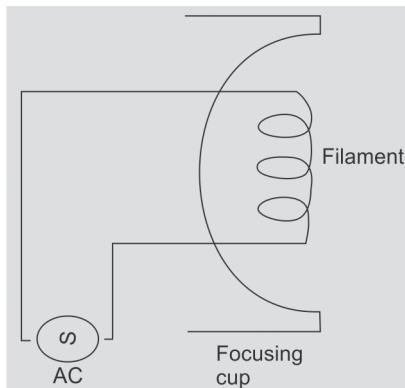
The tube should be designed in such way that, it should withstand voltages from 20–150 kV and currents up to 1000 mA. In radiography, tube current may vary from 100 to 1000 mA, whereas in fluoroscopy, it is 1–5 mA. In addition, exposure time has to be varied over a wider range.

## CATHODE

The cathode is made of tungsten wire in the form of helical filament, surrounded by a focusing cup (Fig. 3.5). Tungsten is used as filament material because of its high melting point, low vapor pressure, good ductility (easily drawn into fine wire) and low work function (4.5 eV). Tungsten exhibits thermionic emission well below its melting point. The filament is made of tungsten wire, about 0.2 mm in diameter,

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that is coiled to form a vertical spiral about 0.2 cm in diameter and 1 cm in length. The coil format provides large surface area for electron emission.



**FIG. 3.5.** Cathode assembly

The filament circuit supplies a voltage of 8–12 V and selectable filament current of 3–7 amperes. Electrical resistance to electron flow heats the filament to very high temperature, releasing surface electron through thermionic emission process. The rate of emission depends on the temperature and it can be adjusted by the filament current. A trace of thorium in the filament not only increases the efficiency, but also prolongs the filament life. If the applied voltage between anode and cathode is zero, the electrons form a cloud near the cathode, which is called space charge. As the applied voltage increases, the electrons are accelerated towards the anode, helps the production of X-rays.

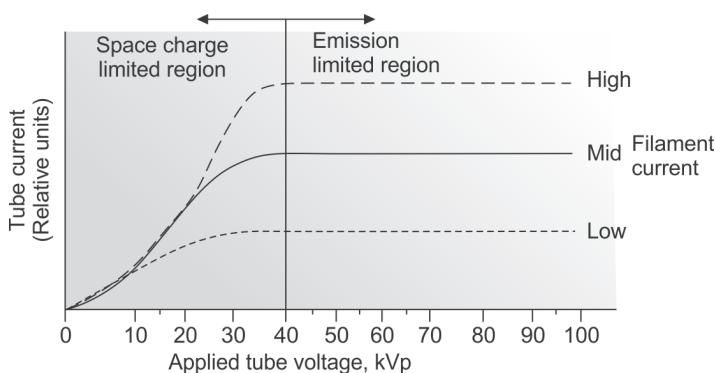
The focusing cup controls the width of the electron distribution, and directs the electron toward the target. Usually, the focusing cup is at the same potential as the filament, which is called nonbiased X-ray tube. Usually, X-ray tubes are provided with two filaments of different length. Selection of a particular filament determines the focal spot length or area.

### SPACE CHARGE EFFECT

To understand the operation of an X-ray tube, it is essential to know how the tube current depends upon the tube voltage, for a given filament excitation (Fig. 3.6). When the applied kV is zero or small, the electrons surrounding the filament forms a cloud, resulting in space charge effect. These electrons tend to repel electron back into the filament and hence

the tube current is very small. As the  $kV_p$  is increased, (0–40 kV) the effect of space charge reduces gradually and the tube current also increases. This is called space charge limited region. In this region, the tube current strongly dependent on applied  $kV$ , for a constant filament current.

Above 40  $kV_p$ , the space charge effect is overcome, and the tube current is controlled by the filament current. This is called the saturation or emission limited region. In this region, the tube current undergoes little change with an increase in tube voltage. The tube current is 5–10 times lesser than the filament current in this region.



**FIG. 3.6:** Space charge effect

Most of the X-ray tubes are operating in between the space charge region and saturation region. Thus, the tube current is determined by both  $kV$  and filament current. In the space charge region ( $< 40$  kV), the tube current is influenced by the applied voltage only. To deliver the selected tube current, a space charge compensating circuit is used. This circuit also corrects the small increase in tube current, at higher applied voltage ( $> 40$  kV).

The above characteristics of the tube (curves) are depend upon many factors, including the distance between anode-cathode, the configuration of focusing cup, the focal spot size, and the filament temperature. Particularly the change of potential of the focusing cup, will drastically alter the curves.

## ANODE

The anode is the target electrode, which is maintained at a positive potential. The target material should possess the following properties:

- high melting point to withstand high temperature,
- high atomic

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number to increase the X-ray production efficiency, (iii) high thermal conductivity to dissipate heat quickly, (iv) low vapor pressure at high temperature to prevent the evaporation of target material, and (v) easily machined to make smooth surface.

Tungsten (W) is the metal widely used as target because of its high melting point,  $3387^{\circ}\text{C}$  and high atomic number, 74. However, its thermal conductivity is low ( $174 \text{ Wm}^{-1}\text{K}^{-1}$ ) and hence tungsten is embedded over a thick block of copper. The thermal conductivity of copper is  $400 \text{ Wm}^{-1}\text{K}^{-1}$ , so that the heat will be removed very quickly to the surrounding. The vapor pressure of tungsten is 5000 kPa, which is low, by which it releases less vapor into the vacuum. In the stationary anode, the tungsten is a square or rectangular plate of 2 or 3 mm thick and dimension greater than 1 cm. However, rotating anode design employs disk of 75–200 mm diameter, with beveled edges. Large diameter anodes are used in CT and fluoroscopy, which increases the heat capacity and heat dissipation. However, they are prone for mechanical damage, which is prevented by making radial slots in the anode. The above type of anode is called stress relieved anodes.

The anode has a tendency to crack under severe stress caused by heating. Therefore, tungsten-rhenium alloy (90% tungsten + 10% rhenium) is always used, which makes the target tougher and reduces surface pitting. Molybdenum (Mo, Z = 42) and rhodium (Rh, Z = 45) are commonly used as anode materials for mammographic X-ray tubes. These targets are capable of giving characteristic X-rays, suitable for soft tissue contrast studies.

### **FOCAL SPOT SIZE**

The area of target with in which the electrons are absorbed and X-rays are generated is called focal spot or focal area. If the focal area is very small, penumbra will be lesser, and the picture sharpness will be good, but heat removal is difficult. On the other hand, if the focal area is large, heat will be removed quickly, penumbra is larger and the picture sharpness is bad. This can be compromised by careful design of the tube.

Usually, focal spot is defined in two ways namely, actual and effective focal spots. The actual focal spot size is the area on the anode that is struck by electrons. The effective focal spot size is the length and width of the emitted X-ray beam as projected down the central axis

of the X-ray tube. The effective focal spot length is always smaller than the actual focal spot. The relation between them is given as follows:

$$\text{Effective focal length} = \text{Actual focal length} \times \sin \theta$$

where,  $\theta$  is the anode angle. The focal spot size is usually expressed in terms of effective focal spot size, which varies from 0.3 mm to 2 mm square. The focal spot sizes of 0.3 mm, 0.6 mm, 1.0 mm, and 1.2 mm are commonly employed in radiology. The focal spot size has a major effect on spatial resolution, particularly when appreciable magnification of the object occurs. Focal spot size can be measured by using a pinhole camera, slit camera and star pattern.

### LINE FOCUS PRINCIPLE

X-ray tube requires a specialized focal area in the target, which is larger in size to spread the heat easily and smaller in size to act as a point source. The point source reduces penumbra effect resulting sharp images. Hence, the target of an X-ray tube is mounted at a very steep angle ( $\theta$ ) with respect to the motion of the incident electrons. The X-rays will appear to come from a small focal area (effective focus), whereas the electrons bombard relatively a larger focal area (actual focus). Therefore heat is removed very quickly and also the image sharpness is preserved. This is known as *line focus principle* (Fig. 3.7). Consider the electrons that are made to strike on a target of length  $ab$ , width  $cd$  and anode angle  $\theta$ , then

$$\begin{aligned}\text{effective focus} &= \text{actual focus} \times \sin \text{ of anode angle} \\ cd &= ab \times \sin \theta, \text{ for example, if } \theta = 17^\circ, \\ &\quad cd = 1 \text{ mm, } ab = 3 \text{ mm, then} \\ \text{effective focus} &= 3 \text{ mm} \times \sin 17^\circ \\ &= 3 \text{ mm} \times 0.2924 \\ &= 0.877 \text{ mm}\end{aligned}$$

In such tubes electrons bombard on a rectangular area of 3 mm  $\times$  1 mm, while the X-rays appear to come from an area of  $\approx$  1 mm  $\times$  1 mm.

Now, the loading gain is given by the relation,

$$\text{loading gain} = \frac{\text{actual focal area}}{\text{effective focal area}}$$

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$$= \frac{3 \times 1}{1 \times 1} = 3.0$$

If the  $\theta$  is made smaller the loading gain may be increased, but the angular width of the useful X-ray field is reduced.

### ANODE ANGLE

Anode angle is defined as the angle of the target surface with respect to the central ray in the X-ray field as shown in Figure 3.7. It has strong relationship with focal spot size and usable X-ray field size. A small anode angle gives smaller effective focal spot, but its usable X-ray field is limited. Large anode angle gives larger usable X-ray field, but the effective focal spot is larger. To optimize the design, larger anode angle with small filament length is used. This will provide smaller effective focal size, with wide field coverage.

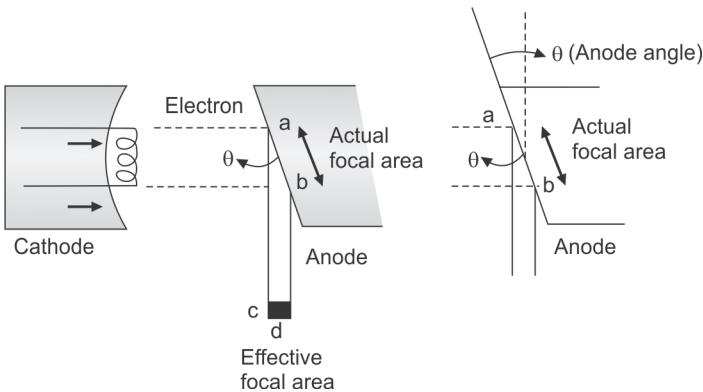


FIG. 3.7: Line focus principle

The optimum choice of anode angle depends on the clinical application. A small anode angle ( $7\text{--}9^\circ$ ) is useful in small field of view (FOV) imaging such as celiacangiography and neuroangiography, where the FOV is limited by image intensifier. Larger anode angles are necessary for general radiographic work to achieve larger FOV coverage at short focus to image distances (FID). Modern X-ray tubes are designed with anode angle of  $10\text{--}13^\circ$  with focal spot sizes of  $0.6\text{--}1.3$  mm.

### TUBE INSERT AND VACUUM

The tube insert or envelope is made up of borosilicate glass (Pyrex). The pyrex glass can withstand high temperature and also act as an electrical insulator. It contains vacuum, which support the electrodes.

The tube insert (i) absorbs the X-rays emerging in undesired directions, (ii) maintains the required vacuum, (iii) acts as an electrical insulator and (iv) also contain the cooling system which removes the heat from the target. Glass is not an ideal insert for X-ray tubes since tungsten vapor condenses and forms electrically conducting thin layer. This may lead to arcing and loss of vacuum due to puncture of the glass. Glass is also susceptible to damage from electron bombardment. Hence, metal envelope have been developed with low attenuation beryllium window ( $Z = 4$ ), for X-ray transmission. However, metal may short circuit cathode and anode due to its conductivity. To eliminate this, ceramic or glass insulations are done at the end of the tube. This type of envelope is called metaceramic or metal glass design.

A high vacuum is maintained between the anode and cathode. This is necessary (i) to avoid the collision between electrons and gas molecules, which gives raise to ionization that reduces the kinetic energy of the electrons, (ii) to prevent oxidation of electrodes and (iii) act as an electrical insulator. The required vacuum is less than  $10^{-5}$  mm Hg.

## TUBE COOLING

In a X-ray tube, only less than 1% of the electrical power supplied is converted to X-rays. The remaining electrical power (over 99%) is converted into heat. This large amount of heat may melt the target and therefore heat should be removed very quickly from the target. Hence, efficient cooling systems are necessary for the X-ray tube.

In general, targets are made by inserting a layer of tungsten in copper block, and X-ray tubes are usually enclosed in metal cases, which are filled with oil for insulation purpose. The heat produced on the focal area is conducted quickly into the anode disk, stored temporarily, later transferred to the insulating oil by radiation. This oil surrounds the glass envelope as well as copper block (static oil cooling). The heat taken up by the oil is transferred to the housing (metal case) by convection process. In some designs, fan is used to assist the convection process and removes the heat from the housing.

In the case of rotating anode tube, the molybdenum neck is so long and prevents the heat conduction to the rotor. If the anode assembly is coated black, it will promote the heat radiation process. The rate of heat radiation is proportional to fourth power of anode temperature. Hotter the anode, greater the rate of heat dissipation. For prolonged

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operation, the oil around the tube is connected by two pipes to a oil reservoir with radiator and pump, where the oil is cooled additionally by an air current and water (circulating oil cooling). In X-ray tubes used in CT scan and angiography work, oil is pumped through an external heat exchanger.

In some modern tubes, the anode is earthed and water is allowed to circulate through the anode. Sometimes the water is additionally cooled by Freon gas.

### HISTORICAL X-RAY TUBES

#### GAS TUBE

In the beginning, gas filled tubes were used to produce X-rays. These tubes consist of two electrodes called cathode and anode, kept in a sealed glass envelope, at opposite ends (Fig. 3.8). The cathode is an aluminum stem carrying a concave aluminum disc. The anode or target is tungsten or platinum backed with copper. The radius of concavity of the cathode is such that its center of focus lies on the surface of the target. The face of the target makes an angle  $45^\circ$  with the axis of the cathode neck. A provision is made to evacuate the tube, to different degrees of vacuum. The tube vacuum must be always less than 0.001 mm Hg.

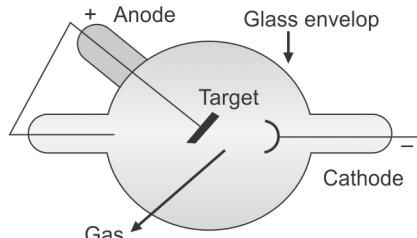


FIG. 3.8: Gas X-ray tube

In a gas tube, a small residue of air is always left. When a high potential is applied between the cathode and anode, the air inside the tube gets ionized and electrons are produced. These electrons move at higher speed towards the anode, and produce an avalanche of electrons by collision with air molecules. The positive ions left are attracted towards the cathode and release many more electrons by its impact.

When the electron stream strikes the target, X-rays are emitted. The relation between the voltage applied to the tube and the minimum wavelength of X-rays emitted are given by the *Duane and Hunt's law*

$$\lambda_{\min} = \frac{12.4}{kV_p}$$

The wavelength is expressed in angstrom ( $\text{\AA}$ ) units and  $1 \text{ \AA} = (10^{-10} \text{ m})$ .

### Defects of Gas Tubes

- i. Ionization potential of the gas depends upon the degree of evacuation of the tube. A tube with much residual gas will produce X-rays at lower voltages. If the voltage increases beyond ionization potential, the tube current increases enormously, and may melt the target. The whole tube may be filled with vapor of the target material, making the tube gassy.
- ii. The energy of the X-rays and the intensity of X-rays are interlinked. To increase the penetrating power, evacuation has to be increased, while applying higher voltage. Hence, a vacuum pump is always required through out the operation. Thus, the apparatus require more skill to operate.
- iii. As the tube operates, the ions gradually diminishes, resulting higher vacuum. Hence, fresh gas has to be injected into the tube by the use of gas regenerators. Because of these disadvantages, the gas tubes are now practically obsolete.

### COOLIDGE TUBE

Flaming and Richardson showed that, when a metal is heated in vacuum to incandescence, it emits large amount of electrons. Such a emission of electrons by the supply of heat is called thermionic emission.

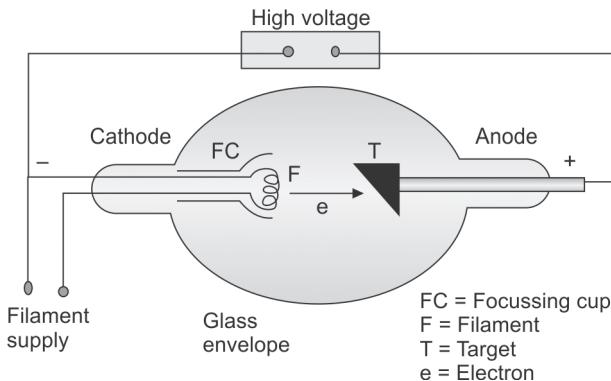
These electrons can be collected by using another electrode, kept at a positive potential. It was proved that all the emitted electrons can be collected at a particular potential called the saturation potential. Further increase of potential does not increase the current flowing between the electrodes.

A new X-ray tube was designed by Coolidge (1913), by using above thermionic emission principle. This was known as hot cathode or electron tube (Fig. 3.9). This tube is completely evacuated, so that there is no production of electrons by ionization. A spiral of wire, which serves as a hot cathode, is heated to incandescence by means of a current of 4 ampere and 10 volts. The filament commonly used is tungsten. The target is placed opposite to the cathode similar to that in a gas tube.

When the high voltage is applied between the cathode and anode, electron current flows from the cathode to anode. As the electron reaches the target, X-rays are produced. These tubes are always

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operated at saturation potential. The wavelength of the X-rays can be controlled by altering the voltage applied to the tube. The tube current can be varied by altering the temperature of the filament. Thus, the applied voltage and tube current are independent of each other, and can be controlled separately, which is the advantage of this tube. All the modern X-ray tubes are based on the principle of Coolidge tube only.



**FIG. 3.9:** The Coolidge X-ray tube

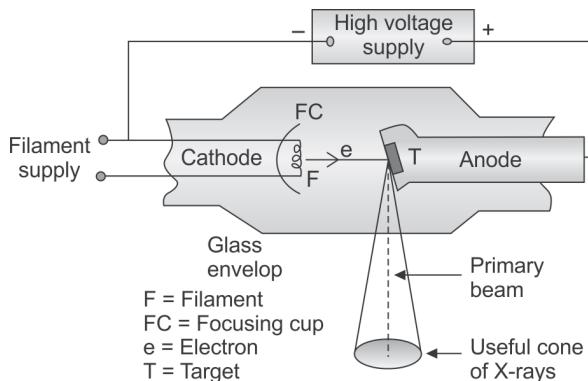
## MODERN X-RAY TUBES

### STATIONARY ANODE X-RAY TUBE

In earlier periods, gas tubes were used to produce X-rays. But, these tubes suffered with disadvantages. Hence, Coolidge proposed a prototype X-ray tube based on the thermionic emission principle. On the basis of Coolidge tube, several X-ray tubes have been designed. The stationary anode X-ray tube is one of the modern X-ray tubes in which the anode is stationary.

The stationary anode X-ray tube consists of a cathode and an anode which are kept in a evacuated glass envelope (Fig. 3.10). The cathode consists of a tungsten filament in the form of a coil placed in a shallow focusing cup. The filament is heated by passing an electric current through it from a low voltage supply.

The anode is made of copper block in which a small tungsten plate is embedded. The tungsten plate serves as a target. The target is positioned on line focus principle, in order to increase the ratio of the actual focal area to the effective focal area. The anode angle is usually 15–20°. A high voltage supply is applied between the cathode and



**FIG. 3.10:** Stationary anode X-ray tube

anode to accelerate the electrons. A vacuum of the order of  $10^{-5}$  mm Hg is maintained in the tube.

When the filament is heated to white light, it emits electrons. The focusing cup (made up of nickel) produces an electric field (negative) that focuses the electron to the focal area. The focusing cup also protects the adjacent parts of the tube wall from damage by electron bombardment. If the anode is made positive with respect to the filament, these electrons will be attracted to the anode. This will constitute an electron current around the circuit in the anticlockwise direction. The tube current is measured by the milliammeter (mA).

Since the space between anode and cathode is a high vacuum, the electrons do not collide with gas molecules in crossing the gap and so acquire very high velocities. The electrons which are accelerated by the applied voltage possess high kinetic energy. When they suddenly stop in the target, X-rays are emitted in all directions. About one-half of these are absorbed in the target itself. The remaining portion emerges as a useful primary X-ray beam. During the production of X-rays, large amount of heat is produced in the target. The tube is also provided with suitable cooling system to remove the heat very quickly.

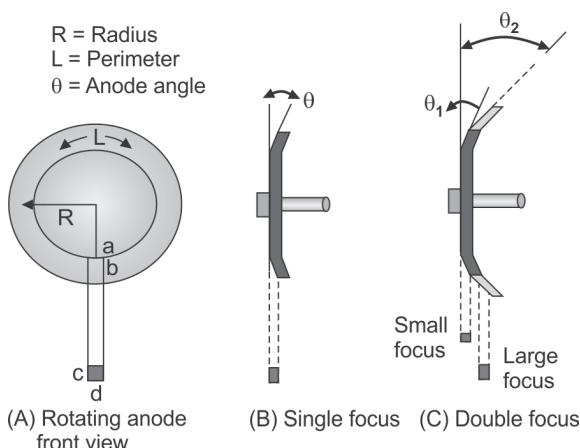
Stationary anode tubes have a small target area that limits the heat dissipation and this limit the X-ray output, but they are small in size and weight. Dental X-ray units (intraoral and ortho-pantomography), portable X-ray units, and portable fluoroscopy systems use stationary anode X-ray tubes.

### ROTATING ANODE X-RAY TUBE

In 1933, the rotating anode X-ray tube was invented, in which the anode is made to rotate before the electron is emitted. It was developed to increase the heat loading with higher X-ray output. In these tubes, the electrons transfer their energy over a large area of a rotating target. Rotating anode tubes are larger in size, but the principle and function are similar to that of stationary X-ray tube.

#### Principle

Consider a rotating anode of radius  $R$  and circumference  $L$  as shown in the Figure 3.11. The electrons bombard a region of height  $ab$  and width  $cd$ . The length may range up to the circumference ( $L = 2\pi R$ ) depending on the exposure time. But, the X-rays always appear to come from a focal spot of area  $cd \times cd$ .



**FIG. 3.11:** Rotating anode assembly: front and side view

Let us consider a stationary anode of actual focal area  $7.3 \text{ mm} \times 2 \text{ mm}$ , and rotating anode of  $R = 30 \text{ mm}$  and length  $7.3 \text{ mm}$ , then

$$\begin{aligned}
 \text{Loading gain} &= \frac{\text{Actual focal area of rotating anode}}{\text{Actual focal area of stationary anode}} \\
 &= \frac{2 \pi \times 30 \times 7.3}{7.3 \times 2} = 94.2
 \end{aligned}$$

Thus, the rotating anode arrangement helps to increase the loading to a greater extent of the order 100. The construction of such a rotating

anode is a remarkable technological development. The diameter of the tungsten disk determines the total length of the target track, and obviously affects the maximum permissible loading of the anode.

### Cathode

Rotating anode X-ray tube consists of a cathode and an anode which are kept in a glass bulb (Fig. 3.12). The cathode is a tungsten filament which is offset from the long axis of the X-ray tube to face the target near the periphery of the anode disk. Usually, rotating anode tubes are fitted with two filaments (Fig. 3.13), one larger than the other set side by side in the cathode assembly. One filament is designed to focus the electrons on a larger area of the anode, which require heavy tube loading. The other filament is used to focus the electrons on a

The diagram illustrates the internal components of a rotating anode X-ray tube. It shows a central vertical axis with a 'Cathode assembly' containing a 'Filament supply' at the bottom. A 'High voltage supply' is connected across the cathode and an anode. An 'Induction motor' drives a 'Rotor' which holds the 'Anode' (a large circular disk). The 'Anode' has a central 'Anode stem' (S) and is at a 'Rotating anode distance' (R) from the 'Target' (T). Electrons (e) from the filament are focused onto the target. A 'Glass envelop' surrounds the tube. Labels include F (filament focusing CL), e (Electron beam), R (Rotating anode distance), S (Anode stem), and T (Target). A 'Cone of X-ray beam' is shown originating from the target.

**FIG. 3.12:** Rotating anode X-ray tube

This diagram shows a cross-section of a cathode assembly. It features a circular 'Focusing cup' containing two vertical columns of 'filaments'. The left column is labeled 'Filament for small focus' and the right column is labeled 'Filament for large focus'. Both columns consist of multiple small, curved segments.

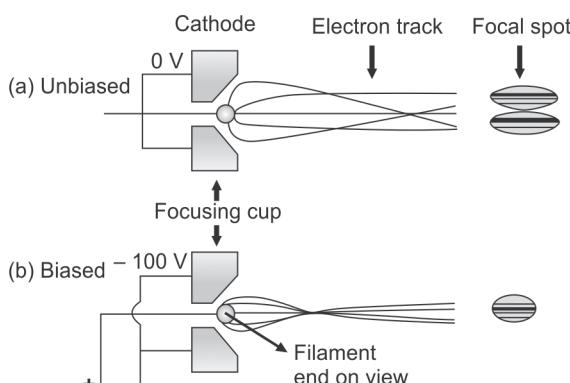
**FIG. 3.13:** Cathode with dual filament

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smaller area of the target. This type is used when high resolution is required. Both filaments should focus the electron on the same part of the anode, so that focal spot is at the same point for both modes of operation. Some tubes provide two target angles for two filaments, so that each filament will have a separate focal spot. Smaller angle is used with the smaller focal spot.

### Focusing Cup

The focusing cup (cathode block) surrounds the filament, shapes the electron beam width. It is used to focus the electrons on a small area (focal spot) in the anode. There are two ways by which the focusing cup is energized, namely, unbiased and biased (Fig. 3.14). In unbiased setup, same voltage is applied to both focusing cup and filament. In this type, the electron spread is wider and the focal spot width is larger. In biased X-ray tubes, insulated focusing cups are used and it is given more negative supply ( $-100$  V) than the filament. This creates a tighter electric field around the electron, reduces the electron spread and gives smaller focal spot width. Thus, the focusing cup width determines focal spot width and the filament length determines the focal spot length.



**FIG. 3.14:** (A) Unbiased focusing cup potential is 0 V, relative to filament,  
(B) Biased focusing cup, potential is -100 V, relative to filament

### Anode

The anode is made in the form of large disk of tungsten, or an alloy of tungsten, which is saucer shaped. The target track is near the periphery of the disk, to maximize the length. The track is a mixture of 90% tungsten and 10% rhenium ( $Z = 75$ ), which reduces the crazing effect caused by thermal stresses. Modern rotating anodes are made from

solid molybdenum onto which a thin tungsten-rhenium focal track is coated. The specific heat capacity of molybdenum is higher than that of tungsten ( $250$  vs  $130 \text{ J Kg}^{-1}\text{K}^{-1}$ ). The mass of molybdenum anode is also lower due its lower density. Some high output X-ray tubes have radial slots cut into the anode disc to reduce thermal stresses caused by repeated heating and cooling. Heavy duty X-ray tubes have graphite layer (Carbon) in the back of the anode disk. The anode disk has a beveled edge and the angle of the bevel may vary from  $6$  to  $20^\circ$ . The bevel is used to achieve the line focus principle.

### **Anode Stem**

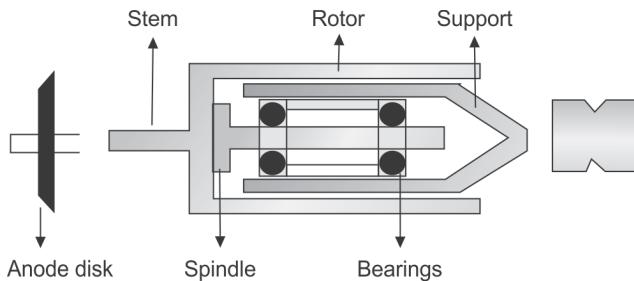
The anode disk is mounted on a stem, which is attached to the rotor. The anode assembly rotates with the help of bearings. The stem is made of molybdenum, which is having high melting point ( $2620^\circ\text{C}$ ) and poor heat conduction. The molybdenum stem prevents the flow of heat from the tungsten to the bearings of anode assembly, due its small cross section. Thus, the bearings are protected from heat, which may cause them to expand and bind. Higher the length of the stem, higher the inertia of the tungsten disk, more will be the load on the bearings. Hence, it is desirable to keep the stem as short as possible.

### **Rotor**

The anode disk is connected to a rotor, which is made up of copper bars arranged around a cylindrical iron core. There are electromagnets surrounding the rotor, outside the glass envelope is called stator (Fig. 3.15). Both the stator and rotor is called as an induction motor. When the stator coils are energized, a rotating magnetic field is produced, that induces current in the copper bars of the rotor. This induced current produces an opposing magnetic field that causes the rotor to spin.

The rotor rotates at a speed of about  $3000$ – $9000$  revolutions per minute (rpm). This will facilitate the electrons to bombard a constantly changing area of the target. Since the electrical conductivity of copper is higher, it will facilitate induction of strong currents from the induction coil supply. The surface of the rotor is blackened to enhance heat dissipation by radiation process. The rotor support is made of steel and the positive high tension supply is made at the end of the rotor support outside the glass envelope.

Low speed rotor operates from  $60$  Hz power (single phase) and gives rotation of about  $3000$  rpm. This speed is too slow for short exposure of the order of mill-seconds. High speed rotor operates from



**FIG. 3.15:** Rotating anode X-ray tube and anode assembly inner view

180 Hz power (3 phase) and gives a rotation speed of 9000 rpm. If the speed of rotation is increased, the heat generated at the focus will be spread over a larger area. Modern X-ray tubes employ higher rotor speeds by increasing the frequency of the stator supply with frequency multiplying circuits. The X-ray machines are designed so that the tube cannot be energized until the anode attains its full speed. This delay time (1–2 s) is incorporated in the exposure buttons. The power supplied to the induction coils, produce eddy current which causes heating of the rotor.

### Bearings

The anode assembly is rotated with the help of bearings, which are made of steel ball races. Bearings are in the high vacuum environment and require special heat insensitive, nonvolatile lubricants. The bearings are coated with lead or silver to act as (metallic lubricants) lubricant. Commonly available lubricants such as oil and grease cannot be used, since it will vaporize while heating and destroy the vacuum. Even dry lubricant (graphite) would wear off as a powder and destroy the vacuum.

When the X-ray unit is turned on, the motor alone is energized first for a few seconds until the rotor reaches its operating speed. Then, a high voltage is applied to the tube for the required exposure. After the exposure, the rotor is slowed down quickly by dynamic braking to avoid the wear in the bearings. Since the electrons are striking the whole circumference of the anode, no part of the anode attains very high temperature. The heat from the tungsten disk is dissipated by radiating through the vacuum to the wall of the tube, and then into the surrounding oil and tube housing.

The life of a rotating anode X-ray tube is limited, because of pitting due to continuous electron bombardment on the anode surface.

These changes are due to thermal stress. This pitting reduces the X-ray yield and changes the spectral distribution. The decreased output, results in excessive scattering of X-rays and increased absorption of X-rays by the target itself. A pitted anode will affect the electric field between cathode and anode, so alter the size of the focal spot.

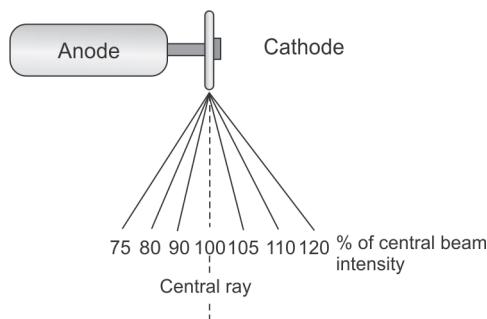
### Grid Controlled X-ray Tubes

A grid controlled X-ray tube contains three electrodes, the anode, cathode and the focusing cup. The focusing cup acts as a third electrode (grid), and controls the flow of electrons from the filament to the target. The grid is electrically negative relative to the filament. The voltage across the filament-grid produces an electric field along the path of the electron beam, that pushes the electrons even closer together. If the voltage is made large enough, the tube current may be completely pinched off, and no electrons go from the filament to target. The voltage applied between the focusing cup and filament acts like a switch to turn the tube current on and off. Since the cup and filament are close together, the voltage necessary to cut off the tube current is not very large.

For example, a 0.3 mm focal spot tube operating at 105 kV<sub>p</sub> require about -1500 V between the filament and cup. This type of grid controlled X-ray tubes are useful in some procedures involving rapid switching and short exposure times, e.g. cini-angio-cardiography and pulsed fluoroscopy.

### HEEL EFFECT

The heel effect refers to the reduced intensity of the X-ray beam towards the anode side of the X-ray field (Fig. 3.16). The X-ray photons that are emitted on the anode side of the field must pass through a greater



**FIG. 3.16:** Heel effect

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thickness of the anode than those directed toward the cathode side. This results in a reduced intensity on the anode side of the field. The magnitude of the heel effect depends on the anode angle, focus to film distance (FFD) and field size. The heel effect is less important at large FFD, because the film subtends a smaller beam angle. To reduce heel effect, anode angle should be increased and field size should be decreased. For better balance of the transmitted X-rays, the cathode side of the tube is oriented over the thicker parts and anode over the thinner parts of the patient.

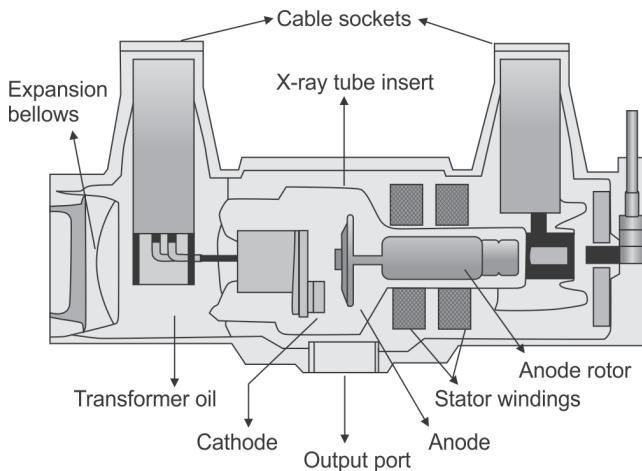
### **OFF-FOCUS RADIATION**

Off-focus radiation is produced by an X-ray tube when high speed electron interacts the anode surfaces, other than the focal spot area. The main source of off-focus radiation is scattered electrons at the target. They are accelerated back to the anode, outside the focal spot. They create a low intensity X-rays over the face of the anode. Off-focus radiation increases the patient exposure, geometric blurring and background fog, resulting poor image quality. To reduce the off-focus radiation, small lead collimator may be placed very close to the X-ray tube port. Grounded anode X-ray tubes (anode and metal tube envelopes are given same electrical potential) reduce off focus radiation since the scattered electrons are attracted by the metal envelope. Tubes that are used in mammography also reduce off-focus radiation.

### **X-RAY TUBE AND HOUSING**

The tube housing supports, insulates and protects the tube insert from the environment (Fig. 3.17). The tube housing is internally shielded with lead to attenuate X-rays emitted in other directions except through the window. The shield should perform four functions namely, (i) radiation protection, (ii) electrical protection, (iii) thermal protection and (iv) physical protection. Steel casing is lined with lead to prevent radiation emerging in all directions. The Perspex/beryllium window is convex upwards to reduce filtration of the X-ray beam by oil. To prevent electrical shock, the shield is earthed. Wherever high tension cables enter the shield, insulated sockets are used. The shield is filled with mineral oil, which act as a electrical insulator and prevents sparking across the insert.

The oil also acts as a cooling medium and expands at higher temperatures. The oil expansion activates bellow to operate a micro



**FIG. 3.17:** X-ray tube and housing anatomy

switch, so that further use of the tube is prevented. The oil expansion also helps prevent the entry of air into the tube insert. The shield also protects the insert from accidental damage caused by knocks and bumps.

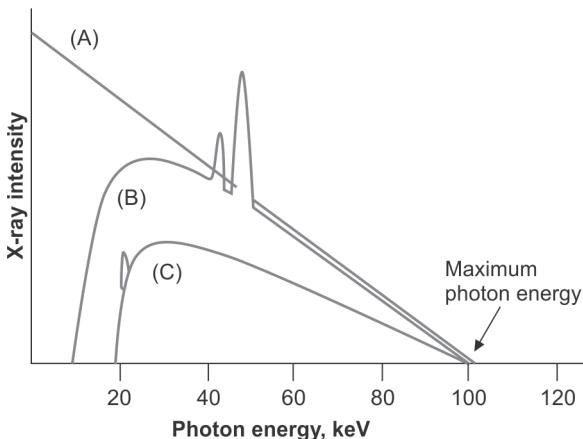
The effectiveness of the tube housing in limiting the leakage radiation must meet the specifications, listed by the Atomic Energy Regulatory Board (AERB). The leakage radiation measured at a distance of 1 m from the source shall not exceed 115 mR (air kerma of 1 mGy) in one hour, when the tube is operating at each of the ratings specified by the manufacturer.

## FILTERS

A filter is a metallic sheet introduced in the path of X-rays, in order to reduce the patient dose. Diagnostic X-rays consist of both low energy and high energy X-rays. When X-rays pass through a patient, only high energy X-rays penetrate through the patient and form the radiological image. Whereas, the low energy X-rays are absorbed in the first few centimeters of tissue, thereby increasing the radiation dose. The introduction of filters absorb these low energy X-rays and reduce the patient dose. This process of removing the low energy X-rays, by introducing metallic sheets is called filtration (Fig. 3.18).

Filtration has two components namely, (i) inherent filtration and (ii) added filtration. Filtration resulting from the absorption of X-rays by

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**FIG. 3.18:** Effect of filter, (A) Unfiltered spectrum, (B) Filtered spectrum with inherent filter, and (C) Filtered spectrum with added filter

the X-ray tube and its housing is called inherent filtration. This usually varies between 0.5 mm and 1.0 mm of Al equivalent. Added filtration results from absorbers placed in the path of the X-ray beam. The sum of the inherent and added filtration gives the total filtration.

$$\text{Total filtration} = \text{Inherent filtration} + \text{Added filtration}.$$

Al and Cu are the materials usually used in diagnostic radiology. The thickness of added filter varies from 1.0 mm to 1.5 mm of Al. Aluminum ( $Z = 13$ ) is an excellent filter material for low energy X-rays. Copper ( $Z = 29$ ) is a better filter for high energy radiation. Copper is always used in combination with aluminum as a compound filter. A compound filter consists of two or more layers of different metals. The layers are arranged in such a way that the high  $Z$  layer always faces the X-ray tube.

The recommended total filtration for diagnostic X-ray unit of  $> 100 \text{ kV}_p$  is 2.5 mm Al. X-ray units with proper filters, reduces the patient dose significantly, up to 80%. Filters are simple and inexpensive. Though filters reduce the intensity of X-ray beam significantly, it does not affect the maximum energy of the X-ray beam spectrum.

Heavy metal filters (Gd, Ho) are also used in general radiography. These filters make use of the K-edge, and offer increased absorption of X-rays, while imaging with contrast agents. They enhance contrast for iodine and barium, reduce patient dose and increase tube loading.

The recommended beam filtration is follows:

- i. General radiography
  - 1.5 mm Al below 70 kVp
  - 2.0 mm Al between 70 and 100 kVp
  - 2.5 mm Al above 100 kVp
- ii. Mammography
  - Be 1 mm + Mo 0.03 mm (Mo target)
  - Be 1 mm + Rh 0.025 mm (Rh target)

In mammography, Mo target with Rh filter is commonly used. However, Mo cannot be used as filter in mammography X-ray tubes with Rh targets.

## **SCATTERED RADIATIONS**

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There are three types of radiation involved in patient imaging, namely, primary, scattered and leakage radiation. Leakage radiation does not contribute to image formation and no discussion is required. However, primary and scattered radiations are responsible not only for image formation but also the degree of image quality. Two vital factors of image quality are spatial resolution and contrast resolution. Spatial resolution is greatly controlled by focal spot, whereas contrast resolution is controlled by scatter radiation or noise. Scatter radiation is produced by Compton interaction, resulting in noise. Hence, scatter radiation needs to be reduced to obtain good quality image. That is why collimators and grids are used in patient imaging.

Scattered radiation mainly depends on kVp, field size and patient thickness. As the kVp increases, the X-ray energy increases. As a result, Compton interaction increases, and photoelectric interaction decreases. Hence, increase of kVp, increases the scatter radiation and reduces image quality. Therefore, X-ray imaging should be done with minimum kVp, with lowest scatter. But, at low kVp, the percentage of transmission may be lesser, which can be compensated with increase of mAs. Increase of mAs may account higher patient dose, hence optimal selection of kVp and mAs is required.

Scattered radiation increases with field size. As the field size increases, scatter radiation also increases, which reduces the contrast of the image. Smaller the field size, lesser the scatter radiation and lesser the optical density. To maintain the optical density, higher exposure techniques are required with smaller field size.

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Scatter radiation increases with patient thickness. More scatter radiation is involved with thicker patients or thicker body parts. Mainly muscle, fat, bone and fluid filled cavity (pathology) are the sources of scatter radiation. Abdomen X-ray produces 3 times higher scatter than that of extremity X-rays. Compression will reduce patient thickness and bring the patient closer to the film. It will improve spatial and contrast resolution, with reduced patient dose. Patient thickness cannot be controlled except in mammography, only proper selection of techniques will help to obtain a good quality image.

### BEAM RESTRICTORS OR COLLIMATORS

An X-ray beam restrictor is a device that is attached to the X-ray tube housing, to regulate the size and shape of an X-ray beam. They can be classified into three categories, namely, (i) aperture diaphragms (ii) cones and cylinders (Fig. 3.19) and (iii) collimators.

Aperture diaphragms consist of a sheet of lead with a hole in the center. The size of the hole determine the size and shape of the X-ray beam. It is simple and the aperture can be altered to any size and shape. The disadvantage of an aperture diaphragm is that it produces large penumbra. The penumbra can be reduced by keeping aperture diaphragm far away from the X-ray target. Aperture diaphragms are used in dental radiography with rectangular collimation. In addition, it is used in trauma and chest radiography.

The use of cones and cylinders will reduce the penumbra considerably. Both have extended metal structures that restrict the useful circular beam to the required size. The position and size of the distal end determine the field size. If the X-ray source, cone and film are not aligned properly, then, one side of the film may not be exposed, which is called cone cutting. Cone is a ideal beam restrictor, but the flare of the cone is greater than the flare of the X-ray beam. These systems provide only limited number of field sizes.

The collimator is the best X-ray beam restrictor. It defines the size and shape of the X-ray field that emerges from the X-ray tube.

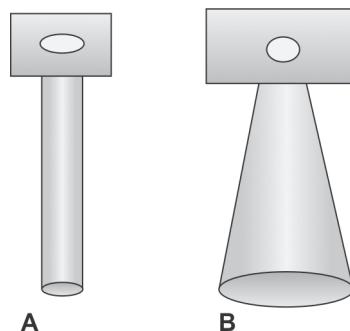
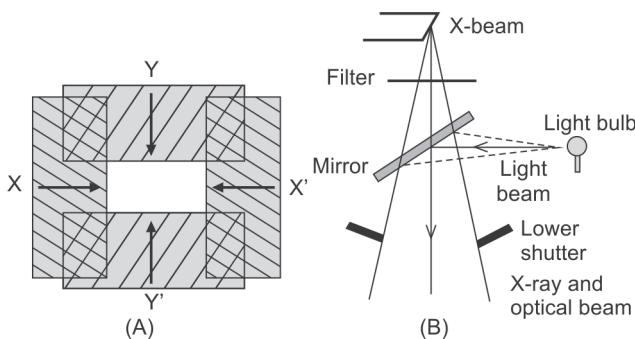


FIG. 3.19: (A) Cylinder and (B) Cone

The collimator assembly is attached to the tube housing at the tube port. A collimator consists of two sets of shutters, which can be moved independently. Each shutter consists of four or more lead plates of 3 mm thick, which can absorb X-rays completely, to provide a well defined X-ray field. When the shutters are closed, they meet at the center of the X-ray field.

The collimator also has a light and mirror arrangement, to illuminate the X-ray field. The light bulb is positioned laterally and the mirror is mounted in the path of the X-ray beam at an angle 45° (Fig. 3.20). The target and the light bulb should be kept at equal distance from the center of the mirror. The collimator provides variety of rectangular X-ray fields and the light beam shows the center of the X-ray field. The light field and radiation field should match exactly with each other. The variation must be within 4% of TFD. The alignment of light beam and X-ray beam should be checked periodically. A well collimated beam covers lesser area of the patient, giving less patient dose. Also it generates less scatter radiation, which improves the image quality.



**FIG. 3.20:** (A) Collimator shutters and (B) Light and mirror arrangement to create radiation and optical field coincidence

Collimators that automatically limit the X-ray field size to the useful area of the detector is also available. These are called positive beam limitation (PBL) collimators. A sensor in the cassette holder, adjust the collimator opening, equal to the cassette dimensions. Thus, PBL collimators limit the irradiated volume and reduce the patient dose.

# 4

# Generation and Control of X-rays

The X-ray generators use transformers and rectifiers to generate suitable DC voltage to the X-ray tube. The generator also has operator console, kV<sub>p</sub> and mA controls, exposure time selection, the kV and mA meters, primary and secondary switching, filament transformer, automatic exposure control circuits, space charge and voltage compensation circuit and exposure timer. The types of X-ray generators are single-phase, three-phase, constant potential, and high frequency inverter generators.

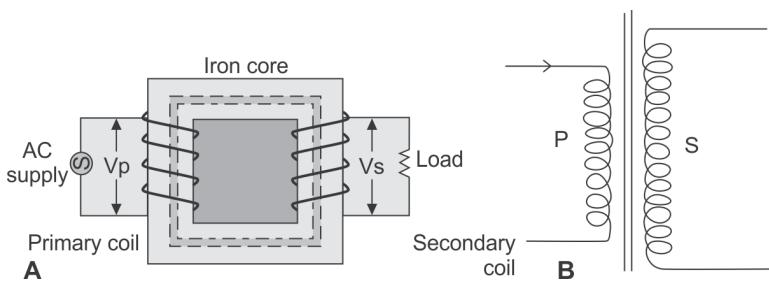
## **TRANSFORMER**

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The transformer is an electrical device, which can convert electrical energy from one coil to another coil. The transformer is working on the principle mutual induction. The transformer basically consists of two coils, namely, primary and secondary (Fig. 4.1). These coils are wound on a iron core. The alternating voltage, which is to be transferred, is applied in the primary coil as input. This produces a changing magnetic flux in the iron core, which produces an alternating emf in the secondary coil. The induced emf's in the coils are directly proportional to the respective number of turns of the coil.

Let N<sub>p</sub> and N<sub>s</sub> are the number of turns in the primary and secondary coils respectively. Let V<sub>p</sub> and V<sub>s</sub> are the voltage in the primary and secondary coils respectively. Let I<sub>p</sub> and I<sub>s</sub> are the current in the primary and secondary coils respectively, then,

$$\begin{aligned} V_p &\propto N_p \\ V_s &\propto N_s \\ \text{or } \frac{V_p}{V_s} &= \frac{N_p}{N_s} \end{aligned}$$



**FIG. 4.1:** (A) Transformer principle, and (B) Symbol

In a transformer, the turns ratio is equal to the voltage ratio. The power input in the primary ( $P_p$ ) =  $V_p \times I_p$  and the power output in the secondary ( $P_s$ ) =  $V_s \times I_s$ . On the basis of the law of conservation of energy, power input is equal to the power output.

$$V_p \times I_p = V_s \times I_s$$

$$\frac{V_p}{V_s} = \frac{I_s}{I_p} = \frac{N_p}{N_s}$$

This shows that the current in the coils are inversely proportional to the number of turns in the respective coils.

Basically, there are three types of transformers, namely, step up, stepdown and isolation transformers. If a transformer, transfers power of low voltage and high current into power of high voltage and low current, it is called step-up transformer. In this type, the secondary coil will have more number of turns than the primary, i.e.  $N_s > N_p$ . If a transformer, transfers power of high voltage and low current into power of low voltage and high current, it is called step-down transformer. In this type, the primary will have large number of turns than the secondary, i.e.  $N_s < N_p$ . If  $N_s = N_p$ , this results in an isolation transformer.

### Efficiency

The efficiency of the transformer is the ratio between the output power and input power.

$$\begin{aligned} \text{i.e. efficiency} &= \frac{\text{output power}}{\text{input power}} \\ &= \frac{P_s}{P_p} \times 100 \end{aligned}$$

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In actual transformer, the output power is always lesser than the input power due to some energy losses. Hence, the efficiency is always less than <100%.

### Transformer Rating

The transformer rating refers the maximum safe output that can be taken from the secondary winding. The ratings are specified in three ways as follows: the highest voltage which the transformer can provide, the maximum current which the transformer can give on continuous running and the maximum current which the transformer can give for a period not exceeding one second.

If the rating is exceeded, the transformer may overheat and burn out its insulation and winding. Usually, it is expressed as the maximum safe output of its secondary winding in kilowatts. For three-phase generators ratings are calculated by the formula,

$$kW = (kV \times mA) \div 1000$$

For example, a three-phase generator operating at 100 kV and 500 mA, will have a rating as follows:

$$\text{Rating} = \frac{(100 \times 500)}{1000} = 50 \text{ kW}$$

For single-phase generators, the formula is,  $kW = (kV \times mA \times 0.7) \div 1000$ . The factor 0.7 comes from the rms value of the voltage. Kilowatt ratings of X-ray generators are determined, when the generator is under load. Information about ratings are useful to compare X-ray generators.

### TRANSFORMER LOSSES

In practice, the output power is always lesser than the input power and hence, the efficiency of the transformer is always less than 100%. This implies that some amount of energy is lost in the form of heat. This energy loss can be considered as copper losses, eddy current losses, hysteresis and flux leakage losses.

#### Copper Losses

Whenever a current  $I$  flows through a resistance  $R$ , an amount of power equal to  $I^2 \times R \times t$  watt is converted into heat. This can arise in both copper coils and iron core. The copper coil has resistance. If current flows through this coil, electrical energy equal to  $I^2Rt$  is

converted into heat. To reduce this loss, the current cannot be reduced because the normal operation of the transformer will be affected. Instead, the resistance of the coil must be minimized by using wire of low resistivity. Therefore, thicker wire should be used as transformer coil. The optimum thickness will be decided by comparing the cost, space and saving of power. Copper is the best coil material available to day and hence, it is commonly used.

### **Eddy Current Losses**

The iron core consists of concentric layers of iron, each acts as a circuited single turn coil. Whenever the magnetic field changes, an emf will be induced in the core. The current produced by the induced emf in the core is called eddy current, which will give rise to  $I^2Rt$  heat losses. These eddy currents can be eliminated by making the iron core in the form of thin sheet of metal, and each sheet is insulated from its neighbor by a thin layer of paper. This type of core is known as laminated core. The core is usually made up of stelloy, an alloy of steel. Some design employ high resistance ceramics as core material.

### **Hysteresis Losses**

The transformer core is a magnetic material. The core is magnetized twice in each cycle of the alternating voltage. When the direction of AC changes, the magnetization is also gets reversed. During this reversal, some energy is lost due to the molecular friction and the energy appears as heat. The loss of energy by molecular friction is called hysteresis loss. This can be reduced in practice by choosing a suitable magnetic material, such as mu-metal, which has low hysteresis loss. Mu-metal is a ferromagnetic alloy containing 78% nickel, 17% iron, and 5% copper. It has high permeability.

### **Flux Leakage**

All the magnetic flux linked with the primary is not linked with the secondary coil. This is said to be flux leakage, which results in loss of energy. This can be minimized by using good core design like shell type of core.

## **TRANSFORMER CONSTRUCTION**

A practical transformer differs considerably from the ideal transformer. The following points should be considered during the construction of the transformer.

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### Winding

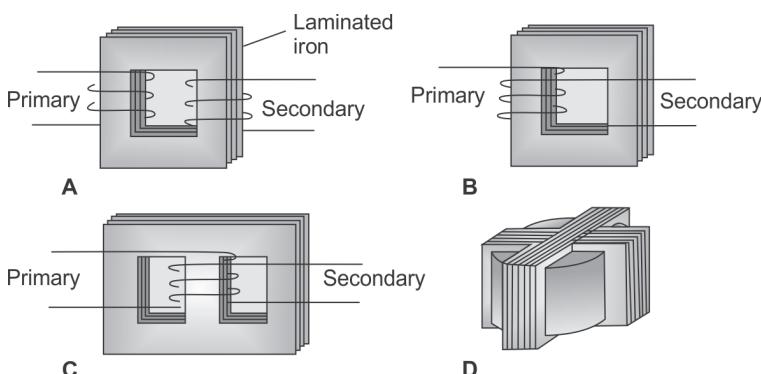
The transformer is usually made with single primary winding, whereas the secondary will have more than one winding. For example, the primary of a transformer used in control equipment, may be designed for 200 V input. The secondary may be designed with three windings for three output such as 500 V low current, 50 V low current and 6 V high current. The primary will have a wire of medium thickness, whereas the secondary have thin and thick wires. Thicker wire offer lesser resistance and allow flow of high current in the primary. Whereas secondary handle only low current, hence, thin wire with relatively higher resistance is preferred to save cost and power loss.

### Core

The transformer cores are always designed so that they form a closed circuit. A core with a closed magnetic circuit has high permeability and is very efficient. At the same time, the core is laminated to eliminate eddy current losses. There are three types of core, namely, (i) core type, (ii) shell type, and (iii) cross type or H type (Fig. 4.2).

In a core type transformer, the primary winding is on one leg and the secondary winding is on other leg. This is easily assembled and has a good cooling surface. Alternatively, both primary and secondary windings are made as two halves. The secondary is wound over the primary winding on each leg. This is the most preferred transformer core type, used in X-ray generators.

In a shell type transformer, the primary and secondary are wound around the central limb, and the magnetic circuit is shorter. Shell type



**FIG. 4.2:** Different types of transformer cores: (A) Core or single window type, (B) Core or single window type, (C) Shell or double window type and (D) Cross or H type

is the most efficient design in terms of energy conservation and efficiency (98%). Hence, it is used most commonly.

The cross or H type core is called modified shell type, since it is a combination of two shell cores set at right angles to each other. In this, the coils are surrounded by four legs. The windings are located over the center core, which is four times the area of the each of the outside legs. This type of core is cooled easily and hence used in large power transformers, where the voltage drop and cost is kept minimum.

Transformer designed for higher output voltages, such as 100 kV, needs special care. The secondary winding must be designed very carefully to avoid electrical break down due to ionization of the surrounding air. Transformers are cooled by oil or forced air, to avoid overheating. Transformers never be immersed in water for cooling. During accidental flooding, if the transformer is immersed in water, immediately the water should be pumped out.

## **Oil**

High voltage transformers are usually enclosed in a metal tank filled with oil. This oil penetrates into the inner spaces of the windings and increases the effectiveness of the insulation. The oil prevents the windings from dust and moisture and also acts as a cooling medium. The oil is a good insulator than air, it avoids electrical short circuiting. Oil also provides effective cooling to the transformer.

## **Autotransformer**

The autotransformer consists of a single winding wound on a laminated iron core and it is working on the principle of self induction (Fig. 4.3). The primary voltage is applied across two of the terminals, and the secondary voltage taken from two terminals, almost always having one terminal in common with the primary voltage. The primary and secondary circuits, therefore, have a number of windings turns in common.

The alternating current applied between the input points will induce a flow of magnetic flux around the core. This magnetic flux will link with all the turns forming the coil, inducing a voltage into each turn of the winding. Since the volts-per-turn is the same in both windings, each develops a voltage in proportion to its number of turns. In an autotransformer, part of the current flows directly from the input to the output, and the other part is transferred inductively, allowing a smaller, lighter, cheaper core to be used as well as requiring only a single winding.

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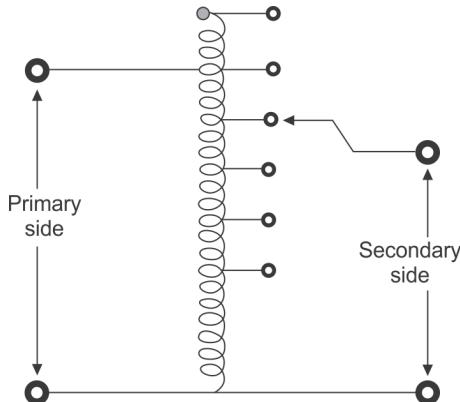


FIG. 4.3: Autotransformer

For example, if 230 V are applied between points A and B (Fig. 4.4), which involves 115 turns of the autotransformer winding, then the volts per turn ( $230 \div 115$ ) will be 2. By a suitable selection of taps, one may select the number of turns to supply the necessary voltage to the other components. A selection of 55 turns will provide voltage of 110 V, while a selection of 160 turns provide 320 V. Thus, an autotransformer can function as a step-up or step-down transformer.

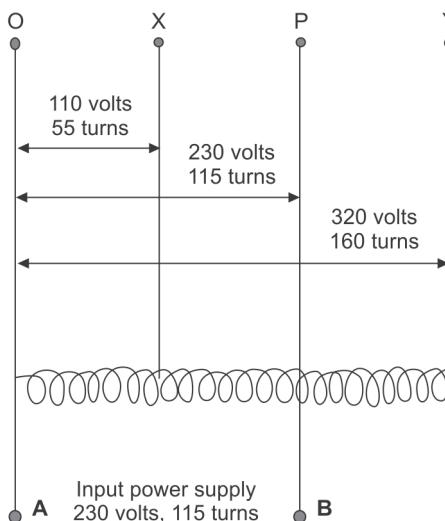


FIG. 4.4: Autotransformer working principle

These transformers are widely used where electrical isolation between primary and secondary is not necessary. Autotransformer

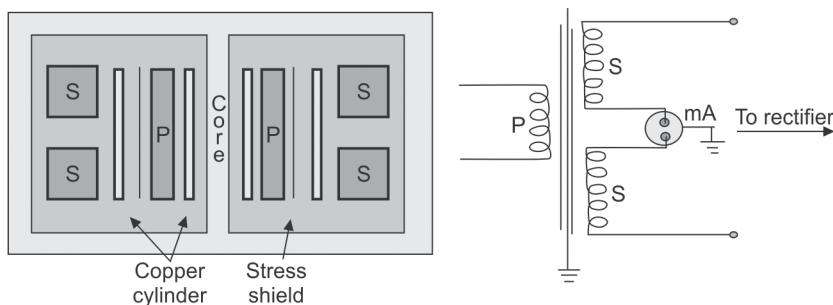
occupy very important place in X-ray generator circuits. In the X-ray generator, the autotransformer is used to adjust the voltage applied to the primary of the high voltage transformer with high efficiency and convenience.

An autotransformer does not provide electrical isolation between its windings as an ordinary transformer does. A failure of the insulation of the windings of an autotransformer can result in full input voltage applied to the output. If there is a break in the part of the winding, then the transformer acts as an inductor in series with the load.

### HIGH-TENSION TRANSFORMER

The high voltage transformer is used to transfer low voltage into high voltages required to operate X-ray tubes. This is known as high tension generator, which provide voltage from 20–150 kV and current up to 1000 mA for the X-ray tubes. It is a step up transformer with two windings and a shell type core. The number of turns in the secondary is higher than that of primary, and it is decided by the voltage ratio. If 400 volt is to be transformed into 80,000 volts, then the voltage ratio is 80,000/400, requiring 200 turns in secondary per primary turn. Thus primary winding consists of a few hundred turns of thick copper wire, which is well insulated and wound on a cylinder. A thin copper sheet is fitted over the primary winding and it is earthed. This is known as stress shield, which protects the primary circuit during breakdown of the secondary insulation (Fig. 4.5).

The secondary winding consists of about 100,000 or more turns of thin copper wire coated with insulated varnish. This is wound in an insulating cylinder, which is placed over the primary winding. The layers are separated from each other by thin paper prepared with wax for insulation. The voltage difference between any two layers



**FIG. 4.5:** High-tension transformer

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is about only 200–300 volts. This method of design reduces the risk due insulation breakdowns. The core of the high tension transformer is rectangular in shape, which is earthed and well laminated.

Usually, the secondary is wound in two equal multiple parts and the center of the winding is earthed through the core. This means that instead of running from 0 to + 150 kV, the two secondary cables run from – 75 kV to + 75 kV. This is to reduce the insulation, size and cost. Since the current measurement in the primary coil is not an accurate representation of the current in the secondary coil, the current must be measured on the secondary side.

A milliammeter (mA meter) is connected between the inner ends of the two secondary windings at which the transformer is grounded, which is also the center of the coil. This minimizes the risk of electrical shock to the operator. Though the mA meter is connected at this point, it is placed at remote distance at the control console.

The entire unit is immersed in an earthed metal tank, which is filled with oil. The metal tank is closed with a tight lid. In the case of dental and mobile X-rays, the heat production is very low and hence, oil is not used. Instead of oil, plastic is used as an insulator. The transformer is immersed in plastic when it is in fluid state. Later on, the plastic solidifies and acts like a solid insulator.

## **RECTIFIER CIRCUIT**

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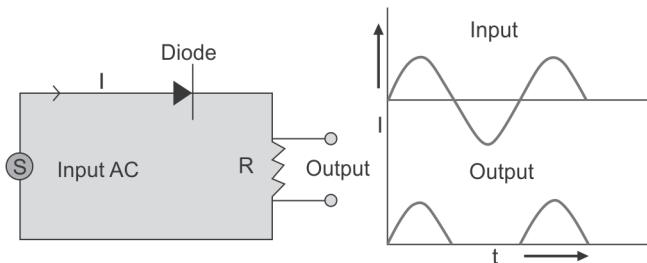
### **RECTIFICATION**

Rectification is the process of changing alternating current into direct current. The device that produces the change is called a rectifier. A rectifier allows an electrical current to flow in one direction but does not allow current to flow in the other direction. Rectifiers are connected into the X-ray circuit in series. They are mainly divided into half wave and full wave rectifiers.

If alternating voltage is applied directly to the X-ray tube, the anode will emit electrons, whenever it is negative with respect to cathode. These electrons will travel towards the cathode and bombard the filament and destroy the filament. This is called back projection which is avoided by the supply of rectified DC voltage. Thus, rectifiers play an important role in X-ray production.

## HALF WAVE RECTIFIER

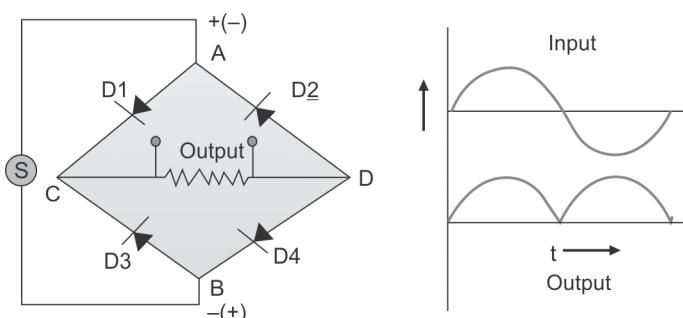
Vacuum tube diodes or solid state (semiconductor) diodes can be used for rectification. In a half wave rectifier, a single diode is used, as shown in the Figure 4.6. An alternating voltage is applied to the diode as input. The output is obtained across the resistance R. When the plate is positive, the diode will allow the current to flow. When the plate is negative, the diode will not allow the current. Therefore, the diode will allow the current only during those half cycles when the plate is positive. Hence, the output current is always in one direction. This circuit is known as half wave rectifier and it is mainly used in mobile and dental X-ray units. A single solid state diode cannot prevent reverse current at higher voltages. Hence, many diodes are placed in series in a stick to do rectification.



**FIG. 4.6:** Half wave rectifier

## FULL WAVE RECTIFIER

In the half wave rectifier, the input voltage is used only in one half of the cycle. The other half of the cycle is not used. Therefore, there is a need for a rectifier, which will use the full cycle of the input. This is possible by having two or more number of diodes, as shown in the Figure 4.7. The alternating voltage is applied between A and B. The output is obtained across the resistance R.



**FIG. 4.7:** Full wave rectifier

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When end A is positive, D<sub>1</sub> and D<sub>4</sub> will conduct and a current flows through R. During the next half of the cycle end A is negative, and end B is positive. Now, the diodes D<sub>2</sub> and D<sub>3</sub> will conduct and a current flows through R. Thus, the current flows through the resistance R during full cycle of the input voltage, in the same direction. X-rays are produced in two pulses per cycle, irrespective of the polarity of the transformer. Three phase generator employ multiple rectifiers in the secondary circuit. Full wave rectifiers are used in high end X-ray tubes which employ rotating anode X-ray tubes.

### THYRISTOR

A thyristor is a silicon-controlled rectifier which has four layer semiconductors (n-p-n-p). It is used to switch larger currents, which the transistor cannot handle. It has two large terminals, namely, anode and cathode, which connects the main circuit (Fig. 4.8). The third terminal is the gate, which is smaller in size. Initially, the junction J<sub>1</sub> and J<sub>3</sub> are forward biased and junction J<sub>2</sub> is reverse biased. Hence, only small current flows through the circuit and the thyristor is said to be in OFF state.

If a positive voltage is applied in the gate terminal, holes flows through J<sub>3</sub> and the barrier across J<sub>2</sub> breaks down. This will facilitates movement of electrons across J<sub>2</sub> junction and makes the thyristor to ON condition. The conduction continues in the circuit, even after the gate voltage is removed. The conduction ceases only when the potential difference across the anode and cathode falls to zero. Thyristor conduct current only in one direction and can be used to switch the alternating current.

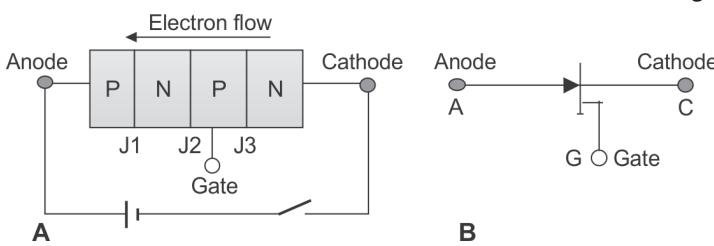


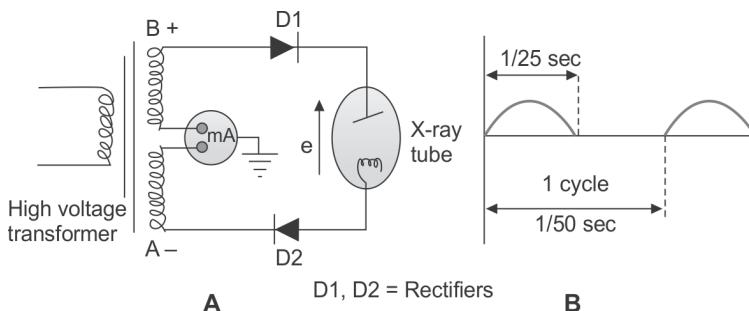
FIG. 4.8: (A)Thyristor rectifier principle and (B) Symbol

### HALF WAVE RECTIFICATION X-RAY CIRCUIT

The half wave rectification is the most commonly used circuit. The circuit uses two rectifier stacks connected in series with the X-ray tube, as shown in Figure 4.9. The center point of the secondary winding of the step up transformer is grounded. The electrons flow through

the X-ray tube from the cathode to anode in the first half cycle. When the voltage reverses, in the second half cycle, the rectifier stops current flow. Since there are two rectifiers, the circuit is symmetric and each has to withstand only half of the peak voltage ( $V_p$ ). The anode goes only to  $+V_p/2$  at the peak of the conducting cycle and the cathode to  $-V_p/2$ . The tube voltage, tube current and X-ray pulse are shown as a function of time.

The discontinuous nature of X-ray yield reveals that the tube is inoperative at least half the time. This means that the exposures must be twice as long to get the same X-ray flux. This increases the chance of organ motion during the exposure, with a loss of diagnostic information. The advantage of the half wave rectification is that they protect the X-ray tube from the full potential of the inverse cycle.



**FIG. 4.9:** (A)The half wave rectifier X-ray circuit and (B) Rectifier output

## FILAMENT CIRCUIT

The tube current can be altered by altering the number of electrons emitted by the filament. The number of electrons can be altered by changing the temperature of the filament. To achieve this, a filament circuit is used, which will regulate current flow through the filament (Fig. 4.10).

The power to heat the filament is provided by a small step-down transformer, called the filament transformer. In addition, the circuit consists of a variable resistor network and a focal spot size selector. This transformer has 10–20 times more turns in primary coil, compared to secondary coil. The filament is connected directly to the secondary coil of a step-down transformer. The primary coil of the transformer obtains its voltage from the auto transformer. Usually, the primary voltage will be around 100–200 V, whereas the secondary voltage

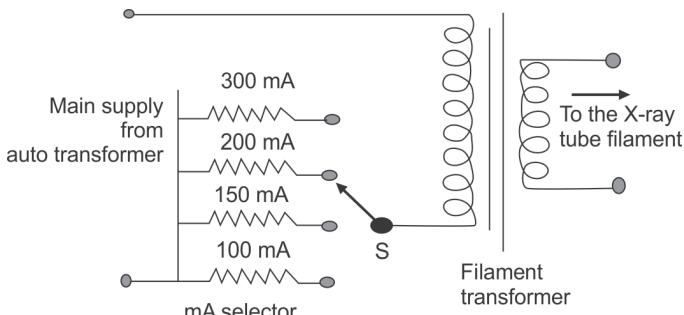


FIG. 4.10: The filament circuit

is around 10 V and current up to 7A. This makes it necessary to provide high voltage insulation between primary and secondary coils. Hence, the filament transformer is placed in the same oil-filled grounded metal tank as the high voltage transformer.

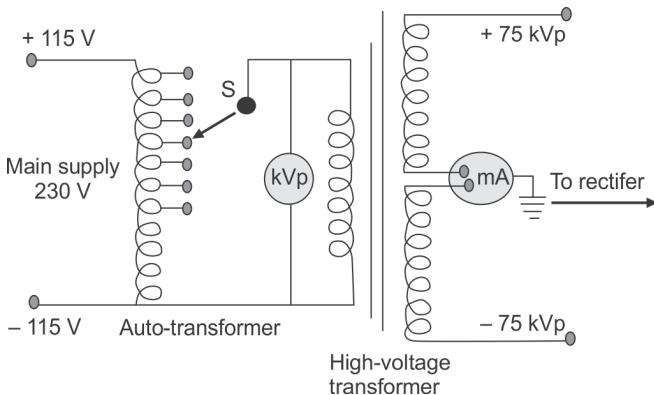
Precise control of filament heating is very essential. A small variation in the filament current results in a large variation in X-ray tube current. A 5% change in filament voltage may bring a change of 20–30% in X-ray tube current. The filament current may be controlled, by altering the voltage to the primary of the step-down transformer, by addition of resistors connected in series. The resistors may be a number of separate resistors or a single variable resister. As the resistance increases, the voltage to the filament decreases. For example, a current of 4 A and a resistance of 1.5 ohms will reduce voltage by 6 V.

When the selector S moves over the resistors, the primary voltage of the transformer is altered. As a result, different values of tube current (mA) are obtained. The selector is either a rotary switch or a push button located on the control panel. The circuit also has other components to stabilize the voltage to the filament transformer that includes a voltage stabilizer and a frequency stabilizer. There is also a circuit that automatically compensates for the space charge effect.

## KILOVOLTAGE (KV) CONTROL CIRCUIT

The kilovoltage applied across an X-ray tube determines maximum energy and hence, the penetrating power of the X-rays. To have a wide range of penetrating power of X-rays, the applied kilo voltage must be varied in small steps. By using a kilovoltage circuit, the kilovoltage can be varied in steps of say  $2 \text{ kV}_p$ . A simplified kilovoltage circuit is shown in Figure 4.11. The circuit has two transformers,

namely, an autotransformer and a step-up transformer. The autotransformer is actually the  $kV_p$  selector and is located in the control panel. The voltage across the primary coil of the step-up transformer can be varied by selecting the suitable number of turns in the auto transformer.



**FIG. 4.11:** Kilovoltage control circuit

The secondary coil of the step-up transformer has more turns than primary, and increases the voltage by a factor of 600. The potential difference across the secondary coil may be as high as 150,000 V, so the step-up transformer is immersed in oil for maximum insulation. There are two meters in the circuit, one to measure  $kV_p$  (voltmeter) and the other to measure mA (ammeter). The meters are located on the control panel, but their connections are in the high voltage circuit. They indicate the potential across the X-ray tube and the actual current flowing through the tube during exposure.

The potential difference across the X-ray tube can be measured indirectly on the low voltage side of the transformer. Therefore, the  $kV_p$  meter is placed in the circuit between the autotransformer and step-up transformer. Because the  $kV_p$  meter records the selected  $kV_p$  before the actual exposure begins, it is usually termed as pre-reading  $kV_p$ . If the  $kV_p$  meter is properly calibrated, it can directly read the applied voltage across the X-ray tube. Since the voltage in the primary circuit is relatively small, the meter can be placed on the control panel. This requires minimum insulation without any risk of electrical shock.

The connections for the mA meter must be in the secondary winding of the transformer. Since the efficiency of the transformer is less than 1, measurement at primary level is not the true representation of the

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current in the secondary. Hence, the mA meter is connected at the centre of the secondary coil, at which the transformer is grounded. This will minimize the risk of electric shock to the operator, since the center of the coil is at zero potential. Though the meter is connected at this point, it may be placed at the control panel.

The main supply is applied to the autotransformer. There are number of tapping in the autotransformer. By moving the stud selector over the tapping, the output voltage of the autotransformer can be varied. This variable output voltage is applied to the primary of the high tension transformer. Finally, the kilovoltage across the X-ray tube is varied.

If the range provided by the stud selector is from 40 to 100  $kV_p$  in steps of 2  $kV_p$ , then there must be 31 tapping. Usually, there are two selectors, one is a coarse control giving step of 10  $kV_p$  and the other is a fine control giving steps of 2  $kV_p$ . The kilovoltage can also be continuously varied by using a variance transformer. This type of control is employed in the diagnostic X-ray units, used for fluoroscopy.

### SINGLE-PHASE X-RAY GENERATOR

A single-phase X-ray generator utilizes a single-phase AC supply as input. These generators employ full wave rectification, which utilizes the full potential of the electrical supply. Figure 4.12 shows the full wave rectified single-phase generator and its wave form. Both half cycles of the AC is used to produce X-rays. Hence, the X-ray output per unit time is twice as large as that of half wave rectification.

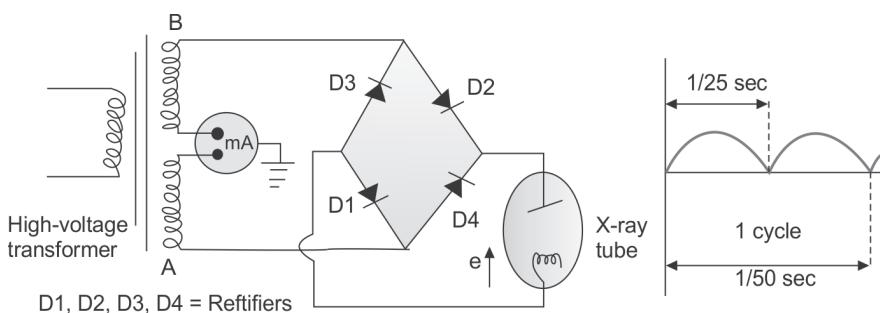
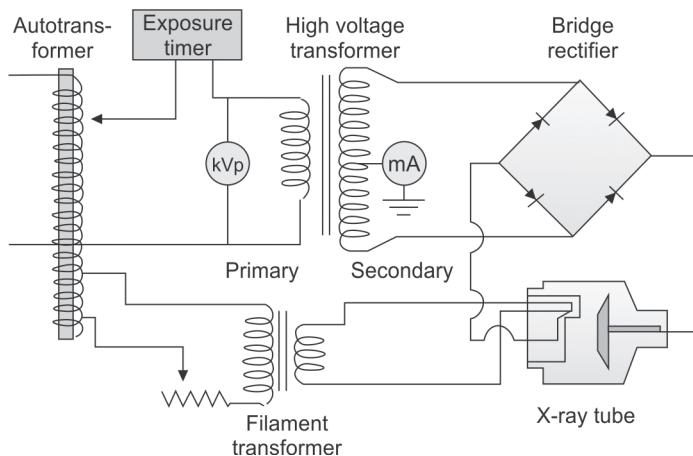


FIG. 4.12: Single phase X-ray generator with full wave rectifier

The voltage across the circuit is supplied by the step up transformer. In the first half cycle (A is negative and B is positive), the electrons

will flow from A through the rectifier D1 to the X-ray tube, and return through rectifier D2 to the side B. In the next half cycle (A is positive and B is negative), the electrons will flow from B through the rectifier D3 to the filament and return through the rectifier D4 to side A. Thus, the four rectifiers produce pulsating DC through the X-ray tube and the voltage across the tube fluctuates from zero to maximum.

The generated X-rays have 100 short pulses in one second (2 pulses/cycle, frequency = 50 cycles/sec). The exposure time for each X-ray pulse is  $1/100 \text{ s} = 10 \text{ ms}$ . The AC waveform can be easily switched off, when the voltage in the circuit is at zero level. It is at this point, the primary voltage switches can be opened easily. Hence, the timer is calibrated in fractions of seconds in most of the single-phase generators. Most of the X-ray pulses are generated during the peak value of the applied voltage. The tube current follows the kV in a nonlinear way below  $40 \text{ kV}_p$  due to space charge effect. A typical single-phase X-ray generator circuit design is shown in the Figure 4.13.



**FIG. 4.13:** Single-phase X-ray generator circuit

Thus, half wave and full wave rectifier circuits are generating only pulsating potential. The principal disadvantage of pulsed radiation is that a considerable portion of the exposure time is lost while the voltage is in the valley between two pulses. This will enable the low energy electron to bombard with the target, by giving heat and low energy X-rays. These X-rays are absorbed in the patient and raise patient dose.

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Hence, there is a need for constant potential circuits, which can give better X-ray output with more penetration. To achieve this, a condenser C is connected parallel to the X-ray tube. As a result, sufficient charge may be stored on it to maintain a constant voltage to the X-ray tube. Alternatively, three-phase generator can be used to produce constant potential across the X-ray tube.

### THREE-PHASE X-RAY GENERATOR

The three-phase X-ray generator uses a 3-phase AC line supply. There are three wires, each with a single phase AC sinusoidal wave. Each wave is out of phase with the other two for one-third ( $120^\circ$ ) of a cycle. A three-phase transformer is used to convert the low voltage AC to high voltage AC. It has three sets of primary and secondary windings. These windings are connected in one of the two configurations, namely, delta and wye (star). Generally, the primary windings are of delta configuration and the secondary is connected with wye configuration.

When the voltage is rectified, the circuit produces two pulses per cycle for each line, resulting six pulses per cycle. Hence, this is named as 3-phase 6-pulse generator. It is also possible, to produce 12-pulse per cycle, by using different configurations of transformers and rectifiers. This is called the 3-phase 12-pulse generator.

### SIX-PULSE THREE-PHASE GENERATOR

This type employs a delta-wound primary transformer with a wye-wound secondary transformer. The output of the secondary winding is rectified with six solid state rectifiers. The wye winding and 6 rectifiers are connected together, as shown in Figure 4.14. The rectified output will have six positive maximum voltages per cycle. Suppose, A is negative with respect to B, electron will flow from A, through

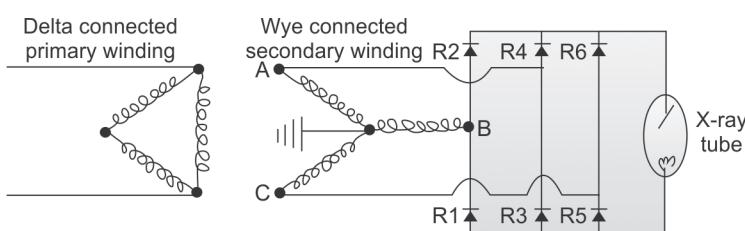
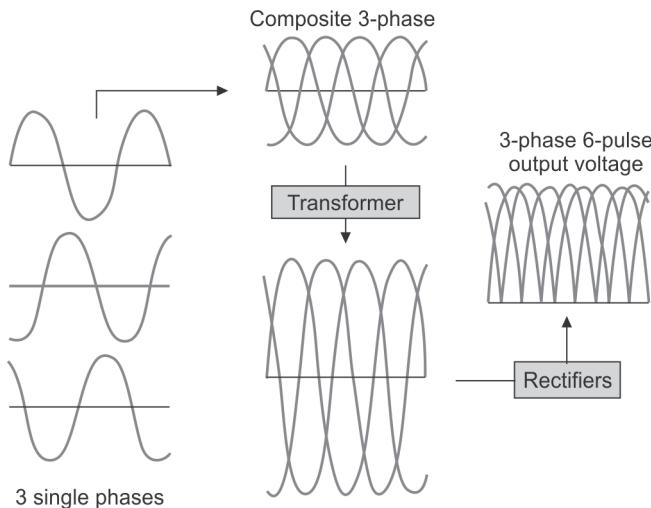


FIG. 4.14: A six-pulse, three-phase X-ray generator circuit



**FIG. 4.15:** The wave form of 3 phase, 6 pulse X-ray generator

rectifier R3 to the filament of the X-ray tube, then to the target of the tube and through the rectifier R2 and to the coil B. During the next half cycle, B would be negative with respect to A and the electron flow from B, through R1, X-ray tube, through R4 to A. By this method, full wave rectification of all three phases will produce six pulses per cycle.

Since voltage supplied to the X-ray tube never falls to zero, the ripple factor is very low (13.5%), as shown in Figure 4.15. The ripple factor of an DC voltage is the ratio of the difference between the maximum and minimum voltage divided by the maximum voltage.

$$\text{Ripple factor (\%)} = \frac{(V_{\max} - V_{\min})}{V_{\max}} \times 100$$

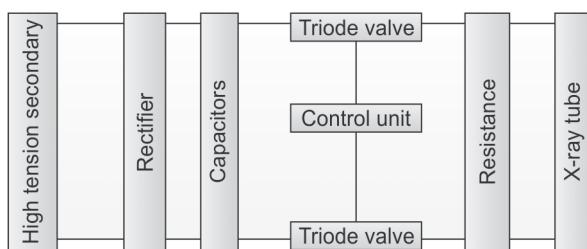
The ripple factor is the variation of voltage across the X-ray tube expressed as a percentage of the maximum value. For example, a ripple factor of 13.5% means that at 100 kV voltage fluctuates between 86.5 and 100 kV. The single phase X-ray generator ripple factor is 100, but in practice it is less than 100, due to capacitance effect of the cable. It means that the cable offers capacitance, which smoothes the DC voltage. The ripple factor for high frequency and constant potential generators are 4–15% and < 2%, respectively.

Three-phase generators produce a nearly constant potential. This is a major advantage over single-phase generators that produce a

pulsating potential. Three-phase generators produce X-rays efficiently throughout the exposure and the average X-ray energy is higher, because no time is left for low energy electron to bombard the target. A second major advantage of three-phase generators is a much higher tube rating for extremely short X-ray exposures. Since timer circuits use triode, tetrode or pentode in the secondary circuits, the beam on and off is possible with short exposure times. Nowadays three-phase generators are available to withstand a tube current of up to 2000 mA. This will make the X-ray tube to give short exposure times and high repetition rates, which is required for angiography. However, 3-phase generators are more expensive and difficult to install.

### **CONSTANT POTENTIAL GENERATOR**

This generator provides a constant voltage across the X-ray tube. It consists of a 3-phase AC voltage and a rectifier circuit. Vacuum tubes, such as triode or tetrodes are connected in line on the cathode side and on the anode side (Fig. 4.16). They control the kV and exposure time, on the high voltage side of the transformer. A comparator circuit measures the difference between the set kV (console) and the actual kV in the circuit, and adjusts the grid of the triode or tetrode tubes. These vacuum tubes provide extremely fast kV and mA regulation and exposure timing, so that flat output waveform is obtained. This generator also gives higher average X-ray energy, with shortest exposure time (1 ms). But these generators are bulky, higher cost, and involves inefficient power consumption.

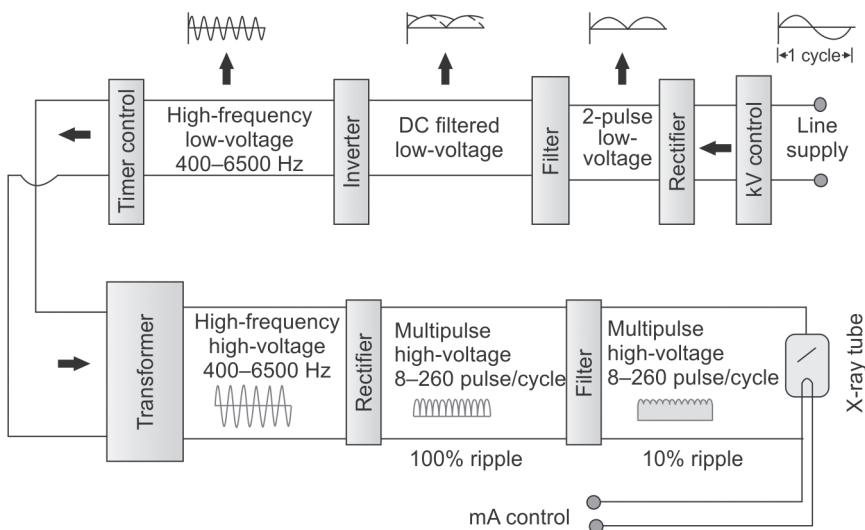


**FIG. 4.16:** Constant potential X-ray generator block diagram

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### **HIGH-FREQUENCY GENERATOR**

A high-frequency generator provides high-frequency AC waveform up to 50,000 Hz. After rectification and smoothing, it will provide a constant voltage supply. Both single phase and three phase line supply can



**FIG. 4.17:** Three phase high frequency X-ray generator

be used in high frequency generator. They provide accurate kV and mA with reproducibility. The high frequency generator transformers are efficient, compact and less costly.

A single-phase or three-phase AC line supply is rectified and smoothened, before fed into an inverter circuit (Fig. 4.17). The inverter gives a high frequency AC waveform, which is fed into the transformer. The transformer provides a fixed high voltage with low current. After rectification and smoothening, it is fed to the X-ray tube. At the same time two capacitors accumulate charges from the smoothened voltage.

When the X-ray tube is switched on, feedback circuits measure the difference between the set kV and the available kV. The comparator circuit generates trigger pulses, whose frequency is proportional to the difference of set and available kV. Based on the trigger pulse, the inverter produce corresponding output pulse. This is passed on to the transformer for further change in its output. The capacitor which has stored charges, increase the potential difference across the X-ray tube. Thus, the desired kV is obtained across the X-ray tube.

The feedback pulse rate depends upon the tube current and there is no need of autotransformer for kV control. The mA is also controlled similar to that of kV. If the available mA is low, the trigger pulse boosts the power to the filament and increase thermionic emission. The feedback circuit also eliminates the need for space charge compensation circuits and also correct for filament aging effects.

## EXPOSURE SWITCHING AND TIMERS

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### EXPOSURE SWITCHING

A switch is a device that turns the high voltage applied to the X-ray tube on and off. The current must be switched off very rapidly and it should remove all the energy that is stored in the voltage-smoothing networks. Then, only the voltage across the X-ray tube become zero. Improper switching off current creates high voltage spikes, which can damage the equipment. There are two categories of switching in X-ray generators. One method is switching in the primary circuit of the high voltage transformer, and the other is switching in the secondary circuit.

In a single-phase low-power X-ray generators, mechanical contactor is used as switch in the primary circuit. In this, a timer controlled electrical circuit energizes an electromagnet, which is connected to the contactor that closes the circuit. When the set time is elapsed, the electromagnet is de-energized and the contactor opens the circuit. So that the applied voltage is turned off across the X-ray tube. These switches are poor in accuracy and not useful for short exposures, i.e. < 8 msec. They depend upon the power in the circuit.

Three phase and constant potential X-ray generators use triode or tetrode switches on the secondary circuit of the transformer. They use electronic or phototimer with an accuracy of 1 msec. High-frequency generator uses electronic timers and switching on and off takes place in the primary side of the transformer. A rapid response is possible with an accuracy of 2 msec. A grid controlled X-ray tube can be used to switch ON and OFF exposure in any generator.

### EXPOSURE TIMERS

The exposure timers control the length of an X-ray exposure. There are two basic types of exposure timers, namely, electronic timers, and phototimers (automatic exposure control).

#### Electronic Timers

In electronic timer, the length of the X-ray exposure is determined by the time required to charge a capacitor through a resistance. When the exposure button starts exposure, it also starts charging a capacitor. The exposure time is terminated, when the capacitor is charged to a specified value, necessary to turn on associated electronic circuit. This time can be varied by varying the value of the resistance in

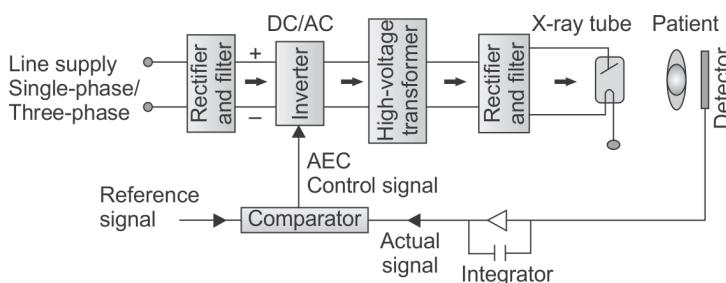
the charging circuit. Modern generators use electronic timers, in order to obtain very accurate exposure duration. These timers can accurately control exposures from  $< 1$  ms to  $> 1$  s. Of late digital timers are used instead of electronic timers, which has microsecond accuracy with good reproducibility.

### Phototimers (Automatic Exposure Control)

Phototimers terminate the exposure, when the X-ray receptor (film/screen) receives a preselected amount of radiation. They use any one of the radiation detector, namely, (i) ionization chambers, (ii) solid state diodes, and (iii) scintillators with photomultiplier tubes. In addition, there is an amplifier, density selector, comparator circuit, termination circuit and a back up timer. The X-ray coming out of the patient falls on the ionization chamber which is amplified, which is fed to the comparator and integrator circuit. When the integrated signal is equal to the pre-selected value, the exposure is terminated. In case of failure, back up timer terminates the exposure.

Most manufacturers use flat, parallel plate ionization chamber, which is mounted between the patient and the film. Since the chamber is radiolucent, it does not cause much shadow on the film. Solid state diodes and scintillators are mounted behind the cassette, to avoid their shadow on the film. A typical automatic exposure control (AEC) used in a high frequency generator is shown in Figure 4.18.

In order to match the film speed to the signal collected by the photo timer, proper calibration is required. Nowadays phototimers are provided with multiple settings, so that X-ray exposure may be increased or decreased in 10–20% increments. This will enable the operator to vary the film density for a given patient imaging. This is the most common type used today, which eliminates human error.



**FIG. 4.18:** Automatic exposure control with high frequency X-ray generator

## **QUALITY AND INTENSITY OF X-RAYS**

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### **QUALITY**

The term quality describes the penetrating power of the radiation. If a radiation consist photons of single energy (mono-energetic), then the quality can be described either by photon energy or wavelength. But, X-ray beam consist many photon energies (heterogeneous), and hence, its quality cannot be described by the photon energy. Therefore, the X-ray beam quality is usually specified by the following: half-value layer, applied voltage (kV), filtration and effective photon energy.

### **HALF-VALUE LAYER (HVL)**

The half-value layer or half-value thickness of a radiation beam is the required thickness of a material, which reduces the beam intensity to one half. The half-value layer is always stated together with the value of the applied voltage and the filtration. Aluminum and copper are the materials commonly used to specify HVL.

### **INTENSITY**

The intensity is a measure of quantity of radiation. The intensity of a radiation beam is the energy flowing in unit time through a unit area. It is equal to the number of photons in the beam multiplied by the energy of each photon. The intensity is commonly measured in roentgens per minute (R/min).

The term exposure is often used in radiology, which is proportional to the energy fluence of the X-ray beam. It refers both quality and quantity of the beam. The term quantity refers the number of X-ray photons in the beam.

### **FACTORS AFFECTING QUALITY AND INTENSITY**

The X-ray production efficiency, quality, quantity and intensity are affected by seven factors, namely, applied voltage, tube current, filtration, target material, exposure time, generator waveform and distance.

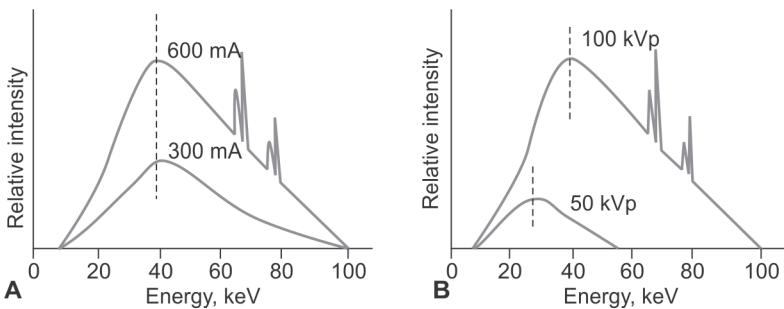
#### **The Applied Voltage ( $kV_p$ )**

The applied voltage affects both the quality and intensity of the X-rays. The energy of the photon emitted from the X-ray tube depends on the energy of the electrons that bombard the target. The energy of the electron is, in turn, determined by the peak kilovoltage used. As the applied voltage increases, the effective photon energy in the

bremsstrahlung also increases. The maximum photon energy is proportional to the peak value of the applied voltage. In addition, the X-ray production efficiency is related with applied voltage. The intensity increases with increase of applied voltage (Fig. 4.19). The amount of radiation produced increases as the square of the kilo voltage,

$$\text{i.e. Radiation exposure} \propto (\text{kV}_p)^2$$

Thus, increase in  $\text{kV}_p$  increases the quality, quantity and efficiency of X-ray production.



**FIG. 4.19:** Effect of (A) Tube current and (B) Kilovoltage on X-ray spectra

### Tube Current (mA)

The number of X-rays produced depends on the number of electrons that strike the target of the X-ray tube. The number of electrons depends directly on the tube current (mA) used. Greater the mA, higher the electrons that are produced, and hence, more X-rays will be obtained. The tube current affects only the intensity but not the quality of the X-rays. As the tube current increases the intensity also increases.

$$\text{i.e. the intensity is} \propto \text{mA.}$$

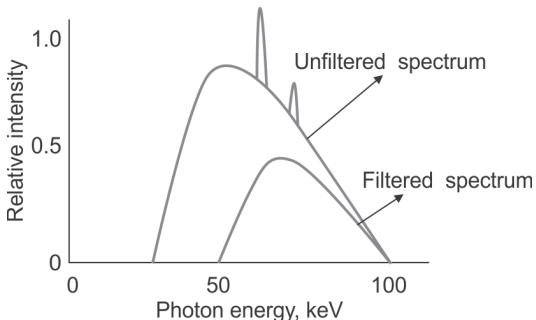
Increase of applied voltage is compensated by the reduction of tube current, which is required to maintain same exposure. The ratio of mAs varies with 5th power of  $\text{kV}_p$  ratio

$$(\text{kV}_{p1}/\text{kV}_{p2})^5 = \text{mAs}_2/\text{mAs}_1$$

### Filtration

Filters are thin sheet of material (Al,Cu,Mo), which offer high attenuation for low energy photons. The purpose of using filter is to reduce patient exposure at the skin level. Filters alter both the quality and quantity of X-rays by selectively removing the low energy photons in the spectrum.

This reduces the photon number (quantity) and shifts the average energy to higher values by increasing the quality. A filtered beam consists of higher photon energies and is said to be hardened (Fig. 4.20).



**FIG. 4.20:** Effect of filter on X-ray spectra

### Target Material (Z)

The atomic number of the target material affects the intensity of the X-rays. The intensity increases with increase of atomic number. The X-ray production is more efficient if higher the atomic number of the target material. For example, tungsten ( $Z = 74$ ) would produce much more bremsstrahlung than tin ( $Z = 50$ ), if both were used as target material under identical  $kV_p$  and mA.

The atomic number of the target material also determines the energy (quality) of characteristic X-rays. Thus, the atomic number of the target material determines the intensity of bremsstrahlung and quality of characteristic X-rays produced.

### Exposure Time

Exposure time determines the length of X-ray production. The total quantity of X-rays is directly proportional to the product of the tube current and exposure time (mAs).

### The Generator Waveform

The generator wave form (single-phase, 3-phase or constant potential) directly affects the quality of the emitted X-ray spectrum because of the average potential difference across the tube. For example, a single phase generator provides a lower average applied voltage potential difference, than a 3-phase generator.

### Distance

The X-ray beam intensity decreases with distance from the target because of the divergence of the X-ray beam. The decrease in intensity

is proportional to the square of the distance from the target. The nonlinear fall-off in intensity with distance is called the inverse square law. In general, if the distance from the X-ray source is changed from  $x$  to  $y$ , then the X-ray beam intensity changes by  $(x/y)^2$ . If  $x = 1$  m and  $y = 2$  m, then the intensity decreases by a factor 4. Thus, doubling the distance from the X-ray source, decreases the X-ray beam intensity by a factor of 4.

In summary, the intensity ( $I$ ) of X-ray radiation is given by the relation

$$I \propto \frac{kV_p^2 \times mAs \times Z}{d^2}$$

where,  $d$  is the distance between the target and the point of measurement.

## **GENERATOR RATING AND HEAT LOADING**

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The X-ray tube is designed for a particular type of application and hence, it should be loaded carefully. First, the focal spot must not be loaded beyond a certain power input. Second, the anode must not be loaded by successive exposures beyond a certain limit. Finally, the housing must not be expected to dissipate its energy at rates beyond a certain value.

If a X-ray tube is overloaded, enormous amount of heat will be produced. Therefore, the surface layers of the target may evaporate. The evaporated tungsten will deposit on the inner side of the glass tube by increasing the inherent filtration and reducing the tube output. The tungsten deposit may also cause irregular conduction paths within the tube, resulting in electrical instability and breakdown.

## **POWER RATING**

The term tube rating refers the maximum load that may be applied on the tube which may not cause any breakdown or damage to the tube. The power rating refers the energy supplied to the X-ray tube in one second and it is given the relation:

$$\text{Power (kW)} = 100 \text{ kV}_p \times A_{\max} \text{ for } 0.1 \text{ sec}$$

Thus, power rating is the average power delivered for 100 kVp and 0.1 sec for a maximum current of  $A$ . High power rating are seen in short exposures, large focal spots, higher anode rotation, small

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anode angle and large anode diameter. The power rating tells us the maximum combinations of  $kV_p$ , mA and time that may be used.

### HEAT LOADING

The thermal rating tells us the number of exposures that may be given in a given period of time. To know this, we have to calculate the energy deposited in the target in heat unit (HU) as follows:

$$\text{Energy (HU)} = kV_p \times \text{mA} \times \text{sec}$$

The above relation is true for single phase generators. It cannot be applied for three-phase and high-frequency generators, since their ripple factor is low. Hence, the above relation is multiplied by 1.35 for three-phase generators and 1.4 for constant potential generators. The thermal rating in HU is a artificial one and hence, the SI system of expressing energy in joule is proposed. This is given by the relation

$$\text{Energy (J)} = kV_p \times \text{mA} \times \text{s}$$

In a single phase system,  $kV_p$  is not a constant, and the average voltage is obtained by multiplying the  $kV_p$  by 0.7, the rms value

$$\text{Energy (J)} = 0.7 \times kV_p \times \text{mA} \times \text{s}$$

The rating in HU may be related to rating in joule as  $HU = 1.4 \times J$ . Thus, the X-ray tube ratings give all the necessary information for the safe operation of the tube. Rating charts are provided (Fig. 4.21) for easy and permissible operation of X-ray tubes. These include single exposure chart, and multiple exposure chart for angiography systems. Fluoroscopy use continuous radiation and needs anode heat input and cooling chart and housing heat input and cooling chart. Rating chart is influenced by the X-ray tube design and generator type. It is specific to a given X-ray tube and should not be exchanged.

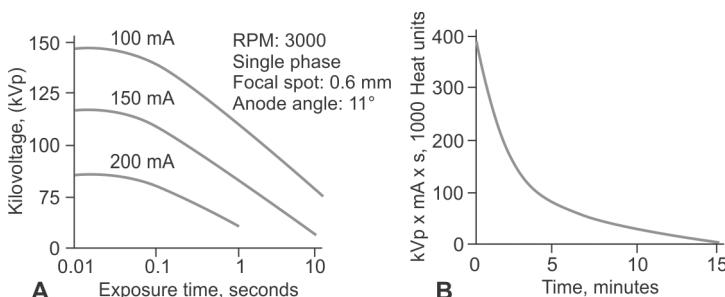


FIG. 4.21: Model (A) X-ray rating chart and (B) Anode cooling chart

# 5

# Radiation Units and Interactions with Medium

## RADIATION UNITS

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Radiation units are necessary to express quantities of physical entities in a numerical scale for comparative purposes. Quantification of physical entities is normally done by estimating a measurable physical effect of the physical entity, e.g. heat is quantified on the basis of expansion it produces in materials. The condition for quantification is the amount of physical entity, and its effect should have a linear relationship.

In radiological physics, the quantities of interest are the amount of (i) nuclear disintegration takes place per unit time, (ii) ionizing photons present in a field, (iii) energy transferred from radiation to tissue, (iv) energy absorbed in the tissue, and (v) biological effectiveness of energy absorption, etc. In 1981, the International commission on radiation units and measurements (ICRU) issued the units based on SI units as explained below.

### Fluence and Flux

The number of photons passing through a unit area is called fluence ( $\phi$ ) and its unit is  $\text{cm}^{-2}$ . The rate at which the photons are passing the unit area is called flux. The flux is the photon fluence per unit area per sec and its unit is  $\text{cm}^{-2}\text{s}^{-1}$ . The amount of energy passing through an unit area is called energy fluence ( $\psi$ ) and its unit is joules/meter<sup>2</sup>.

### ACTIVITY

Activity refers to number of unstable nucleus that regains stability through radio-disintegration per unit time. This information is significant as the quantity of radiation released from radioactive material is directly proportional to the activity. Activity gives an idea about the quantity of radiation released per time (disintegration rate) from the radiation source.

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The disintegration rate of a radioisotope is measured by the unit curie (Ci). One curie is the number of disintegration per second (dps) from 1 gram of radium (Ra-226) and it is found to be  $3.7 \times 10^{10}$  dps. Smaller units such as milli curie (mCi) and microcurie ( $\mu$ Ci) are also used to measure activity.

$$1 \text{ mCi} = 3.7 \times 10^7 \text{ dps}$$
$$\text{and } 1 \text{ } \mu\text{Ci} = 3.7 \times 10^4 \text{ dps.}$$

Becquerel (Bq) is the SI unit of activity, and it is equal to 1 dps. In practice, mega Becquerel (MBq) and giga Becquerel (GBq) are used as units and  $1 \text{ MBq} = 10^6 \text{ Bq}$  and  $1 \text{ GBq} = 10^9 \text{ Bq}$ . In day-to-day practice  $1 \text{ mCi} = 37 \text{ MBq}$  is a useful relation.

$$\text{Hence, } 1 \text{ Ci} = 3.7 \times 10^{10} \text{ Bq} = 37 \text{ GBq}$$

Since activity deals with nuclear emissions, it does not have any practical application in diagnostic radiology. These units do not tell us the dose delivered to the biological system. Hence, new quantities like exposure and absorbed dose are defined.

### EXPOSURE – ROENTGEN

Exposure indicates the amount of ionizing photons present in a field. The number of charged particles produced by ionization is directly proportional to the ionization events. Hence, ionizing photons are quantified on the basis of total charge produced by them in a given medium. Air is the universal medium to measure exposure. Average minimum energy necessary for photons to produce ionization in air is about 34 eV.

The term exposure (X) refers the radiation quantity measured in terms of ionization in air, in a small volume around a point. Exposure is a source related term. Exposure from an X-ray source obeys inverse square law. The unit of exposure is roentgen (R).

One roentgen (1928) shall be the quantity of x or gamma radiation such that the associated corpuscular emission per 0.001293 grams of air (1 cc of dry air at STP), produces in air, ions carrying 1 esu of quantity of electricity of either sign (STP refers standard temperature and pressure: 273 K and 760 mm of Hg). If the open air chamber is used, temperature and pressure variation can alter the amount of air molecules present in the chamber. Hence, temperature and pressure corrections are required.

In SI unit, these practical difficulties are avoided as volume of air is replaced with mass of air, which is not altered by temperature and

pressure variations. One exposure unit is defined as that amount of photons, which produces one coulomb of charge in one kilogram of air and it is equal to 1 coulomb/ kilogram in air (C/kg). The roentgen unit may also be defined in terms of SI unit as

$$1R = 2.58 \times 10^{-4} \text{ C/kg of air.}$$

In practice, following submultiples of roentgen are used:

$$1 \text{ milliroentgen (mR)} = 1/1000 \text{ roentgen} = (10^{-3} \text{ R})$$

$$1 \text{ microroentgen } (\mu\text{R}) = 1/1000 \text{ mR} = (10^{-6} \text{ mR or } 10^{-6} \text{ R})$$

Radiation monitors are usually calibrated in roentgen and milliroentgen (mR) and it is used to measure the output of X-ray machines. The output of a X-ray machine is often expressed in mR/mAs, e.g. a 75 kV X-ray unit with 2 mm Al filtration may give 5 mR/mAs at 100 cm distance. The measurement of exposure rate in air by an air filled ion chambers is easy, since the effective atomic number of air is equal to that of soft tissue. Thus the measured exposure is proportional to the dose in soft tissue in the diagnostic X-ray energy.

There are some difficulties in the unit of roentgen. It is not a unit of dose, which is a measure of absorbed energy. It can be used only up to a photon energy of 3 MeV. It is defined only for x and gamma radiations in air.

## **KERMA**

Kerma stands for Kinetic Energy Released in the Medium, which describe the initial interaction of the photon with an atom in the medium. When X and gamma rays passes through a medium, they transfer kinetic energy to the charged particles (electrons and protons). Kerma (K) is the measure of kinetic energy transferred to the charged particles. It is defined as the sum of the initial kinetic energy of all the charged ionizing particles, liberated by photons in a material of unit mass. The unit for kerma is jouls per kilogram (J/kg). The SI unit is gray and the special unit is rad. When the reference material is air, the quantity is called air kerma.

$$1 \text{ air kerma (Gy)} = 114 \text{ R}$$

## **Mass Energy Transfer Coefficient**

For X and gamma rays, kerma can be calculated from the mass energy transfer coefficient ( $\mu_{tr}/\rho_0$ ) of the material and the energy fluence ( $\psi$ ). The mass energy transfer coefficient is the product of mass attenuation

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coefficient and fraction of photon energy transferred as kinetic energy to the charged particle. It is always lesser than the mass attenuation coefficient. The scattered photon does not contribute to the kinetic energy of the charged particle. The ratio of the mass energy transfer coefficient and mass attenuation coefficient is 0.68 for 20 keV photons in tissue, it decreases to 0.18 at 50 keV, due to increase of Compton scattering. If  $\psi$  is the energy fluence and  $(\mu_{tr}/\rho_0)_E$  is the mass energy transfer coefficient at energy E then, kerma (K) is given by

$$K = \psi (\mu_{tr}/\rho_0)_E$$

### **ABSORBED DOSE – GRAY/RAD**

Exposure to radiation results in the transfer of energy from radiation to the interacting medium (kerma). Transferred energy may not be fully absorbed in the medium because, partly it is radiated out in the form of bremsstrahlung. Hence, absorbed dose is equal to the difference between kerma and the bremsstrahlung loss.

The term absorbed dose (D) refers the amount of energy absorbed per unit mass of the substance. The unit of absorbed dose is rad (r), which is a short form of radiation absorbed dose and  $1\text{rad} = 100 \text{ ergs/gram}$ . This unit is independent of type of radiation and the medium. The SI unit of absorbed dose is gray (Gy).

$$1\text{Gy} = 1 \text{ J/kg.}$$

$1\text{Gy} = 10^7 \text{erg}/10^3\text{g} = 10,000 \text{ erg/g} = 100 \text{ rad}$  (since  $1 \text{ rad} = 100 \text{ erg/g}$ ). Hence, the unit rad is related to gray as  $1\text{Gy} = 100 \text{ rads}$ . In practice, the following submultiples of gray are used:

$$1 \text{ milligray (mGy)} = 1/1000 \text{ Gray} = (10^{-3} \text{ Gy})$$

$$1 \text{ microgray (\mu Gy)} = 1/1000 \text{ mGy} = (10^{-6} \text{ Gy})$$

Since  $1 \text{ Gy} = 100 \text{ rad}$ ,  $1 \text{ mGy} = 100 \text{ mrad}$  and  $1 \text{ \mu Gy} = 100 \text{ \mu rad}$

The units rad or gray is defined based on the energy delivered in the medium and hence, it is suitable to describe a biological effect. It is used to quantify the radiation dose to the patient in radiotherapy.

### **Mass Energy Absorption Efficient**

The initial kinetic energy transferred to the charged particle is absorbed in the tissue as absorbed dose. In this, some part of the kinetic energy appear as bremsstrahlung X-rays, which escape the small volume of

tissues. The initial kinetic energy minus the energy that appears as bremsstrahlung X-rays is the true absorbed dose. It is represented by the mass energy absorption coefficient ( $\mu_{\text{en}}/\rho_0$ ) of the medium. The mass energy absorption coefficient is always lesser than that of mass energy transfer coefficient. However, in the diagnostic X-ray energy with low Z materials, both are almost equal, as the bremsstrahlung loss is very small.

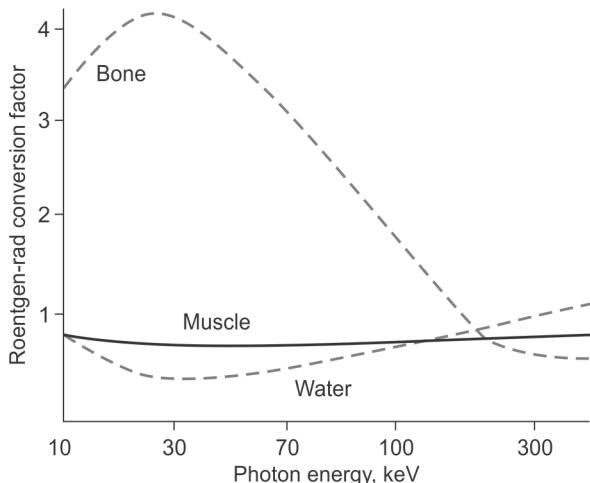
### ROENTGEN-RAD CONVERSION FACTOR

The absorbed dose (D) is related to the exposure (X) as follows:

$$D = f \times X$$

$$\text{i.e. Absorbed dose} = f \times \text{exposure}$$

where, f is the roentgen-to-rad conversion factor. In diagnostic X-ray energies, the f-factor for air, muscle and other soft tissue is close to 1. The factor is 4 for bone, because of higher photoelectric absorption. The variation of roentgen-rad conversion factor with photon energy is given in Figure 5.1.



**FIG. 5.1:** Roentgen-rad conversion factor with photon energy

### Conversion of Exposure from Dose in Air

Let D is the dose in air in Gy and X is the exposure in C/kg, then radiation exposure of 1 roentgen is given by

$$1 \text{ Roentgen (X)} = 2.58 \times 10^{-4} \text{ C/kg}$$

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The average energy deposited in air per ion pair is 33.97 J/C, and the total energy associated with 1 roentgen exposure is

$$\begin{aligned}1 \text{ Roentgen (X)} &= 33.97 \text{ J/C} \times 2.58 \times 10^{-4} \text{ C/kg} \\&= 0.00876 \text{ J/kg} \\&= 0.00876 \text{ Gy (since } 1 \text{ Gy} = 1 \text{ J/kg)} \\&= \text{dose in air (D)} \\1 \text{ Dose in air (D)} &= 0.00876 \text{ Gy/Roentgen}\end{aligned}$$

Thus, dose in air is equal to 0.00876 Gy per roentgen or 8.76 mGy per roentgen. Based on the above, the following relations may be used between dose and the exposure:

$$\begin{aligned}\text{Dose in air (mGy)} &= 8.76 \times \text{Exposure (R)} \\1 \text{ Dose in air } (\mu\text{Gy}) &= 8.76 \times \text{Exposure (mR)}\end{aligned}$$

### **RELATIVE BIOLOGICAL EFFECTIVENESS (RBE)**

All radiations are capable of producing same type of biologic effects, but the magnitude of the effect per unit dose differs. In other words, different radiations of equal dose do not produce the same level of biologic response. To evaluate the effectiveness of different radiations, the term "relative biological effectiveness" (RBE) was introduced and it is defined as follows:

$$\text{RBE} = \frac{\text{Dose of } 250 \text{ kVp X-rays required to produce certain effect}}{\text{Dose of reference radiation required to produce the same effect}}$$

The effects or endpoints include chromosomal mutation, cataract formation and acute lethality, etc. The RBE depends upon the linear energy transfer (LET) of the radiation in the medium. The linear energy transfer (LET) is a parameter that describes the average energy deposition per unit path length of the incident radiation and it is expressed in keV/ $\mu\text{m}$ . RBE also depends on the total dose and dose rate of the radiation. The RBE values of different radiations are as follows;

X-rays, $\gamma$ rays, electrons	:	1
Thermal neutrons	:	5
Fast neutrons, protons	:	10
Heavy particles	:	20

### **EQUIVALENT DOSE AND EFFECTIVE DOSE**

Exposure to radiation may be whole body exposure or partial body exposure. That means it can be uniform or nonuniform exposure. Radiosensitivity of tissues in human body varies towards radiation

exposure. Individual tissues contribute differently to the total detriment of health of the exposed person, depending upon the seriousness of the damage and its curability.

### Equivalent Dose

Hence, all radiation cannot cause the same biological damage per unit dose. In order to account the above factors, ICRP (Report-60,1990) has introduced radiation weighting factor ( $w_R$ ). It will modify the dose to reflect the relative effectiveness of the type of radiation causing biological damage. The product of the absorbed dose and radiation weighting factor is called equivalent dose (H):

$$H = D \times w_R$$

where, D is the absorbed dose and  $w_R$  is the weighting factor for the radiation type. The weighting factor is 1 for X-rays, gamma rays, and electron of all energies. High LET radiation may cause higher biological effect, hence have higher radiation weighting factor. The radiation weighting factor for various radiations are given in Table 5.1

The SI unit of equivalent dose is sievert (Sv) and 1 Sv=1J/kg. Rem is the special unit of equivalent dose, used early, when absorbed dose is measured in rad. Rem is the short form of radiation equivalent men and 100 rem is equal 1 Sv. In practice, millisievert (mSv) is mostly used and

$$1 \text{ Sv} = 1000 \text{ mSv}$$

$$100 \text{ rem} = 1000 \text{ mSv} \text{ (since } 1 \text{ Sv} = 100 \text{ rem)}$$

$$100,000 \text{ mrem} = 1000 \text{ mSV} \text{ (1 rem} = 1000 \text{ mrem)}$$

Hence, 100 mrem = 1 mSV or 100 mR = 1 mSV (if f- factor is 1)

**TABLE 5.1 Radiation weighting factors ( $w_R$ )**

Radiation type	$w_R$
Photons (all energies)	1
Electrons	1
Neutrons, < 10 keV	5
Neutrons, 10 keV–100 keV	10
Neutrons, 100 keV–2 MeV	20
Neutrons, 2 MeV–20 MeV	10
Neutrons > 20 MeV	5
Protons	5
Alpha particles	20

**TABLE 5.2 Tissue weighting factors ( $W_T$ )**

Tissue	$W_T$ (ICRP 2005)
Bone marrow	0.12
Breast	0.12
Colon	0.12
Lung	0.12
Stomach	0.12
Bladder	0.05
Esophagus	0.05
Gonads	0.05
Liver	0.05
Thyroid	0.05
Bone surface	0.01
Brain	0.01
Kidneys	0.01
Salivary glands	0.01
Skin	0.01
Remainder	0.10

### Effective Dose ( $H_T$ )

To account for the variation of radiosensitivity of different tissues and the non-uniformity of radiation exposure, ICRP has established tissue weighting factors ( $W_T$ ) as given in Table 5.2. The weighting factor of a particular tissue or organ is the risk of stochastic effects being induced in the organ when singly irradiated, compared to the total risk of inducing stochastic effects if the same radiation dose is received by the whole body. The sum of the products of the equivalent dose to each tissue irradiated ( $H_T$ ) and the corresponding weighting factor of tissue is called the effective dose ( $E$ )

$$E = \sum W_T \times H_T$$

where,  $W_T$  is the weighting factor of tissue  $T$  and  $H_T$  is the mean equivalent dose received by the tissue  $T$ . This quantity ( $E$ ) expresses the overall measure of health detriment associated with each irradiated tissue as a whole body dose and considers the radiosensitivity of each irradiated tissue. It is used to evaluate the probability of stochastic effects at low doses.

It was understood that testes and ovaries are the most radiosensitive tissues as per ICRP (1990). However, bone marrow and breast tissue

are the most radiosensitive organs as per ICRP (2005). Organ of higher sensitivity carries a higher risk for a given dose. The sum of the weighting factors is unity. The unit of effective dose is sievert (Sv). The tissue weighting factors are developed from a reference population, having equal number of sexes and wide range of age. The effective dose can be measured with the same unit sievert.

## **RADIATION INTERACTION WITH MATTER**

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When X or  $\gamma$  radiation passes through a medium, it interacts with an atom and produces moving electrons. These electrons travel in the medium, interact with other atoms and produce ionization and excitation. As a result, energy is deposited on the cells, which are either damaged partially or completely. In addition, sufficient amount of heat is also produced. In summary, the x or  $\gamma$  photon transfer energy to the electrons, which in turn transfer the energy to the cell system and produce the biological effect. That is why they are called as indirectly ionizing radiations.

The above interaction is said to have wavelike and particle like properties. X and gamma rays interacts with structures that are similar in size to their wavelength. Low energy photons tend to interact with atoms, medium energy to that of electrons and high energy photons with that of nuclei. The above structural level interactions may be performed by five mechanisms, namely (i) coherent scattering, (ii) photoelectric absorption, (ii) compton scattering, (iii) pair production, and (v) photodisintegration. The compton scattering and photoelectric absorption are the two most important interactions in diagnostic radiology.

## **ATTENUATION**

Attenuation is the product of absorption and scattering. It is the removal of photons from the beam due to absorption and scattering. If a beam passes through an absorber of thickness  $x$ , both absorption and scattering takes place (Fig. 5.2). As a result, the transmitted beam will have less number of photons and it is given by the relation

$$I = I_0 e^{-\mu x}$$

where  $I$  is the number of transmitted photons,  $I_0$  is the number of incident photons,  $e$  is the base of natural logarithm and  $\mu$  is the linear attenuation coefficient of the absorber material.

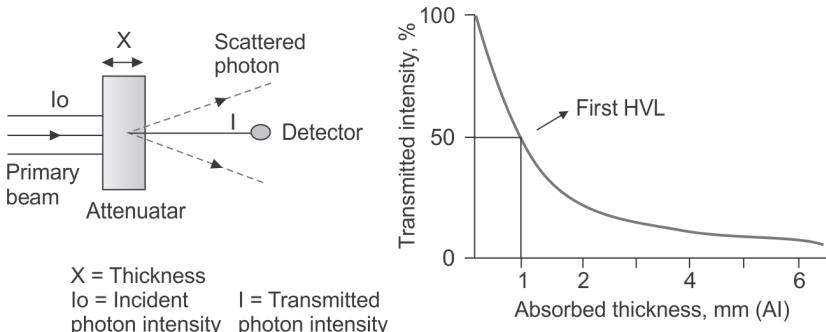


FIG. 5.2: Attenuation through an absorber

### Linear Attenuation Coefficient

The linear attenuation coefficient is defined as the reduction in the radiation intensity per unit path length and its unit is  $\text{cm}^{-1}$ . It refers to the fraction of photons removed from a monochromatic radiation beam. The above equation may be simplified and rewritten as:

$$\mu = - \frac{1}{x} \log_e \frac{I}{I_0}$$

The term  $I/I_0$  represent the fraction of photons removed, per unit thickness of the medium. Since the relation is logarithmic, higher numbers of photon are removed at the initial thickness of the absorber and lesser number at later thickness. That is why the relationship is exponential and the radiation intensity never reduces to zero. The linear attenuation coefficient depends on the energy of the photons and material density. Hence, linear attenuation coefficient varies with density for the same material.

In diagnostic energy range (30–100 keV), the linear attenuation coefficient decreases with increasing energy except at K-edges. The  $\mu$  for soft tissue ranges from 0.35 to 0.16  $\text{cm}^{-1}$ .

### Mass Attenuation Coefficient

The mass attenuation coefficient is obtained by dividing the linear attenuation coefficient by the density ( $\rho$ ) and has the symbol  $\mu/\rho$  and unit  $\text{cm}^2/\text{g}$ . The mass attenuation coefficient is independent of density. It is used to quantify the attenuation of materials independent of their physical state. The product  $\rho X$  is called mass thickness, and it is expressed in  $\text{g}/\text{cm}^2$ .

**Worked Example 5.1**

Calculate the linear attenuation coefficient of a material of thickness 1.8 mm, which reduces the intensity to 50%.

Here  $x = 1.8 \text{ mm}$ ,  $I/I_0 = \frac{1}{2} = 0.5$ ,  $\log 0.5 = 0.693$

$$\mu = - (1/x) \times 0.693 = 0.38 \text{ cm}^{-1}.$$

**Attenuation Coefficient and Beam Energy**

The attenuation coefficient varies with photon energy. The linear attenuation coefficient is defined for monoenergetic beams. There is difficulty in the application of this coefficient to polychromatic beams. Basically X-rays are polychromatic radiation beams. The low energy components are removed, while it is passing through the matter. Hence, the effective energy of the beam increases, resulting beam hardening effect. In addition, X-ray tube and filter also hardens the beam. Hence, diagnostic X-rays are heavily filtered beam that can be approximated to monochromatic X-rays.

In computed tomography, the CT number is defined on the basis of linear attenuation coefficients. Therefore, energy of the X-ray beam also influences the CT number. Same tissue lying in front and back of the body will experience different beam energy due to body attenuation. Hence, the attenuation coefficient will vary at these points, resulting variation of CT numbers. Similarly, different CT machines may give different CT numbers, due to variation of X-ray beam energy.

**Half Value Layer**

The half value layer (HVL) is the thickness required to reduce the beam intensity to half of its original value. The linear attenuation coefficient is related to the term half value layer as follows:

$$\text{HVL} = 0.693/\mu$$

HVL is an indirect measure of photon energies. It refers the quality of the beam, in a narrow beam geometry. A narrow beam geometry is similar to an experiment, in which scattered photons are not accounted by the detector. In the case of broad beam geometry, the beam is wider and scattered photons are present always. If measurements are done under broad beam geometry, it will under estimate attenuation. Most of the patient imaging conditions is broad beam geometry. If a attenuator has  $n$  number of HVL thickness, then the reduction of beam intensity is given by the relation  $(1/2)^n$ . Table 5.3 indicates the decrease of radiation intensity with increasing HVL's (for heavily filtered X-ray beam).

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**TABLE 5.3** Relation between Half value layer and % of transmission

No. of HVLs	% transmission
0	100.0
1	50.0
2	25.0
3	12.50
4	6.25
5	3.12
6	1.56

HVL is measured by thickness of aluminium in diagnostic X-rays and it is given for its effective energy. For diagnostic X-ray beam energies, the HVL for soft tissue ranges from 2.5 to 3.0 cm.

There is another term called tenth value layer (TVL), which gives the thickness of material that attenuates an X-ray beam by 90%. This quantity is very much useful in room barrier or shielding calculations. The TVL and HVL can be related by using the equation:

$$\begin{aligned} \text{TVL} &= 2.303/\mu \\ &= 2.303/(0.693/\text{HVL}) \\ &= 3.32 \text{ HVL} \end{aligned}$$

### Worked Example 5.2

Calculate the HVL of a X-ray beam, which passes through an absorber whose linear attenuation coefficient is  $0.35 \text{ cm}^{-1}$ .

$$\mu = 0.35 \text{ cm}^{-1}$$

$$\text{HVL} = 0.693/0.35 = 1.98 \text{ cm}$$

### Worked Example 5.3

A X-ray beam passes through an absorber of thickness 2 mm with transmission of 25%. Calculate the HVL of the beam.

Here,  $x = 2 \text{ mm} \text{ or } 0.2 \text{ cm}$ ,  $I/I_0 = 1/4 = 0.25$ ,

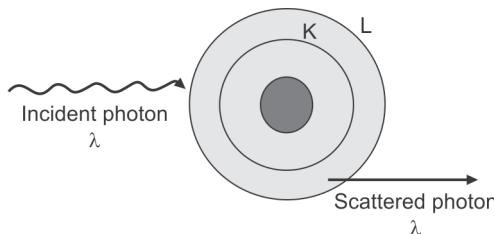
$$\mu = -(1/0.2) \times \log 0.25 = 6.93 \text{ cm}^{-1}$$

$$\text{HVL} = 0.693/\mu = 0.693/6.93 = 0.1 \text{ cm}$$

## COHERENT OR RAYLEIGH SCATTERING

The photon interacts with electron of an atom and sets the atom in the excited state. The excited atom releases excess energy as scattered X-ray with wavelength equal to the incident photons. The emitted radiation will have the same energy of the incident photon. But the direction

of the scattered photon is different to that of the incident photon. Thus, in coherent scattering, the photon undergoes a change in direction without change in wavelength (Fig. 5.3). In this process, no energy is transferred and no ionization occurs and most of the photons are scattered in the forward direction. The scattering angle increases as the X-ray energy decreases. This interaction occurs mainly in low energy photons, may be in mammography (15–30 keV).



**FIG. 5.3:** Coherent scattering, the incident photon and scattered photon have same wavelength

## COMPTON SCATTERING

In Compton scattering, a photon interacts with a free electron (valence) of an atom and gets scattered with partial energy (Fig. 5.4). The other part of energy is given to the valence electron, which is ejected from the atom. The ejected electron loses energy by ionization and excitation of atoms in the tissue, thereby contributing to patient dose. The scattered photon may travel in the medium with or without interaction in the medium by Compton scattering or photoelectric absorption. The scattered photon will have longer wavelength compared to the incident photon. The energy of the incident photon ( $E_0$ ) is equal to the sum of the energy of the scattered photon ( $E_{sc}$ ) and the kinetic energy of the ejected electron.

$$E_{sc} = \frac{E_0}{1 + [E_0 (1 - \cos \theta) / 511 \text{ keV}]}$$

where,  $\theta$  is the angle of scattered photon. As the incident photon energy increases, both photons and electrons are scattered in forward direction. The fraction of energy transferred to the scattered photon decreases with increase of incident photon energy, for a given angle of scatter.

If the photon makes a direct hit on the electron, the electron will travel straight forward ( $\phi = 0$ ) and the scattered photon will be scattered back with  $\theta = 180^\circ$ . In this collision, the electron will get its maximum energy, while the scattered photon goes with minimum energy. If the

photon makes a grazing hit with the electron, the electron will be emitted at right angles ( $\phi = 90^\circ$ ) and the scattered photon will go in the forward direction ( $\theta = 0$ ). In this collision, the electron receives minimal energy and the scattered photon goes with maximal energy.

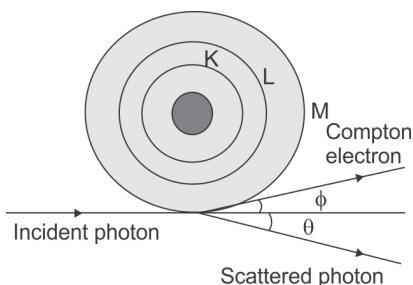
The Compton scattering involves an interaction between a photon and a free electron, resulting ionization of atom. The incident photon energy is shared between the scattered photon and ejected electron. The probability of Compton scattering depends upon the electron density (number of electrons per gram  $\times$  density) in the medium. Except hydrogen, the electron density is constant for tissue and it is independent of atomic number (Z). Probability of Compton interaction decreases with increase of X-ray energy and it is  $\propto 1/E$ . The probability per unit volume is proportional to the density of the material. Hydrogenous material has higher probability for Compton scattering.

Compton scattering occurs in all energies in tissue, important in X-ray imaging. It is a predominant interaction in the diagnostic energy range with soft tissue (100 keV-10 MeV). Scattered X-rays provide no useful information, reduces image contrast, create radiation hazards in radiography and fluoroscopy. In fluoroscopy, large amount of radiation is scattered from the patient and contributes to occupational radiation exposure.

## PHOTOELECTRIC EFFECT

In the photoelectric effect (PE), a photon of energy E collides with an atom and ejects one of the bound electrons from the K or L shells (Fig. 5.5). The ejected electron is called photoelectron and it has kinetic energy equal to  $E - \text{orbital binding energy}$ . In this process, all the incident photon energy is transferred to the electron. The incident photon must have energy equal or greater than the orbital binding energy of the electron, to perform photoelectric effect.

After the photoelectric effect, the atom is said to be ionized and there is a vacancy in the shell. This vacancy is filled by an electron of lower binding energy from higher orbit. This will create a cascade of electron transition event from outer orbit to inner orbit. The difference

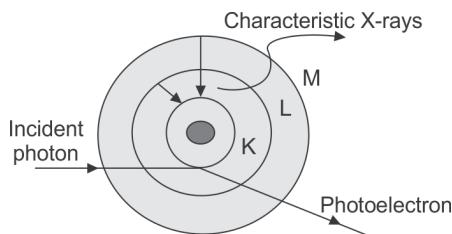


**FIG. 5.4:** Compton scattering

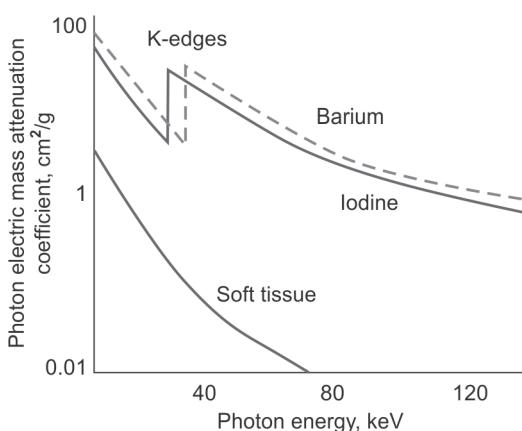
in binding energy is released as characteristic X-rays or Auger electrons. The photoelectric effect involves tightly bound electrons. The tightly bound electrons are mostly available in the K shell and hence, most photoelectric interactions occur at the K shell.

The probability of photoelectric cross section per unit mass is proportional to  $Z^3/E^3$ , where Z is the atomic number and E is the incident photon energy. As the X-ray photon energy increases, the subject contrast decreases. As the atomic number increases, the subject contrast increases, that is why barium ( $Z = 56$ ) and iodine ( $Z = 53$ ) are used as contrast agents.

Even though photoelectric effect decreases with increase of energy, there are exceptions. The absorption of photon increases markedly as the incident photon energy is increased from below to above the binding energy of the K-shell. This is known as K-edge absorption. The relation between probability of photoelectric absorption with photon energy may be plotted (Fig. 5.6). The elements exhibit sharp discontinuities called absorption edges. For example, the biding energy of the K-shell in iodine is 33.2 keV, which will have 6 times higher absorption at the K-edge.



**FIG. 5.5:** Photoelectric effect

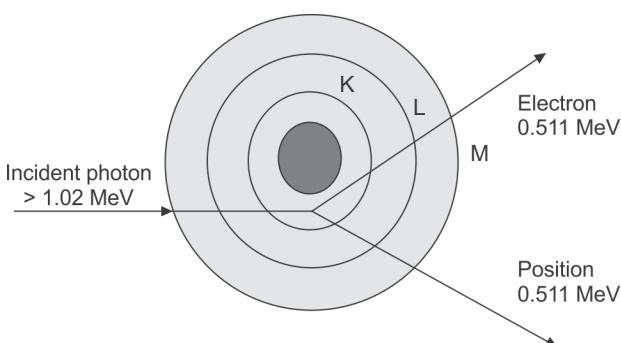


**FIG. 5.6:** Photoelectric absorption as a function of photon energy

The photon energy corresponding to an absorption edge increases with atomic number. The absorption edges of elements H,C,N, and O present in the soft tissue are well below  $< 1$  keV. The iodine, barium and lead absorption edges are 33.2, 37.4 and 88 keV, respectively. The photoelectric effect is very important in soft tissue imaging for photon energy  $< 50$  keV. It will differentiate attenuation between two tissues with slightly varying atomic number.

## PAIR PRODUCTION

When a photon having energy  $> 1.02$  MeV, passes near the nucleus of an atom, will be subjected to strong nuclear field (Fig. 5.7). The photon may suddenly disappear and become a positron and electron pair. For each particle 0.511 MeV energy is required and the excess energy  $> 1.02$  MeV, would be shared between the positron and electron as kinetic energy. Actually, the interaction is between a photon and the nuclear field. This process is an example for the conversion of energy into mass as predicted by Einstein. The threshold energy for the pair production process is 1.02 MeV. The probability of pair production increases with energy for a given material and also increases with atomic number ( $Z^2$ ). It is very important for photons having energy  $> 5$  MeV.



**FIG. 5.7:** Pair production

The electron loses energy by excitation and ionization and filling vacancy in the orbital shells. The positron travels in the medium and loses its energy by ionization, excitation and bremsstrahlung process. Finally, the positron combines with a free electron and produces two photons of each energy of 0.511 MeV that are ejected in opposite directions, to conserve momentum. Pair production is a true absorption because all the energy of the original photon is transformed.

### Positron Annihilation

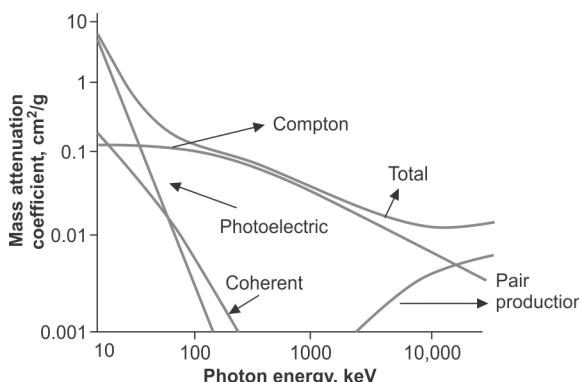
After the pair production process, the electron comes to rest, by joining with an atom. The positron comes to rest by combining with an electron and the two particles annihilates each other. The combined mass of the two particles is converted into energy in the form of two photons. The combined mass of two particles is 1.02 MeV and this energy is shared by the two photons. Hence, each photon will have energy of 0.511 MeV. The above process is called the positron annihilation. This is an example of conversion of mass into energy and forms the basis for positron emission tomography.

### RELATIVE IMPORTANCE OF ATTENUATION PROCESS

The total attenuation coefficient in tissue for a given energy, is the ability of the tissue to remove photons from the X-ray beams. It is caused by photoelectric absorption coefficient ( $\tau$ ), Compton attenuation coefficient ( $\sigma$ ) and pair production coefficient ( $\pi$ ). The total linear attenuation coefficient in tissue ( $\mu_{\text{Tot}}$ ) is given the relation

$$\mu_{\text{Tot}} = \mu_{\tau} + \mu_{\sigma} + \mu_{\pi}$$

Figure 5.8 shows the individual and total mass attenuation coefficients for soft tissue as a function of energy. At low photon energy (< 26 keV) photoelectric process dominates the attenuation in soft tissue. As the energy increases photoelectric absorption decreases. It is known that photoelectric absorption vary with  $Z^3$ . Hence, photoelectric absorption is more important than Compton scattering, in high Z materials and in low energy X-ray photons. It is the main mode of interaction with bone, contrast materials and screen phosphors.



**FIG. 5.8:** Mass attenuation coefficient for soft tissue as function of photon energy

Compton process is present at all energies and it is very important than photoelectric absorption, with low Z materials and in high energy X-ray photons. Both the processes are equally important at 30 keV energy for air, water and soft tissue, at 50 keV energy for bone and at 300 keV energy for iodine and barium contrasts. The pair production is not much important in diagnostic radiology, but is important in mega voltage radiotherapy.

Overall in the diagnostic X-ray energy, Compton dominates in air, water and soft tissue. Photoelectric absorption dominates in contrast media, lead and screen-film systems, whereas both process are important in bone.

## **IMPORTANCE OF INTERACTION IN TISSUE**

### **Differential Absorption**

When X-ray passes through the human body, (i) it partly interacts with the human body by Compton scattering, and photoelectric effect and (ii) partly transmits through the body without interaction. The Compton scattered X-ray do not give useful information for image formation. But it creates noise, by degrading the diagnostic image. Hence, suitable techniques are used to reduce scatter radiations that reaches the detector.

Photoelectric effect gives diagnostic information and helps the detector for image formation. Bone like anatomic structures are radiopaque, and shows high absorption characteristics, resulting light areas (white) in the radiographs. The X-rays that are transmitted through the body without interaction, reaches the detector, resulting dark areas (black) in the radiograph. The anatomical structures appear to radiolucent to above X-rays.

Hence, the radiographic image is due to the difference between the X-rays that are absorbed by photoelectric process and those transmitted with out interaction. This difference is called differential absorption. Reduction of kVp increases differential absorption, and image contrast, but with high patient dose.

### **Atomic Number**

The probability of photoelectric absorption is  $\propto Z^3$  of the soft tissue. The atomic number of bone and soft tissues are 13.8 and 7.4, respectively. The probability of photoelectric effect in bone is 7 times  $(13.8/7.4)^3$  more than in soft tissue. This probability decreases with increase of energy. Hence, in high X-ray energy, only few interactions occur and more X-rays are transmitted without interaction.

**TABLE 5.4** Mass density of tissues and contrast

Substance	Effective atomic number, Z	Mass density, kg/m <sup>3</sup>
Lung	7.4	320
Fat	6.3	910
Soft tissue, muscle	7.4	1000
Bone	13.8	1850
Air	7.6	1.3
Barium	56.0	3500
Iodine	53.0	4930

The Compton scattering is independent of Z of the tissue. The probability of Compton scatter is equal in bone and soft tissue and decreases with increasing energy. This decrease is slow compared to photoelectric absorption, which decreases rapidly. Hence, Compton scatter dominates at high photon energies.

### Mass Density

The interaction of X-rays with tissue is proportional to the mass density, regardless of the type of interaction. Mass density is the quantity of matter per unit volume expressed in kg/m<sup>3</sup>. It is related to mass of each atom and tells us how tightly the atoms are packed. The atomic number and mass density of various tissues are given in Table 5.4. If the mass density increases, the electron number increases, which accounts for higher interaction.

In addition to the Z related photoelectric effect, mass density also contributes to differential absorption. The X-rays are absorbed and scattered 2 times ( $1850/1000 = 18.5$ ) in bone than in soft tissue. Mass density helps in imaging lungs in radiography. In case of air ( $Z = 7.6$ ) and soft tissue ( $Z = 7.4$ ), the atomic numbers are almost same. But, an air filled soft tissue cavity can be imaged, due to their mass density difference.

The contrast agents barium and iodine has high atomic number and high mass density. Hence, they can be used with low kVp technique, to see internal organs. However, use of high kVp technique, will help in visualizing the lumen of the organ. In this case, the X-ray penetrates the contrast and helps to outline the organ.

### Photon Energy

In diagnostic radiology, the kilo voltage range is about 20–150 kVp and effective photon energy ranges from 15–100 keV. The relative

importance of interaction changes over this range. At low energy, photoelectric absorption is the main cause of attenuation than Compton scatter. Hence, soft tissue and bone appear as dark and light in the X-ray film. The thickness of tissue is also important in the attenuation process. Thick layer of soft tissue and thin layer of bone may produce same amount of attenuation. However, use of low energy with photoelectric effect interaction can differentiate the above, overcoming the thickness effect.

In high energy, Compton scatter is dominant and the differentiation between soft tissue and bone reduces. Though Compton scatter depends upon density, tissue differentiation is still possible with use of high kVp techniques.

Overall at low kV X-rays like mammography, about 75% of the attenuation in soft tissues is due to photoelectric absorption. The Compton scatter plays a minor role at this energy. At higher energy, the photoelectric absorption accounts for 15–20% attenuation in soft tissues and Compton plays a dominant role, e.g. chest radiography or gamma imaging.

## PARTICLE INTERACTIONS

Particle radiation includes alpha and beta particles, proton, electron, positron and neutrons. Alpha and protons are heavier particles compared to that of electrons and positrons. The behavior of heavy particles is different from lighter particles.

Charged particles interact with matter with electrical forces and lose energy in a medium by ionization and excitation. In this type of interaction, the charged particle interacts with orbital electron and loses energy. In excitation, the charged particle transfer its energy to the orbital electron and moves it to the higher orbits. The transferred energy should be less than the binding energy of the electron. The excited electron returns to lower energy by emission of characteristic X-rays and Auger electrons. About 70% of the charged particle energy is spent via excitation.

If the energy of the particle exceeds the binding energy of the electron, ionization occurs, where an electron is ejected from the atom. In ionization, an electron and a positive atom is produced, which are called as ion pairs. Sometimes, the ejected electron produces further ionization, known as secondary ionization. These electrons produced by secondary ionization are called delta rays. The energy required to produce an ion pair in soft tissue is 34 eV.

The ionization and excitation occur along the path of a charged particle in a medium and the rate at which the ion-pairs are formed depends upon the charge and energy of the incident particle and the atomic number of the medium.

### Specific Ionization

The number of primary and secondary ion pairs produced per unit path length of charged particle is called specific ionization, expressed in IP/cm. Specific ionization increases with particle charge and decreases with particle velocity. Larger particles interact with greater columbic field and lose energy and slows down. It permits greater time for interaction. Thus, the specific ionization of alpha particle is higher than that of proton and it is in the order of 7000 IP/mm.

The plot of specific ionization as a function of the particles path in the medium is shown in Figure 5.9. As the particle slows down, the specific ionization increases to a maximum, called the Bragg peak. Beyond the peak, the particle picks up electrons and become electrically neutral and hence, the specific ionization decreases rapidly. This property of the particle is used to treat cancer patient in radiotherapy. The particle delivers maximum radiation at a given depth, without exit dose. Protons are the well established external beam particle in precision radiotherapy.

### Path Length and Range

Electron undergo multiple deflection and attraction and hence their path is zig-zag, not in straight line. Large charged particle gives a dense and linear track as shown in Figure 5.10. The path length of a particle is the actual distance the particle travels in the medium. The range is the actual depth of penetration in the medium. The path length is always greater than range for electrons. But, in the case of heavy particle, the path length is equal to the range.

### Linear Energy Transfer

The amount of energy deposited per unit path length is called the linear energy transfer (LET) and it is expressed in eV/cm. LET of a charged

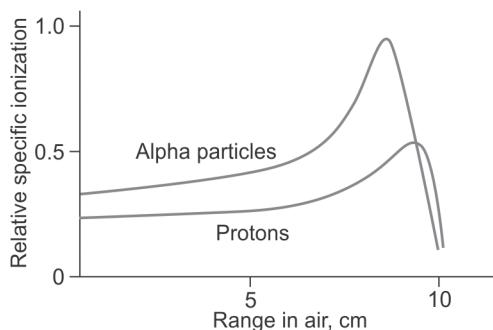
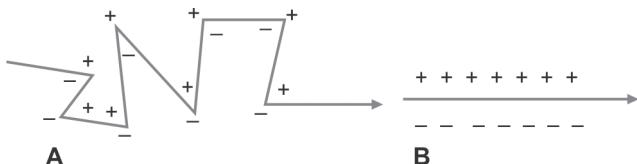


FIG. 5.9: Bragg curves for protons and alpha particles



**FIG. 5.10:** Path and range of a charged particle: (A) Electron path, path > range, (B) Alpha path, path = range

particle is proportional to its charge square and inversely to its kinetic energy. It is the product of specific ionization and the average energy deposited per ion pair. The LET tells the energy deposition density and largely determines the radiobiological effect of radiation. Alpha and protons are high LET radiations, whereas X-rays, electrons and positrons are low LET radiations. High LET radiation cause more damage than low LET radiation in a biological system.

### Scattering

Scattering refers the deflection of the particle from its original track. A scattering event, in which the total kinetic energy of the colliding particle is unchanged is called elastic scattering. A scattering event, in which loss of kinetic energy is involved is called inelastic scattering. If the binding energy is negligible, the ionization event is said to be elastic. The ionization event, in which the binding energy is significant is called inelastic.

### Electron Interaction

Electron particle enters in a medium, and undergoes both elastic and inelastic collisions. They lose energy by excitation, ionization and radiative process. When an electron is deflected by a positively charged nucleus, it loses kinetic energy, that appears as electromagnetic radiation. When electron undergo deceleration, bremsstrahlung radiation is emitted. When the electron energy is low, bremsstrahlung is emitted at angles of 60 and 90 degrees, relative to the electron track. If the electron energy is high, it is emitted in the forward direction. The probability of bremsstrahlung emission per atom is proportional to  $Z^2$  of the absorber. It also varies inversely with square of the mass of the particle. Hence, the bremsstrahlung emission from alpha and proton particles is very low. Bremsstrahlung rays have all possible energies and multiple electron interaction and gives continuous spectrum.

**Neutron Interaction**

The neutrons are uncharged particle and do not interact with electrons. Hence, they are called indirectly ionizing particles. They interact with light atomic nuclei (e.g. H,C), producing recoil nuclei that lose energy through excitation and ionization. In tissue, neutron interacts with hydrogen in water, producing recoil protons. Neutrons may also be captured by the atomic nuclei, with the emission of gamma rays. Neutron interaction has no role in diagnostic radiology and imaging.

# 6

# Radiation Detection and Measurements

## **RADIATION DETECTION**

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Radiations cannot be perceived by our normal senses, such as sight, feeling and smell. Hence, a suitable device is required, to detect and measure the amount and energy of the radiation. Generally, these devices consist of a detector in which interaction takes place and a measuring device to record the interaction. Some important effects on which the detection of radiation is based are: ionization, luminescence, photographic effect, thermoluminescence, chemical effect, and biological effect.

## **IONIZATION**

This effect consists of removing electrons from originally neutral atoms, or molecules, thereby giving rise to positive ions and negative ions. These are known as ion pairs, which can be collected by applying an electric field, to give rise to current or pulses. Gaseous and solid media are used for ionization. The ionization chambers, proportional counters, Geiger-Muller (GM) counters and semiconductor detectors fall under this category.

## **LUMINESCENCE**

This is the process in which radiation excites the atoms of a material and its energy is converted into visible light flash. Using light sensitive photomultipliers, the light flashes are converted into electrical pulses, which can be detected and registered with the help of electronic circuit. Detectors making use of this effect for radiation detection are known as scintillators. A scintillator coupled to a photomultiplier forms a scintillation detector.

## **PHOTOGRAPHIC EFFECT**

Similar to light, radiation also affects X-ray film. If the film is exposed by X or gamma rays, it will form a latent image, with black metallic silver. The degree of blackness can be measured by means of optical density, which is proportional to radiation exposure.

## **THERMOLUMINESCENCE**

Radiation can impart energies to certain crystalline materials (lithium fluoride, calcium sulfate) or certain glasses, which can store these energies for long time. The energy thus stored can be later released in the form of light or luminescence by heating these materials. The quantity of light released can be measured and correlated to radiation dose. The devices based on these effects are called TL dosimeters.

## **CHEMICAL EFFECTS**

Ionizing radiation can cause chemical changes, e.g. oxidation of ferrous sulfate to ferric sulfate. These effects can be quantitatively measured to correlate to the radiation dose. Some of these chemical systems may be used in conjunction with indicators to measure high personnel exposures. Radiation can also cause change of coloration in certain plastics. Such color changes can also be measured and correlated to radiation doses.

## **BIOLOGICAL EFFECTS**

Radiation exposure to the body can be measured by biological methods, e.g. analysis of blood for chromosomal aberrations in persons exposed to radiation doses of range 10–1000 rem. This is perhaps the only method of dosimetry, when there is no other information is available to assess the radiation exposure.

## **TYPES OF DETECTORS**

Radiation interacts with detector materials and deposits energy by ionization and excitation. The energy so deposited by a single interaction is very small and hence all detectors need signal amplification. In addition, detector system performs signal processing and signal storage with the help of electronic circuit. Signal processing can be done either by pulse mode or current mode. In pulse mode, the signal from each interaction is processed individually. In current mode, the

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electrical signals from individual interactions are averaged together, and give net current signal.

In pulse mode, two interactions must be separated by a time interval, so that it can produce two distinct signals. This time interval is called dead time of the detector. If interaction occurs in the dead time interval, either it will be lost or distort the signal from the previous interaction. Dead time depends upon the components in the detector system, e.g. multichannel analyzer in GM counter detector. Dead times of different detector system vary widely.

In current mode, all the information of individual interaction is lost. But, the electrical signal of each interaction is integrated to give the net electrical signal, which is proportional to the dose rate in the detector. Hence, current mode operation is suitable for high interaction rates, to avoid dead time loses.

### **Detector Efficiency**

Detector efficiency is a measure of its ability to detect radiation. In pulse mode, it is the probability that a particle/photon emitted by a source is detected. It is the product of geometric efficiency and intrinsic efficiency. Geometric efficiency is the fraction of emitted photon that reaches the detector.

$$\text{Geometric efficiency} = \frac{\text{Number of photons reaching the detector}}{\text{Number of photons emitted by the source}}$$

The intrinsic efficiency is the fraction of particle or photons that are detected.

$$\text{Intrinsic efficiency} = \frac{\text{Number of photons detected}}{\text{Number of photons reaching the detector}}$$

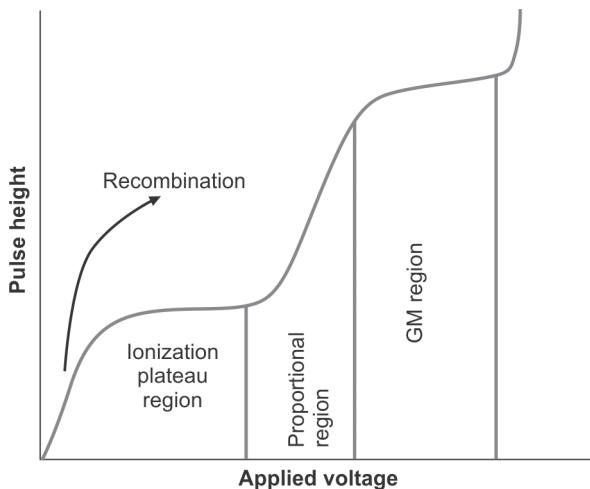
The intrinsic efficiency is often called quantum detection efficiency (QDE), which is determined by the energy of the photon, detector thickness, atomic number and density. The probability of detector efficiency varies from 0 to 1. It increases when the source is closer to the detector. It is 0.5 for a point source placed under a flat surface detector. It is 1 for well type detector system. Radiation detectors are classified as (i) gas filled detectors (ii) scintillation detectors, and (iv) semiconductor detectors, based on their detection mode.

### **GAS-FILLED DETECTORS**

Gas-filled detector has volume of gas in between two electrodes, in which a voltage is applied. When exposed to radiation, the gas is

ionized and ion pairs are formed. Positive ion moves towards negative electrode and negative ion moves towards positive electrode. The electron travels through the circuit, and reaches the cathode and recombines with positive ions. This forms an electrical current that can be measured by meter.

There are three type of gas-filled detectors, namely, ionization chamber, proportional counter, and Geiger Muller (GM) counter. These detectors are classified based on the applied voltage. Figure 6.1 shows detector current for various applied voltages between the electrodes. When the voltage is zero, the ion pairs produced by the radiation recombine and no current flows through the circuit. If a small voltage is applied, current start flows through the circuit. As the voltage increases further the current also increases, reducing the recombination of charges. This region is called *recombination region* of the curve.



**FIG. 6.1:** Gas-filled detector: Relation between applied voltage and pulse height

As the voltage is increased further no increase in current is observed and a saturation is reached. This is represented by a plateau in the graph stating that all the liberated charge is collected. This region is called *ionization chamber region*, and ionization chambers operate in this region.

If the applied voltage is increased further, the current also increases further and the region is called *proportional region*. In this region, the electron travels with high kinetic energy and cause additional ionization. This is called gas multiplication, which amplifies the detector

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current. The amplification increases as the voltage increases. In this region, the charge collected is proportional to the amount of energy deposited in the gas.

After the proportional region, the amount of charge collected is the same, regardless of energy deposited. This region is called *GM region*, in which the gas multiplication spread the entire length of the detector. GM counters cannot differentiate energy of the radiation. If the voltage is increased further, discharge takes place in the detector and should not be operated further.

### Ionization Chambers

An ionization chamber usually consists of an outer cylinder coated inside with graphite to make it conducting and a central electrode insulated from the chamber wall (Fig. 6.2). The cylinder is filled with either air or with suitable gas for radiation interaction and detection. Air requires 34 eV to produce 1 ion pair, and a 100 keV photon can create about 3000 ion pairs.

When the chamber is exposed to radiation, ion pairs are formed and are collected by the electrodes. The flow of ions through the circuit, create a current signal. The amplitude of the signal depends upon the number of ion pairs formed and independent of applied voltage. The amount of signal obtained from single interaction is very small and require amplification. Therefore, ion chambers are used in current mode, not in pulse mode. They are free from dead time losses and operated with wide range of voltages.

If the chamber is filled with air and the effective atomic number of the wall material is equal to air, then the amount of current produced is proportional to exposure rate. Thus, an average rate of energy dissipation in the chamber is measured. Such a measurement is often

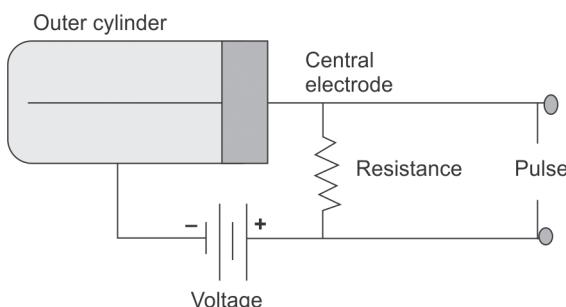


FIG. 6.2: Ionization chamber

called a dose-rate or exposure rate measurement. The minimum current that can be conveniently measured is about  $10^{-14}\text{A}$ . The ionization current is a measure of intensity of radiation. Such chambers are used as survey meters and dosimeters in radiotherapy. However, their intrinsic efficiency is low because of lower air density.

Instead of air, high atomic number gases, argon ( $Z = 18$ ) or xenon ( $Z = 54$ ) and pressurizing gases can be used, to increase the sensitivity towards X and gamma rays. Such chambers are used in isotope calibrator and CT scans as detector. The advantages of ion chamber are: (i) walls can be made tissue equivalent, (ii) all type of radiation can be measured and (iii) can be calibrated to any energy. The disadvantage includes small signal current which require high amplification and has restricted sensitivity.

### **Proportional Counter**

Proportional counters are designed with specific gas medium, operated with higher voltages (1000 V). Operating voltage varies with gas medium. In general, inert gases, such as nitrogen, and argon are used. Krypton and xenon are also used at high energies with higher efficiencies. Butane is a cheap high density gas but need frequent replacement. Gas pressure increases density and gives higher quantum detection efficiency.

The higher electric field accelerates the electrons to high kinetic energy, capable of producing secondary ionization. About  $10^3\text{--}10^5$ , secondary events are produced per initial ionization event. This is called amplification and it can produce 100–1000 times higher charge per interaction. The above secondary ionization is proportional to the initial ionization and hence the name proportional counters. They can be operated either in pulse mode or in spectrometers. Proportional counters provide large surface area and serve as detector in CT scan (xenon gas). They are also used in health physics research and industry.

### **Geiger Counter**

The Geiger Muller (GM) counter consists of a cylindrical cathode with a fine wire anode along its axis (Fig. 6.3). The device is filled with a special mixture of gases at a pressure of about 10 cm of Hg. When X or gamma ray passes through the counter, photoelectrons are ejected from the metal cathode. These electrons are accelerated by the large positive potential and gain energy. As a result, more electrons are

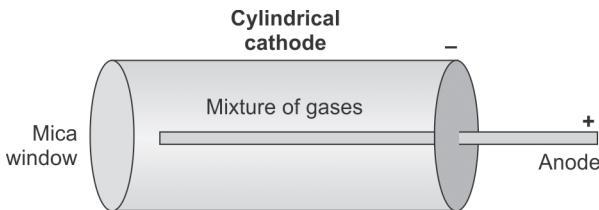


FIG. 6.3: Geiger-Muller tube

ejected from the gas atoms in the chamber, that create large number of ion pairs. Thus, a huge amplification in the order of  $10^{10}$  is obtained as signal.

In a single interaction, it can produce billions of ion pairs, hence require little amplification. The Geiger counter is operated at a higher applied voltage (900–1200 V). The plateau region is about 150–200 voltage long and independent of the applied voltage. It is usual to operate the counter at a voltage midway along the plateau.

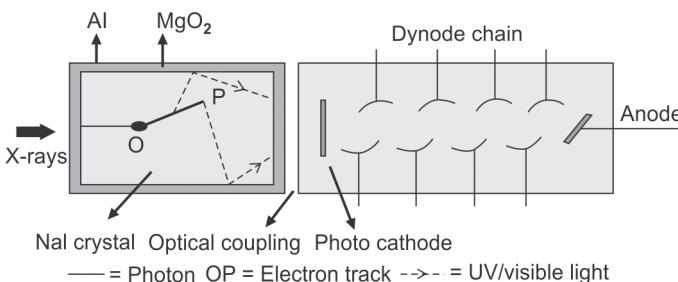
GM counter has higher efficiency for charged particles and record every particle separately. However, beta particle cannot penetrate the window. Hence, they are provided with window, that are opened for beta particle and low energy photon detection. But, they are inefficient towards X and gamma rays. GM counter voltage pulse size is independent of radiation energy, e.g. 1 keV and 5 keV radiation give same size pulse. Hence, they cannot be used as spectrometers and dose rate meters. The discharge produced by the ionization must be quenched, then only the counter can return to the original state and ready for next event. The GM counter is insensitive until quenching is complete, and this time period is called *dead time*. The dead time of GM counter is in the order of 100 ms, and not suitable for accurate measurements. GM survey meter get paralyzed in high radiation field and shows zero reading.

## SCINTILLATION DETECTOR

Scintillation crystal emits visible or ultraviolet light (UV), when exposed to radiation. This light can be seen by eyes that are dark adapted. However, signal needs to be amplified, hence all scintillation crystals are provided with photomultiplier tubes (PMT). Scintillation material type includes (i) organic compounds, and (ii) inorganic crystals. In organic compounds, the molecular structure decides the scintillation property and it is available in liquid form also. In inorganic crystal, the crystal structure gives the scintillation property. They have higher

average atomic number and higher density and widely used in radiology. Most of the inorganic crystals are grown with traces of impurity elements called activators, e.g. NaI: Tl

A scintillation detector consists of the following: (i) a luminescent material, (ii) an optical device to facilitate the collection of light, (iii) an optical coupling between the luminescent material and the photomultiplier, (iv) the photomultiplier tube, and (v) electrical circuit to record the pulse appearing at the output of the photomultiplier tube (Fig. 6.4).



**FIG. 6.4:** Scintillation detector with photomultiplier tube

When a photon interacts with the crystal, the electrons are raised to the excited state. The excited electrons return back to the low energy, with the emission of visible and UV light. This is called luminescence and each luminescence event has its own decay constant. That means the light emission takes some time to cease out, which is called as *after glow*.

These photons are radiated in all directions. The magnesium oxide layer reflects the photons towards the window side. The low energy light photons fall on the photo cathode of the multiplier tube and eject a number of photoelectrons. These photoelectrons are accelerated by the potential applied between the cathode and the dynode of the tube.

Finally, the PMT produces a voltage pulse in the output condenser, which is coupled to an external pulse amplifying circuit. Thus, the initial energy of a single ionizing particle, is transformed into a single voltage pulse. The whole system is enclosed in a light tight box to eliminate effects other than those due to incident ionizing radiation.

The number of electrons ejected from the photo cathode is proportional to the amount of light that reaches it. The amount of light is in turn proportional to the energy absorbed from the photon beam by the crystal. Thus, the size of the pulse that emerges from

the anode is proportional to the energy of the incident photon in the crystal. Hence, this detector can be used to distinguish between photons of different energy. Due to statistical variations within the crystal and PMT, there is a spread of pulses about the peak value.

Scintillation detectors are operated at pulse mode, where the after glow is less important and the electronic circuit can identify individual interactions. If it is operated in current mode, individual interactions can not be identified. Scintillation crystals are commonly used in gamma cameras and CT scanners.

### **Radiological Scintillators**

Sodium iodide ( $\text{NaI}: \text{Ti}$ ) is used in all nuclear medicine applications. It is used in gamma camera, thyroid probe, and gamma well counter, under pulse mode operation. It has high density (Iodine  $Z = 53$ ) and provides high photoelectric absorption probability for X and gamma rays. It has very high conversion efficiency (13%) and emits light promptly with a decay constant of 230 ns. It is possible to manufacture large crystals of size about 59 cm long  $\times$  44.5 cm wide  $\times$  0.95 cm thick. But, they are fragile and hygroscopic and require tight sealing.

Bismuth germinate ( $\text{Bi}_4\text{Ge}_3\text{O}_{12}$ ) is used as detector in PET scanners. The high atomic number (Bismuth,  $Z = 83$ ) and high density, provide high intrinsic efficiency for positron gammas with a decay constant of 300 ns.

Calcium tungstate ( $\text{CaWO}_4$ ) has been used in intensifying screens for radiography. It is replaced by gadolinium oxyorthosilicate with terbium and tantalite activators. Cesium iodide with thallium activator is used in thin film transistor technology in digital radiography.

In image intensifier, cesium iodide with sodium as activator and cadmium sulfide with silver activators are used as input and out-put phosphors.

In CT scans, scintillator coupled with photodiodes are used. Since the X-ray flux is very high, it requires current mode operation, to avoid dead time loses. High resolution CT scans operating with subsecond, require crystals with less after glow. Cadmium tungstate and gadolinium ceramics are commonly employed as scintillators in such CT machines. Table 6.1 summarizes the various scintillators used in radiology and their physical properties.

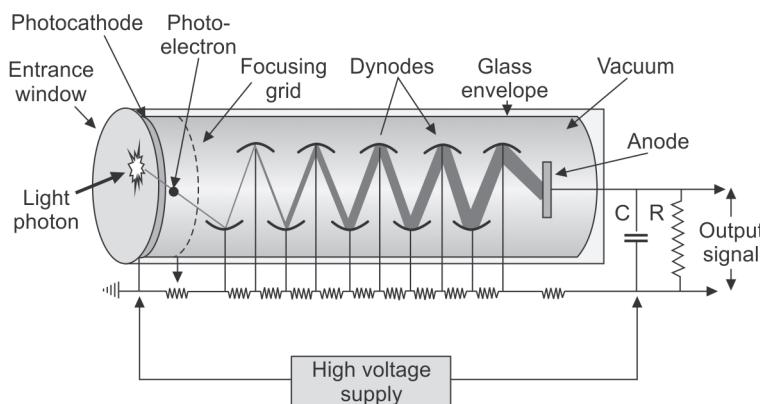
### **Photomultiplier Tube**

Photomultiplier tube (PMT) mainly converts visible and UV light into an electric signal and does signal amplification of the order of millions.

**TABLE 6.1** Radiological scintillators

Material	Density, g/cm <sup>3</sup>	Atomic No. (Z)	Conversion efficiency, %	Decay constant, μs	After glow ms, %
Nal:Tl	3.67	11,53	100	0.23	0.3–5
Bi <sub>4</sub> Ge <sub>3</sub> O <sub>12</sub>	7.13	83,32,8	12–14	0.3	0.1
CsI: Na	4.51	55,53	85	0.63	0.5–5
Cs: Tl	4.51	55,53	45	1.0	0.5–5
CdWO <sub>4</sub>	7.9	48,74,8	40	5	0.1
CaWO <sub>4</sub>	6.12	20,74,8	14–18	0.9–20	–
Gd <sub>2</sub> O <sub>2</sub> S:Tb	7.34	64,8,16	–	560	–

It mainly consists of an evacuated glass tube containing photocathode and 10–12 dynodes and an anode (Fig. 6.5). Photocathode is a thin electrode present inside the glass window. It emits electrons, when visible light incident on it. One electron is emitted for every 5 light photons. A power supply provides 1000 V and a series of resistors divide it into equal amount. The first dynode is given +100 V with respect to the photocathode. The other dynodes are connected with an increment of 100 V each. The electron emitted by the photocathode falls on the first dynode, and gets accelerated. They gain kinetic energy equal to the potential difference between photocathode and the dynode. Additional electrons (5 electrons per 1 incident electron) are produced in the dynode and similar process continues in the second dynode and so on. The total amplification is about  $5^{10}$ , for a 10 number dynode

**FIG. 6.5:** Photomultiplier tube

PMT system. The scintillator and the PMT is coupled together by an optical material, to reduce photon reflection loses.

### **Photodiode**

Photodiode is a semiconductor device that converts light into an electrical signal. It is connected in reverse bias voltage and no current flows in the circuit. When it is exposed to light, an electrical current is generated that is proportional to the amount of incident light. The electrical signal is not amplified, but involves noise. Photo diodes can be used with scintillator alone, without PMT. Photodiodes with CdWO<sub>4</sub> is used in CT scan as detector. It is also used in digital radiography with thin film transistors. Photodiodes are smaller in size and cheaper.

### **Thermoluminescent Dosimeters**

Thermoluminescent (TL) detectors are organic scintillators, which are used in personnel monitoring and patient dose estimation. When exposed to ionizing radiation, valance electrons are raised to the conduction band and trapped in the forbidden gap. The electron traps are well below the conduction band at room temperature. The internal energy of the phosphor is not sufficient enough, to raise the trapped electrons to the conduction band. While heating, the internal energy is raised and some of the trapped electrons escape to the conduction band. These electrons reach the ground state, with emission of light. The amount of light emitted is proportional to the amount of energy absorbed by the TL material.

The intensity of light emission changes with increasing temperature. A plot showing the emission of light intensity against temperature is called glow curve. The peak light intensity is proportional to the radiation dose received. Measurement of peak light intensity forms the basis for thermoluminescent dosimetry (TLD).

Lithium fluoride is a useful TLD material with little fading. Its effective atomic number is close that of tissue. Hence, the light emission is proportional to the tissue dose, over the range of X and gamma energies. It is effectively used for tissue dose monitoring purpose. It is available in the form of powder, rods, discs or chips made up of Teflon (PTFE) impregnated with lithium fluoride. It can be worn by a person, or inserted into a body cavity or pasted on a equipment.

Usually, the heating and the measurement of peak light intensity are done in a special device called TLD reader, after radiation exposure.

### Photostimulable Phosphors

Photostimulable phosphors are scintillators that are used in imaging plate technology. When exposed to radiation, fraction of excited electrons is trapped as absorbed energy. These electrons are released by scanning the plate by laser beam (700 nm). The laser light stimulates the trapped electrons and release visible light. The light may be collected by means of a fiberoptic guide tube and passed on a photomultiplier tube (PMT). The PMT produces an electronic signal. The amount of light emitted is proportional to the radiation exposure. The signal can be digitized and stored. The color of the light that is emitted is different from the laser light.

Imaging plate area made from 85% BaFBr and 15% BaFl, activated with europium (Eu). The Eu creates defects in the crystal and permits electron trapping. When exposed to X-radiation, the electrons in Eu atoms get excited. This makes the divalent Eu ( $\text{Eu}^{+2}$ ) atoms get oxidized into trivalent Eu atoms ( $\text{Eu}^{+3}$ ). The excited electrons are mobile and a fraction of them interact with the F center (Fig. 6.6). The F center traps the electron, and store them over a period of time. In the case of a imaging plate, billions of electrons are trapped in F centers. The number of electrons trapped is proportional to the intensity of exposure at that location.

When the plate is scanned with red laser light, the F center absorbs energy and transfer to the electron. The electron gains enough energy and comes to the conduction band where they become mobile. They reach the valence band with the emission of blue and green light. The electrons are absorbed by the trivalent Eu atoms, which changes into divalent state.

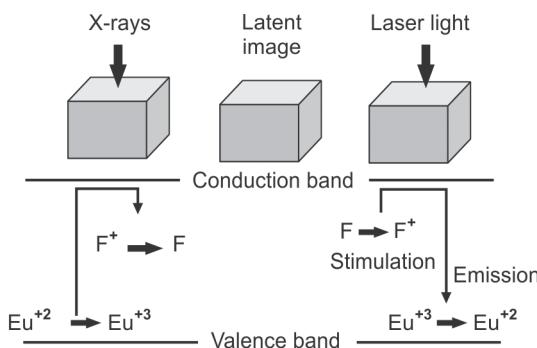


FIG. 6.6: Principle of photostimulable phosphor

This is the principle used in computed radiography (CR). Phosphors such as  $\text{Gd}_2\text{O}_2\text{S}$  and  $\text{BaFBr}$  and  $\text{BaFI}$  can be used as photostimulable phosphors in CR. The imaging plate can be reused by erasing the entire trapped electrons. This is possible with a exposure of bright light, which return all the meta-stable electrons to the valence band.

## **SEMICONDUCTOR DETECTOR**

A semiconductor diode with reverse bias voltage supply can be used to detect visible and UV light. When the diode is exposed to light photons, the low energy electrons (valence band) are excited in the depletion region, and raised to high energy state (conduction band). The holes move towards the p-type semiconductor and electron moves towards the n-type semiconductor. This produces a momentary current flow in the circuit and forms the voltage signal.

When the diode is exposed to ionizing radiation such as X and gamma rays, they produce ionization and excitation that also raise the electrons from valence band to conduction band. Rest of the action is similar to that of light photon exposure. The sensitive volume is the depletion region, in analogy with volume of gas in the ion chamber. Hence, the semiconductor behaves like a parallel plate ion chamber. The amount of energy required is 3 eV to create a electron-hole pair, compared to that 34 eV in ion chambers.

The amount of charge generated per interaction is proportional to the energy deposited in the detector. That means, that the size of the voltage pulse is proportional to the energy deposited in the detector. The voltage pulse is larger than ion chamber and its amplitude is photon sensitive. The pulse raise time is shorter, since the electron, hole move rapidly. Hence, they can be used as spectrometers.

The intrinsic noise is higher, due to (i) semiconductor resistance, (ii) thermal energy produced electron-hole pair, and (iii) charge leakage at the surface. Silicon based p-n junction diode reduce noise significantly than germanium at room temperature. They must be cooled with liquid nitrogen, to avoid excitation of electrons due to thermal energy. Especially germanium based semiconductor must be cooled at  $-190^\circ\text{C}$ , to reduce noise. The intrinsic noise increases with detector volume, and hence in larger volume detectors cooling is a must.

The pulse is narrower than that of ion chambers and hence energy resolution is better compared to ion chambers and scintillation detectors. The efficiency is 100% and its response is linear with exposure.

It is independent of radiation type, and the absorption of energy at the entry window is smaller.

They can be made in small sizes and capable of measuring from 1  $\mu\text{Gy}$  to 16 Gy. Its dose rate response is linear from 5  $\mu\text{Gyh}^{-1}$  to 3  $\text{Gyh}^{-1}$  and can be calibrated better than 5%. It is stable over wide temperature range (-20 to +80°C) and not hydroscopic. Since the effective atomic number of silicon is different from air, it is not tissue equivalent.

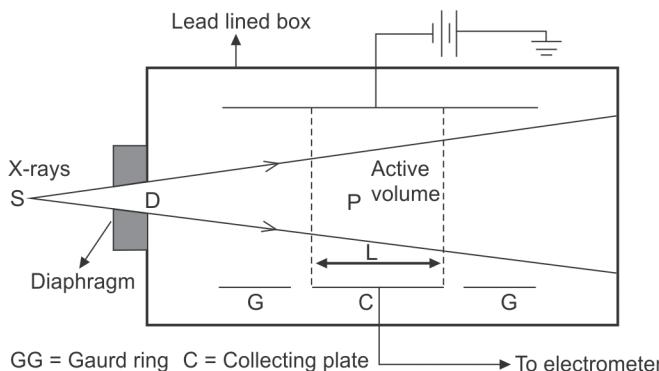
Semiconductor detectors are used in kVp meters, digital pocket dosimeters (silicon), gamma ray spectrometry (Germanium) and X and gamma dosimetry. They can also be used as photodetectors in flat panel detectors.

## PRACTICAL DOSIMETERS

### FREE AIR IONIZATION CHAMBER

Free air ionization chamber is an instrument employed in the measurement of exposure in roentgen, according to its definition. It is a primary standard used only for the calibration of secondary instruments.

A free air chamber consists of a box of air as shown in Figure 6.7. An X-ray beam from a source S, passes into a shielded box, which has two parallel plates. A known mass of air is exposed by the X-ray beam and the charge is collected by the two metal plates. A high voltage is applied between the plates to collect ions produced in the specified volume of air. The potential should be sufficient enough to collect all the charges produced in air, before they recombine.



**FIG. 6.7:** Free air ionization chamber

The total charge collected is measured by an electrometer. The whole chamber is provided with lead lining, to prevent the entry of external radiation into the chamber.

The beam size is controlled by a diaphragm D. The ionization is measured for a length L of the collection plate C. The lines of force are made straight and perpendicular to the collector by a guard ring G. If Q is the charge collected, and  $\rho$  is the density, then the exposure ( $E_D$ ) at the diaphragm is:

$$E_D = \frac{Q}{\rho \times A_D \times L \times 2.58 \times 10^{-4}} \times R$$

where,  $A_D$  is the diaphragm aperture area and  $(\rho \times A_D \times L)$  represents mass (density  $\times$  volume) of air in the specified volume. Thus, exposure measurement, involves only the measurement of Q, L and  $A_D$ . To satisfy the definition of roentgen, there must be an electronic equilibrium in the specified volume. This means that the number of electrons leaving the specified volume must be equal to number of electrons entering into it. At the same time, variation of the beam over the distance L should not be too large, so as to cause any attenuation of the beam. Further, X-ray beam should have a uniform intensity over the volume under consideration. The electronic equilibrium decides the size of the chamber.

The temperature and pressure should be measured during the measurements, which influence the air density. Whenever it deviates from the standard temperature and pressure, suitable correction needs to be applied.

### Temperature and Pressure Correction

The response of ion chamber is affected by air temperature and pressure, since the density of air depends on the temperature and pressure. The density or mass of air in the chamber volume will increase as the temperature decreases, or pressure increases. As a result, the chamber reading for a given exposure will increase. The chambers are usually calibrated under standard atmospheric conditions (760 mm Hg, 22°C). The correction factor  $C_{T,P}$  for conditions other than the calibration conditions is:

$$C_{T,P} = \frac{760}{P} \times \frac{(273 + t)}{295}$$

where,  $p$  is the pressure in mm of mercury and  $t$  is the temperature in degree Celsius. The temperature ratio is expressed in Kelvin, the absolute scale.

### *Worked Example 6.1*

Calculate the exposure in a mass of air having a volume of 1 liter and density  $1.18 \text{ kg m}^{-3}$ , in which  $10 \mu\text{C}$  charge is collected.

$$\begin{aligned}\text{Mass (m)} &= \rho \times V = 1.18 \times 1\text{L} = 1.18 \times 10^{-3} \\ &\quad (\text{1L} = 1000 \text{ cm}^3 = 10^{-3} \text{ m}^3) \\ &= 1.18 \times 10^{-3} \text{ kg}\end{aligned}$$

$$\text{Charge (Q)} = 10 \mu\text{C} = 10 \times 10^{-6} \text{ C}$$

$$\begin{aligned}\text{Exposure} &= Q/m \\ &= \frac{10 \times 10^{-6} \text{ C}}{1.18 \times 10^{-3} \text{ kg}} \\ &= 8.47 \times 10^{-3} \text{ C/kg, dividing by } 2.58 \times 10^{-4} \\ &\quad \text{which is equal to } 1\text{R} \\ &= 32.8 \text{ R}\end{aligned}$$

## THIMBLE IONIZATION CHAMBER

A thimble chamber is basically a ionization chamber with small volume of air. Air equivalent material is used and smaller in size than free air ionization chamber. The air equivalent wall ensures that the energy spectrum of the electrons liberated in the thimble wall is similar to that of air. Hence, the outer electrode is made up of air equivalent wall material. These materials have effective atomic number equal to air, but their density is high.

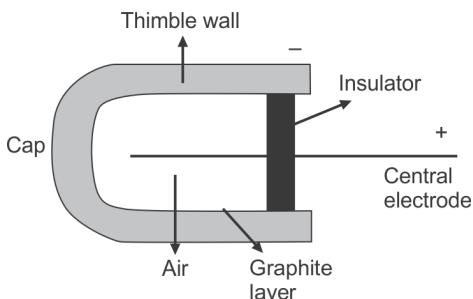
Generally, *bakelite* or *plastic* are used as wall material. This will facilitates the chamber to act like a chamber having a bigger volume of air. The inner surface of the wall is coated by a conducting material like *graphite*. The graphite acts as an outer electrode and participates in the charge collection. The central electrode is made up of thin aluminum and it is made positive with respect to graphite which is earthed. The central electrode is held by an insulating seal which also closes the chamber tightly. A suitable voltage is applied to collect ions in the air cavity (Fig. 6.8).

When radiation passes through the chamber, ion pairs are produced in the air cavity as well as walls of the chamber, under electronic equilibrium. To achieve electronic equilibrium, the wall thickness must be equal or greater than the maximum range of electrons. These ion pairs are collected by the electrodes and it is measured in terms

of ionization charge ( $Q$ ). By knowing the volume ( $v$ ) of air inside the cavity, one can calculate the charge per unit mass. Then, the exposure ( $E$ ) is given by

$$E = Q/(\rho \times v)$$

where,  $\rho$  is the density of air.



**FIG. 6.8:** Thimble ionization chamber

A thimble chamber is small in size (0.6 cc) and very much suitable for routine measurements in hospitals, for calibrating X-ray, tele-cobalt units and linear accelerators. They serve as secondary standard dosimeters and hence needs to be calibrated against the primary standard periodically (once in 3 years). They can be used either in dose mode or dose rate mode. In the dose mode, the charge is collected over a period of time and is stored in a capacitor. The total charge is measured and displayed on a liquid crystal display (LCD). In the dose rate mode, it measures the radiation exposure per unit time. The charge measured per unit time is displayed on the LCD. This mode is suitable for fluoroscopy measurements. The reading may be displayed in roentgens (R) or in  $\mu\text{Gy}$ .

These chambers are susceptible to error due to change in temperature, pressure and humidity and hence require suitable correction factors.

### Chamber Sensitivity

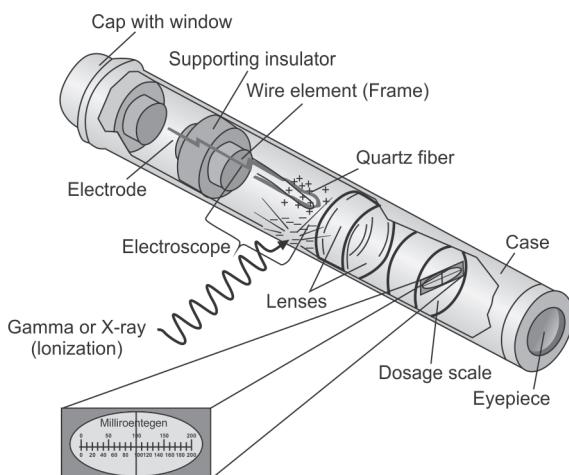
Consider a chamber which is fully charged with the help of an electrometer. Let the potential of the central electrode is  $V_1$ . Now, the chamber is exposed to radiation and ions are produced in the air cavity. During the exposure, ionization causes charge to leak off the central electrode. As a result, the potential of the central electrode reduces to  $V_2$  and the voltage drop is  $(V_1 - V_2)$ . Then, the charge ( $Q$ )

liberated by the radiation is  $= C(V_1 - V_2)$ , where  $C$  is the total capacitance of the system. This is a measure of radiation exposure.

The chamber sensitivity is defined as the voltage drop per roentgen. It can be shown that the voltage drop per roentgen ( $V/R$ ) =  $3.33 \times 10^{-4} v/C$ , where  $v$  is the chamber volume and  $C$  is the total capacitance. Thus, the chamber sensitivity is directly proportional to the chamber volume and inversely to the chamber capacitance. Condenser chambers are widely used for calibration of X-ray generators. One of the most popular was the Victoreen R-meter.

### POCKET DOSIMETER

Pocket dosimeter is an ion chamber with a quartz fiber suspended with in an air filled chamber, on a wire frame as shown in Figure 6.9. It has a built-in capacitance which can be charged by an external potential (charger). The positive charge is placed on the wire frame, by means of the charger. The quartz fiber is bent away from the frame due to columbic repulsion. This can be visible through an optical lens system upon which an exposure scale is superimposed.



**FIG. 6.9:** Pocket dosimeter

These dosimeters should be fully charged prior to their use, so that the initial reading of the dosimeter is set at zero. When exposed to radiation, ion pairs are produced in the air. These ion pairs partially neutralize the positive charge, reducing the columbic repulsion and allowing the fiber to move. Hence, the quartz fiber moves closer to

the wire frame that can be seen as down range excursion of the hair line fiber on the exposure scale (graticule). The movement of the quartz fiber is proportional to the radiation exposure, which is measured in Roentgen (R). The Roentgen is the unit of exposure =  $2.58 \times 10^{-4}$  C/kg. The dose in air can be calculated from the exposure, where 1R exposure is equal to 8.76 mGy (0.876 rad) of air dose.

The dosimeter is available in different ranges varying from 0–200 mR, 0–500 mR, 0–5 R, 0–20 R, 0–200 R, and 0–600 R, for measurement of X and gamma rays. It can detect photon energies from 20 keV-2 MeV. These dosimeters are available both in analog and digital types. Digital dosimeters use either GM tubes or diodes and solid state electronics. The dose measurement range of digital pocket dosimeter is 10  $\mu$ Sv to 100 mSv.

For personnel monitoring, smallest range (0–200 mR) should be employed. The main advantage of pocket dosimeter lies in its ability to provide instant on the spot check of radiation dose received by the personnel. Film and TLD will not show accumulated exposure immediately. In addition to the regular film badges, the radiation doses received by the radiation worker can be assessed by wearing a pocket dosimeter, which gives instantaneous radiation exposure. This is very useful in non-routine work, in which the radiation levels vary considerably and may be quite hazardous, e.g. cardiac catheterization laboratory. Suitable protective measures can be undertaken immediately to minimize future exposures. The dose can be read off directly by the person during or after any radiation work.

The accuracy of the pocket dosimeter is about  $\pm 10\%$ . Pocket dosimeters are small in size and easy to use and do not provide permanent record. Sudden mechanical shock may result in wrong reading. Hence, these dosimeters should be handled with care so as to indicate reliable reading of the doses received.

Nowadays, digital pocket dosimeters are available with easy display of instant radiation measurements. Presently, semiconductor diode based pocket dosimeters with digital display are also available. They have good energy and polar response, with reliable readings, matching to TLD badges. They make loud bleep sounds for every 15–30 minutes on background. The sound become more frequent as dose rate increases, and becomes continuous sound at high radiation fields. The energy range of these dosimeters is 45 keV to 6 MeV, and is available in mR and  $\mu$ Sv display.

## DOSE-AREA PRODUCT METER

Ionization chambers can be used to measure radiation intensity and field area simultaneously, while the beam is ON. It is basically a flat radiolucent air chamber, fitted over the collimator of the X-ray unit. Since air is used as medium, the attenuation is very little. As it measures air dose and radiation field area, it gives the dose area product (DAP) and hence the name:

$$\text{Dose-area product} = \text{Dose} \times \text{area} \text{ (cGy.cm}^2\text{)}$$

The dose depends upon kV, mAs, HT waveform and filtration. Usually, technologist or radiologist decides field area for each patient. It indicates that how much patient area is exposed with a given radiation dose. So that assessment of radiation hazards and associated biological effect is made easy. It is employed in fluoroscopic (angiogram) or cardiac catheterization laboratory examinations, where the procedure is long and radiation levels vary considerably. It is sensitive to exposure rate, fluoroscopy time and area of the beam used. It is also useful in pediatric imaging, to assess patient dose.

However, the DAP meter will not record the source to patient distance, magnification mode, changes in the kVp, etc. which also influences the patient dose. DAP meter is also called as Roentgen-area product (RAP) meter.

## AREA MONITORING

The assessment of radiation levels at different locations in the vicinity of radiation installation is known as area monitoring or radiation survey. These measurements will give an idea about the radiation status of the installation. On the basis of measurements taken, one could confirm the adequacy or inadequacy of the existing radiation protection status. In case, if the radiation levels are found to be higher than the permissible levels, suitable remedial measures can be taken. Hence, the objective of the above measurement is to ensure radiation safety and minimize personnel exposure. An ideal monitor should have uniform response to X and gamma radiation over the range of 15 keV to 3 meV. It should cover a wide range of exposure rates from 0.25 mR/h to a few 10 R/h. It should be able to assess beta radiation levels and be operable with battery cells.

Instruments used for the above purposes are called radiation survey meters and area monitors. In general, any survey meter/area monitor

should consist of two main parts, namely, (i) a device which detect the radiation, and (ii) a display system to measure the radiation. These instruments differ from each other in the medium in which the response takes place and in the method by which the response is detected and quantified. Following are the different type of meters generally used for radiation survey and area monitoring:

- i. Ionization type (air)
- ii. Geigher-Muller (GM) type (Neon and halogen) and
- iii. Scintillation detector type [NaI(Tl), ZnS(Ag)].

Selection of a particular detector depends on variety of factors like type of radiation to be detected and quantity to be measured, response of detector for the energy and type of radiation, etc. They can be used as portable radiation survey meters, capable of measuring radiation count rate in mR/h or  $\mu$ R/h. They are available in the form of vehicle mounted radiation meters, zone monitors, and door way mounted meters, etc.

### **IONIZATION CHAMBER SURVEY METER**

Ionization chambers for low level X-ray monitoring (exposure/exposure rate) are fabricated out of air-equivalent materials (bakelite, tufnol) and they can be used over a wide range of energies from 7 keV to 2 MeV. A typical survey meter consists of a 500 cc chamber connected to a battery operated electrometer and can measure exposure rates from few mR/h to about 10 R/h. Some of these are provided with an end window of thin mylar film for beta radiation detection.

Ion chambers for radiotherapy are fabricated with phenolic wall material with 200–350 cc chamber volume and operated both in dose and dose rate mode. It is recommended to use pressurized ion chambers (8 atmospheres or 125 psi) for radiotherapy. They provide enhanced sensitivity and improved energy response for the measurement of dose and dose rate. They allow fast response time to radiation leakage, scatter beams and pinholes. In addition, the low noise chamber bias supply provides for fast background settling time. It is capable of measuring gamma energy  $> 25$  keV, and beta energy  $> 1$  MeV.

Ionization chambers are used whenever accurate measurements are required. They approximate the condition under which the roentgen is defined. Ion chambers are used to measure X-ray machine outputs,

estimate radiation levels in brachytherapy, and in monitoring radionuclide therapy patients, and survey the radioactive material packages. Ion chambers are influenced by changes in temperature, pressure, photon energy and exposure rate. These limitations are less important in medical applications (5% loss of exposure rate at 10 R/h).

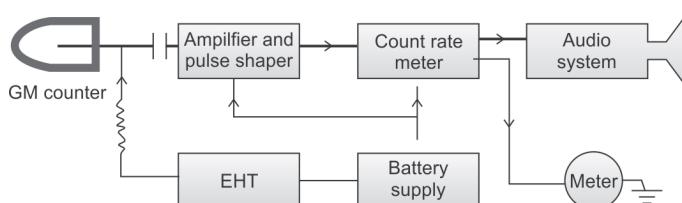
Ion chambers are capable of monitoring higher radiation exposure rate levels, and available in different ranges: 0–5 mR/h, 0–50 mR/h, 0–500 mR/h, 0–5 R/h, and 0–50 R/h. They respond slowly (8–2 seconds) to rapidly changing exposure rates and hence need warm up and stabilization before measurements are made.

Nowadays, survey meters are provided with a lot of special features like autoranging and autozeroing, optional beta slide, simultaneous measurement of dose and dose rate, operated by two 9 volts alkaline batteries, check source, communications interface with windows based excel add-in for data logging, programmable flashing LCD display and audible alarm with dose equivalent energy response (SI units).

### **GM TYPE SURVEY METERS**

GM type instruments are very sensitive and useful for monitoring of low level radiation. Since electronic amplification is not necessary, the electronic circuit of GM is very simple, compared to that of ionization chamber (Fig. 6.10). This feature makes the GM type instruments rugged and less costly. GM counters used for radiation monitoring generally use a mixture of gases (argon, neon and chlorine/bromine). It detects the presence and provides a semi-quantitative estimate of the radiation field magnitude. It provides measurements in counts per minute (cpm). It also provides an approximate measurement of mR/h, since it does not reproduce the conditions under which exposure is defined. But, the relationship between cpm and mR/h is a complicated function of photon energy.

GM counters for X-rays and gamma rays monitoring use copper or chromium cathodes for better efficiency. The primary photons interact



**FIG. 6.10:** Block diagram of GM survey meter

with the cathode materials to produce secondary electrons. Since GM meters are pulsed in nature, they should be used only in X-ray units, that emits continuous X-rays. They should not be used in X-ray units, that emits pulsed X-rays (e.g. linear accelerators).

GM type meters are mainly used as radioactive contamination monitor with thin window ( $1.5\text{--}2 \text{ mg/cm}^2$ ), and large surface area. It will respond to alpha ( $> 3 \text{ MeV}$ ), beta ( $> 45 \text{ keV}$ ), X and gamma ray ( $> 6 \text{ keV}$ ) radiations. GM detector is sensitive to particle radiation, but relatively insensitive to gamma radiations. It is suitable to measure natural background radiations, which are  $50\text{--}100 \text{ cpm}$ . It is mainly used in nuclear medicine for low level contamination surveys. GM counters have long dead time (100 msec) and result in 20% loss at 100,000 cpm measurements. They should not be used in high level radiation fields or when accurate exposure rates are required.

Though GM survey meters often display with  $\text{mR/h}$ , it is not a true exposure rate, but a good approximation. Hence, ionization chamber survey meters are preferred for accurate radiation survey. Geiger counters are available in different forms, namely, flat, thin window counters (pancake) and are suitable for radioactive contamination and low level radiation survey. It has peak sensitivity at the diagnostic energy range and hence, can be used for leakage measurements.

## **PERSONNEL MONITORING SYSTEMS**

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The aim of personnel monitoring program is stated as follows: (i) Monitor and control individual doses regularly in order to ensure compliance with the stipulated dose limits, (ii) Report and investigate over exposures and recommend necessary remedial measures urgently, (iii) Maintain lifetime cumulative dose records of the users of the service. Hence, the radiation received by all the radiation workers during their work should be regularly monitored and a complete up-to-date record of these doses should be maintained. Personnel monitoring is usually done by employing film badges or TLD badges or pocket dosimeter.

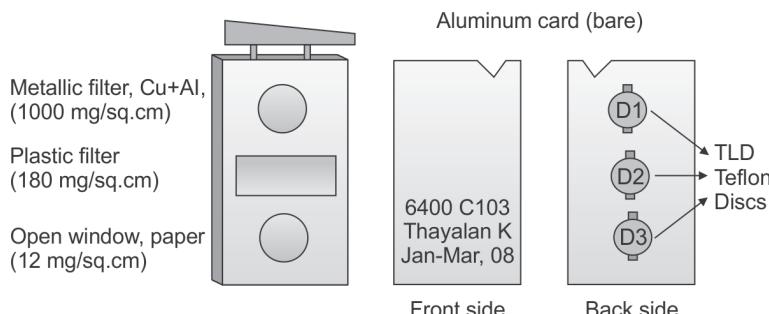
The personnel monitoring devices provide (i) occupational absorbed dose information, (ii) assurance that dose limits are not exceeded, and (iii) trends in exposure to serve as check on working practice. In India, country wide personnel monitoring service is offered by private *agencies*, accredited by Bhabha Atomic Research Centre (BARC), Mumbai.

## **THERMOLUMINESCENT DOSIMETER**

Film badge or thermoluminecent dosimeters (TLD) are used as personnel monitoring devices. The film badge has some disadvantages such as fading at high temperatures and humidity, high sensitivity to light, pressure and chemicals, complex darkroom procedure and limited self-life, etc. Hence, TLD badges are used currently in India, instead of film badges. It is based on the phenomenon of thermoluminescence, the emission of light when certain materials are heated after radiation exposure. It is used to measure individual doses from X, beta and gamma radiations. It gives very reliable results since no fading is observed under extreme climatic conditions. The typical TLD badge consists of a plastic cassette in which a nickel coated aluminum (Al) card is placed as shown in Figure 6.11.

*TLD card:* Three  $\text{CaSO}_4$ : Dy-teflon disks of the TLD card consists of 0.8 mm thick and 13.2 mm diameter each, which are mechanically clipped over three symmetrical circular holes, each of diameter 12 mm, on a nickel plated aluminum plate (52.5 mm  $\times$  29.9 mm  $\times$  1 mm). An asymmetric V cut provided at one end of the card ensures a fixed orientation of the card in the TLD cassette. The card is enclosed by a paper wrapper in which user's personnel data and period of use is written. The thickness of the wrapper (12 mg/cm<sup>2</sup>) makes the measurements equivalent to 10 mm depth below the skin surface. To protect the TLD discs from mishandling, the card along with its wrapper is sealed in a thin plastic (polythene) pouch. The pouch also protects the card from radioactive contamination while working with open sources.

*TLD cassette:* TLD cassette is made of high impact plastic. There are three filters in the cassette corresponding to each disk, namely,



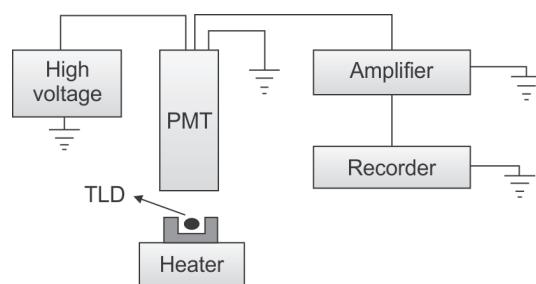
**FIG. 6.11:** TLD badge Al cards and its holder with filters

Cu + Al, Perspex and open. When the TLD card is inserted properly in the cassette, the first disk (D1) is sandwiched between a pair of filter combination of 1 mm Al and 0.9 mm Cu (thick:1000 mg/cm<sup>2</sup>). The copper filter is nearer to the TLD disk and the Al should face the radiation. The second disc (D2) is sandwiched between a pair of 1.5 mm thick plastic filters (180 mg/cm<sup>2</sup>). The third disk (D3) is positioned under a circular open window. A clip attachment affixes the badge to the users clothing or to the wrist.

The metallic filter is meant for gamma radiation, and the Perspex is for beta radiation. The filters are mainly used to make the TLD discs energy independent. When the TLD disk is exposed to radiation, the electrons in the crystal lattice are excited and move from the valence band to conduction band. There they form a trap just below the conduction band. The number of electrons in the trap is proportional to the radiation exposure and thus it stores the absorbed radiation energy in the crystal lattice.

After radiation exposure, the dose measurements are made by using a TLD reader (Fig. 6.12). The reader has heater, photomultiplier tube (PMT), amplifier, and a recorder. The TLD disk is placed in the heater cup or planchet, where it is heated for a reproducible heating cycle. While heating, the electron returns to their ground state with emission of light. This emitted light is measured by the PMT, which converts light into an electrical current (signal). The PMT signal is then amplified and measured by a recorder. The reader is calibrated in terms of mR or mSv, so that one can get direct dose estimation.

Nowadays, windows based computer controlled TLD readers are available. They are capable of analyzing TLD chips, ribbons, powder, discs, pellets, rods and microcubes. They display digital glow curve and temperature profile. They can handle one or more planchets at a time either with manual drawer or computer controlled drawer function. Programmable annealing oven is also available along with the system. In India, Ca SO<sub>4</sub>:Dy-teflon disks are used in country wide personnel monitoring, with an accuracy of  $\pm$  10%.



**FIG. 6.12:** Thermoluminescent dosimeter reader

The disks are reusable after proper annealing, up to 300 times. The annealing process release the residual energy stored from the earlier exposure. A typical annealing cycle consists of 400°C for 1 hour, followed by 300°C for 3 hours. This badge can cover a wide range of dose from 10 mR to 10,000 R with a accuracy of  $\pm$  10%.

TLD badges do not provide a permanent record and it is available for extremity dosimetry and finger dosimetry (ring). LiF can also be used as TLD phosphor, which has wide dose response, 10 mSv to 1000 Sv. Its effective atomic number is close to that of tissue with an accuracy of  $\pm$  2%.

TLD badges are normally worn at the chest level, that is expected to receive the maximum radiation exposure. Most of the radiation workers used to wear the badge at the waist level which is not correct. During fluoroscopy, it is preferable for the radiologist to wear at the collar level inside the lead apron to measure the dose to the thyroid and lens of the eye, since most of the body is shielded from the radiation exposure. Pregnant radiation workers should wear a second badge at waist level (under the lead apron) to assess the fetal dose. Additional wrist badge is advised for procedures involving nuclear medicine, brachytherapy source handling and interventional radiology.

### **GUIDELINES FOR USING TLD BADGE**

1. TLD badges are to be used only by persons directly working in radiation. Administrators, darkroom assistant, sweepers, etc. need not be provided with TLD badges.
2. TLD badge is used to measure the radiation dose. It does not protect the user from the radiation.
3. The name, personnel number, type of radiation (X or gamma), period of use, location on the body (chest or wrist), etc., should be written legibly in block letters on the front side of the badge.
4. A TLD badge once issued to a person should not be used by any other person.
5. Each institution must keep one TLD card loaded in a chest TLD holder as control, which is required for correct dose evaluation. It should be stored in a radiation free area, where there is no likelihood of any radiation exposure.
6. TLD badge should be worn compulsorily at the chest level.

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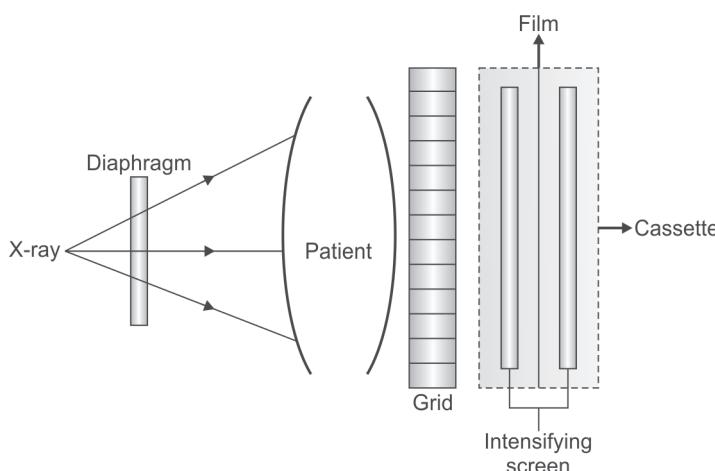
It represents the whole body dose equivalent. If lead apron is used, TLD badge should be worn under the lead apron.

7. While leaving the premises of the institute, workers should deposit their badges in the place where control TLD is kept.
8. A badge without filter or damaged filter should not be used. It is replaced by a new holder.
9. Every radiation worker must ensure that the badge is not left in the radiation field or near hot plates, ovens, furnaces, burners, etc.
10. Every new radiation worker has to fill up the personnel data form, and should be to the BARC accredited agency.
11. All the used or unused TLD badges should be return, after every service period (quarterly) in one lot so as to reach 10th of next month.
12. Contact for all correspondence regarding TLD badge service, to the Officer-in-charge, Personnel Dosimetry and Dose record section, Radiological physics and Advisory division, Bhabha Atomic Research Centre, CT and CRS Building, Anusakti Nagar, Mumbai 400094.

# 7

# Screen-Film Radiography

Radiography is a procedure in diagnostic radiology in which X-rays are used to produce a shadow picture of a patient. This will enable us to visualize the internal structures. It is a method of obtaining two dimensional image of patient's anatomy, by using X-ray film as detector (Fig. 7.1). It was the first imaging modality used in radiology. It is used in variety of diagnosis, starting from bone imaging to chest radiography. In a radiography technique, the radiation from the X-ray tube is transmitted through the patient's body, and then reaches the film. After processing the film, the radiograph is obtained. A radiograph is a negative image. The production of good radiological image require number of accessories such as (i) grid, (ii) cassette, (iii) intensifying screen, and (iv) X-ray film, in addition to the X-ray unit. The principle of the above accessories will be discussed in the following topics.



**FIG. 7.1:** Radiography

This chapter will enlighten the various components that are used to produce a good shadow picture with minimum radiation dose to the patient.

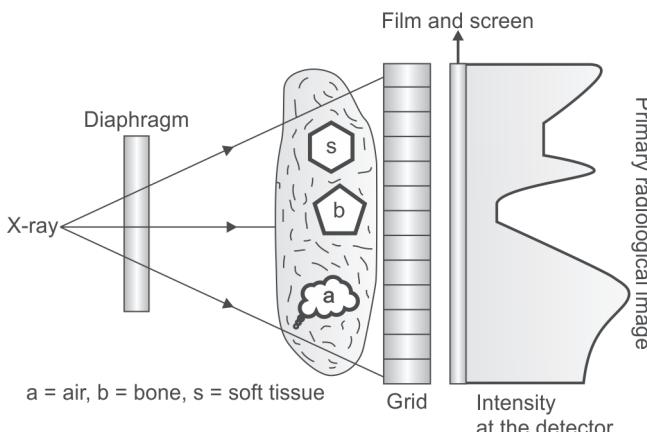
## PRIMARY RADIOLOGICAL IMAGE

Human body is heterogeneous. It is mainly made up of air, fat, water, soft tissue and bone of differing density and atomic number (Table 7.1). When X-rays passes through the body, it gets attenuated differentially by different tissues, and results in a variation of transmitted radiation (Fig. 7.2). This variation is referred as primary radiological image. Since eye is insensitive to X-rays, this image is converted into visible image either by using a X-ray film or fluorescent screen.

The beam that emerges from the patient contains primary and scattered radiation. Only the primary beam contains the useful

**TABLE 7.1** Physical characteristics of air, fat, water, soft tissue, and bone

	Effective At. No. (Z)	Density g/cm <sup>3</sup>	Electrons per gram
Air	7.64	0.00129	$3.01 \times 10^{23}$
Fat	5.92	0.91	$3.48 \times 10^{23}$
Water	7.42	1.00	$3.34 \times 10^{23}$
Soft tissue	7.4	1.00	$3.36 \times 10^{23}$
Bone	13.8	1.85	$3.00 \times 10^{23}$



**FIG. 7.2:** Primary radiological image

information of the patient. Hence, the scattered radiation must be removed, before reaching the film. In the diagnostic range, the X-ray interact mostly by photoelectric effect, which is proportional to  $Z^3$ . The Compton interaction is minimum and mostly with low Z materials. Overall, photoelectric effect dominates over Compton scattering at this energy level. Hence, bone, soft tissue and fat will offer differential attenuation to X-rays. As a result, the transmitted radiation will also have variation. Thus, bone, soft tissue, and fat can be distinguished from one another.

## CONTRAST AGENTS

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Contrast is the difference in X-ray intensities transmitted through different parts of the patient. It is usually called as subject contrast, and is affected by incident photon energy, atomic number and density of the organ. Photoelectric effect contributes to subject contrast at low photon energies. It is also important when materials of high Z (calcium, iodine, barium) are present.

Contrast agents are used to improve subject contrast. They are necessary to visualize many organs in the body with great detail. The contrast agent absorbs X-rays either more or less than the surrounding tissues. The type of contrast agents used are (i) air (ii) iodine compounds, and (iii) barium compounds. Air has negligible density compared to tissues and absorbs very little X-rays. Hence, air cavities can be easily distinguished from other body tissues. Organs filled with barium or iodine absorbs X-rays very strongly and transmits very little radiation, thereby improving subject contrast.

Barium is administered as a contrast agent, because of its high Z(56) and physical density. The barium K-edge is 37 keV, which matches the photon energies used in fluoroscopy. Iodine ( $Z = 53$ , K-edge = 33 keV) is also an excellent contrast agent. It can be injected intravenously, to improve subject contrast.

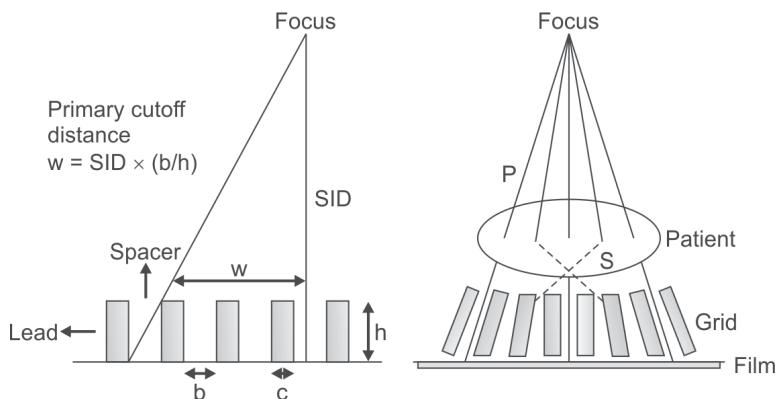
## GRIDS

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When a beam of X-ray passes through the patient, the beam is absorbed and scattered. The absorbed primary beam gives a useful shadow while the scattered radiation will tend to spoil the shadow. Scattered radiation contributes a constant background fog to the film image. This will increase the noise in the image. The ratio between

the amounts of scattered radiation energy to the amount of primary radiation energy at a point is called as scatter to primary ratio (SPR). The SPR increases with thicker patient and larger field sizes. For example, in a abdomen radiography, only 20% of the photons contributes to the image formation and the other 80% energy goes as scattered radiation. Hence, scattered radiation must be removed, in order to increase the image contrast.

The scattered radiation can be removed by a grid, placed in between the film and the patient. The grid consists of a series of parallel lead or tantalum strips of thickness ' $c$ ' (50 mm) and of height ' $h$ ' separated by spacers of low attenuating material of width ' $b$ ' (350 mm) as shown in Figure 7.3. Aluminum or plastic fibers are used as low attenuating spacers. The grid is positioned between the patient and the detector, so that its long axis is pointed towards the X-ray beam. The primary X-rays coming out of the patient, passes through the inter space, since it is parallel in direction. The scattered X-rays, which are in non parallel direction, strike the grid bars and get absorbed. The ratio of the primary transmission to the scatter transmission of a grid is called the *selectivity*.



**FIG. 7.3: Grid design and principle**

## GRID RATIO

The ability of the grid to discriminate against scattered radiation is measured by the grid ratio, which is defined as the ratio of the height ( $h$ ) to the width of the spacer ( $b$ ) between the lead strips.

$$\text{Grid ratio} = h/b$$

As the grid ratio increases, the grid removes more scatter radiations. The strip line density is  $1/(b + c)$  lines per unit length. Typical grid

ratio ranges from 4:1 to 16:1 and strip line densities are 25–60 lines per cm.

The performance of a grid can be understood by contrast improvement factor. It is ratio between image contrast with grid and image contrast without grid at 100 kVp. Higher grid ratio provides higher contrast improvement factor. However, it increases patient dose, as it employs higher exposure techniques. Bucky factor (Gustave Bucky, 1913) is another parameter which relates the patient dose. It is the ratio between the patient dose with grid and patient dose with out grid. Bucky factor increases with increase of kVp and grid ratio.

### **TYPES OF GRID**

Grids may be classified as (i) parallel grid, (ii) crossed grid, (iii) focused grid, and (iv) moving grid (Potter-Bucky). In a parallel grid, the lead strips are parallel to each other in their longitudinal axis. Most of the X-ray tables are provided with linear grids. It is easy to design, but has the property of grid cutoff. This means that the attenuation of primary radiation is greater at the edges and it can be partial or complete cutoff. The distance of grid cutoff may be estimated from the ratio of source to image distance (SID) to grid ratio.

Crossed grid is made up of lead strips that are parallel to the long axis and short axis of the grid. Usually, it is designed with two parallel grids, that are perpendicular each other. The grid ratio of crossed grids is equal to the sum of the ratios of the two parallel grids. Crossed grids are efficient in removing scatter radiations and has higher contrast improvement factor and high grid ratio. It is useful at high kVp and tilt-table exposure techniques. The disadvantage includes difficulty in positioning, proper alignment of tube and table, and higher patient dose. Crossed grid also suffers from grid cutoff.

Focused grid is made mainly to reduce grid cutoff. The lead strip lies in a imaginary radial lines of a circle, whose centre is the focal spot. The strips are parallel to the divergence of the X-ray beam. The grids are marked with focal distance and the side facing the target. If it is reversed, grid cutoff may occur, hence enough care is needed to position focused grid.

### **MOVING GRID**

When a focused or parallel grid is used, each lead strip will appear on the radiograph as very fine line. These lines may spoil the information

in the film. However, these lines may be removed by moving the grid during the radiographic exposure. This is the principle of Potter-Bucky grid (Hollis E. Potter, 1920). Generally, focused grids are used as moving grids. The grid may be made to move continuously in one direction. The grid motion is timed by the exposure control of the X-ray machine. It starts moving just before the X-rays are turned on and continues to move even after the exposure is off. The traveling period of the grid should be greater than the exposure time.

There are two types of moving grids, namely, (i) reciprocating grid and (ii) oscillating grid. The reciprocating grid is driven by a motor and the grid moves back and forth several times, during exposure. The distance traveled may be 2 cm. In the oscillating type, the grid is kept in a frame, which has 2–3 cm clearance on all sides. An electromagnet pulls and releases the grid before the exposure. The grid oscillates in circular path about the frame and comes to rest after 20–30 seconds.

Moving grids increases the distance between patient and film, resulting in magnification, cassette motion and image blur. However, the motion blur is undetectable and hence used widely.

The use of grid will always increase the exposure, because it will absorb some of the primary radiation. In order to reduce the exposures, grids with smaller ratios are preferred. Low ratio grids such as 8:1 is used with energies up to 90 kV<sub>p</sub>. High-ratio grids such as 12:1 are preferred for high energy radiation. In mammography, grid ratio of 4:1 or 5:1 is used. These grids produce films with better contrast with increased patient dose. Grids are generally used for body parts > 12 cm thick or techniques > 70 kV<sub>p</sub>. Grid can produce artifacts when improperly aligned.

## AIR-GAP TECHNIQUES

Air-gap technique is an alternative method of eliminating scatter radiation with large radiographic fields. When X-ray beam passes through the patient, it gets scattered in all directions. The intensity of scatter radiation is maximum at the patient's surface and decreases rapidly at increasing distance from the surface. If sufficient gap is allowed between the patient and film, the scattered photons will not reach the film.

Higher focus to film distance is used with air gap techniques, in order to maintain image sharpness. Hence, it requires greater exposure factors compared to grid. But, the patient exposures are generally

low. Air-gap technique also introduce magnification. Air-gap technique is most effective, when the point of scatter exist closer to the film. Air-gap techniques are used in neuroradiography and mammography.

## CASSETTE

A cassette is a light proof rigid holder that contains screens and film (Fig. 7.4). Cassettes are usually hinged (latches) on one side and can be opened from the other side. The side which is having the latches is the back side and other side facing the patient is called front side. The front side is made of material of low atomic number, like plastic or carbon fiber. This is to maximize the transmission with low attenuation. Carbon fiber ( $Z = 6$ ) absorbs only 50% of X-rays compared to aluminum. Cassette with carbon fiber can be operated with low radiographic techniques, resulting in lesser patient dose. The back side is usually made of heavy metals (lead), having high atomic number, to minimize back scatter.

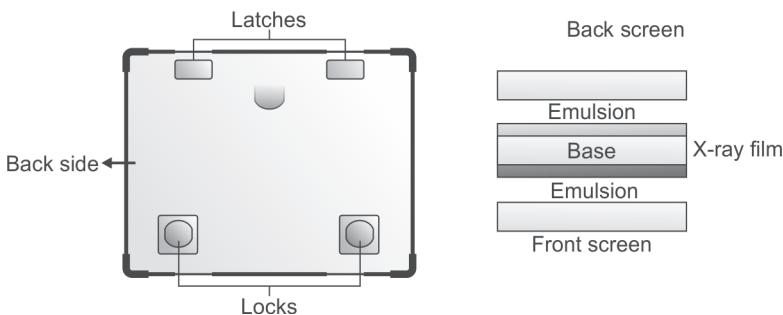


FIG. 7.4: Cassette

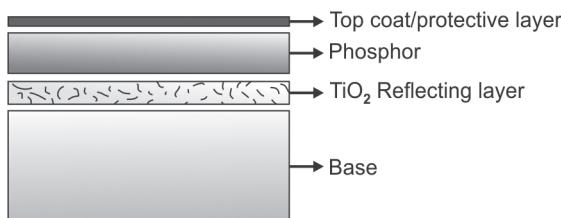
There is a small window at the back side, to provide patient ID. The patient's information can be given in the form of flash card; later on, it is exposed to ID camera. The ID camera optically exposes the window and record the image of flash card. Inside the cassette, there are two permanently mounted intensifying screens, called front and back screens. The X-ray film is loaded between the two screens. The screens may have different thickness or equal thickness.

Compressive materials, such as radiolucent plastic foam, is kept in between the back screen and the cassette cover. The compressive material maintains good screen-film contact, when the film is loaded. Good physical contact between film and screen is necessary, to avoid artifacts and to have good image quality. X-rays, by passing the back

screen get back scattered and reaches the film, which result in image fog. The film loading and unloading into the cassette is done in the darkroom. Cassettes are available in different sizes.

## INTENSIFYING SCREEN

Film is insensitive to X-rays and requires more amount of X-rays to produce an image, which will increase patient dose. To avoid this, intensifying screens are used in cassettes in medical imaging. They absorb X-ray photons and emit more visible light or ultraviolet, for which the X-ray film is more sensitive. The light or UV exposes the film and gives the final image, which will improve the efficiency of radiographic imaging, with lesser patient dose. Thus, the intensifying screens amplify the effect of image formation. Generally, intensifying screen consists of four layers, namely, (i) base, (ii) reflecting layer, (iii) phosphor, and (iv) protective coating (Fig. 7.5).



**FIG. 7.5:** Intensifying screen

## LAYERS OF SCREEN

### Base

The base is made of polyester with 1 mm thick. The base serves as a mechanical support on which the reflecting layer, phosphor and protective layers are mounted. The base material should be moisture free, resistance to radiation damage and discoloration, chemically inert, and flexible.

### Reflecting Layer

The reflecting layer is made of a white substance, such as titanium dioxide ( $Ti O_2$ ) or magnesium oxide. It is a shiny material of thickness 25  $\mu m$ , which reflects light towards the phosphor and make the light emission isotropic. Thus, the reflective layer increases the efficiency of the intensifying screen, by doubling the number of light photons.

## PHOSPHOR

The phosphor is a crystal of inorganic salts, which emits light when exposed to X-rays. The thickness of the phosphor ranges from 50–300  $\mu\text{m}$  with individual crystal size of 5–15  $\mu\text{m}$ . The commonly used phosphors are calcium tungstate ( $\text{CaWO}_4$ ), zinc cadmium sulfide, cesium iodide and barium strontium sulfate. The phosphor should have high atomic number, high conversion efficiency, spectral matching and lesser after glow. It should not be affected by heat, humidity and environmental factors.

In recent times, rare earth phosphors ( $Z = 57\text{--}71$ ) such as Gadolinium oxysulfide ( $\text{Gd}_2\text{O}_2\text{S:Tb}$ ), lanthanum oxysulfide ( $\text{La}_2\text{O}_2\text{S:Tb}$ ) lanthanum oxybromide ( $\text{LaOBr}$ ) and yttrium tantalate ( $\text{YTaO}_4$ ) are used as screen phosphors. They can be manufactured with speed range of 200–1200. They have suitable K-shell absorption edges over the diagnostic photon energy of 35–70 keV. Their absorption and conversion efficiency is high. The spectral emission of rare earth phosphors is discrete and is centered at 540 nm. Therefore, green sensitive film must be used with these phosphors. These phosphors reduce patient dose, due to lower radiographic techniques. They also have less thermal stress and require lesser room shielding, as radiation levels are low. Though cesium iodide ( $\text{CsI}$ ) is used in fluoroscopy and digital radiography, it is moisture sensitive and fragile. Hence, it is not used in screen-film radiography.

## PROTECTIVE COATING

The protective coating (10–20  $\mu\text{m}$ ) is transparent to light and is facing the X-ray film. It is resistant to abrasion and damage caused by handling. It also prevents formation of static electricity, and provides a surface for cleaning. The total thickness of the intensifying screen is about 1.15–1.3 mm.

When X-ray passes through the front screen, the phosphor absorbs the X-rays and emits light in all directions. The reflecting layer reflects the light towards the film, so that no photon is lost. Some portion of the X-rays that are bypassing the X-ray film is absorbed and converted into light by the back screen. Thus, the intensifying screen converts large amount of X-ray photons (95%) into light photons of blue or green wavelength. Since the X-ray film is sensitive to blue or green light, it absorbs the entire light and gives the image.

## **SCREEN CHARACTERISTICS**

### **Quantum Detection Efficiency**

The quantum detection efficiency (QDE) or absorption efficiency of a screen is the ratio between the amount of X-rays absorbed and amount of X-rays incident. Though thicker screens give higher QDE, they suffer with lateral light diffusion, resulting in blurred images. Thicker screen also reduces spatial resolution. This is the reason why two thin screens are used in radiography, which reduces the light diffusion path without loss of spatial resolution. The QDE of  $\text{Gd}_2\text{O}_2\text{S}:\text{Tb}$  is highest for photon energy greater than 50 keV.

### **Conversion Efficiency**

The conversion efficiency of a phosphor is the ratio between the amount of light emitted and the X-ray absorbed. The overall efficiency of screen-film system is product of absorption efficiency and conversion efficiency. It depends upon the intrinsic conversion efficiency of the phosphor. It is 5% for calcium tungstate and 15% for  $\text{Gd}_2\text{O}_2\text{S}:\text{Tb}$  with green light emission of wavelength 545 nm. Presence of light absorbing dye reduces the conversion efficiency.

### **Speed**

The speed of a screen is inversely related to the exposure ( $1/R$ ) required to produce a given density. As the speed increases, the exposure required decreases. Screens are generally classified as fast, medium and slow speed screens. High speed screens (1200) are thicker and provide less spatial resolution with less patient exposure. Slow speed screens (100) are thinner, but have better spatial resolution. The rare earth screens are faster, because they have a higher absorption efficiency and higher conversion efficiency, at the mean X-ray energy used.

Patient exposure is decreased greatly when intensifying screens are used. The reduction in patient exposure is measured by a term called intensification factor. The intensification factor is the ratio of the X-ray exposure needed to produce a given density (optical density of 1) on a film without screen and with screen. It is a measure of speed of a screen. The usual intensification factors are 30–50. The  $\text{Gd}_2\text{O}_2\text{S}:\text{Tb}$  gives an intensification factor of 50, over the diagnostic X-ray range.

### **Noise**

Noise appears on the radiograph as speckled background. It occurs when fast screens and high kVp techniques are used. Screen with

higher conversion efficiency, increases noise. However, increase in absorption efficiency will not affect noise. Increase of conversion efficiency, enhances quantum mottle, resulting higher noise. This may occur in very fast screens, with grainy and mottled image. Rare earth screens are 2 times faster than calcium tungstate, which do not increase noise significantly.

### Spatial Resolution

Spatial resolution refers how an object can be imaged and it is expressed in line pairs (lp) per mm. If this number is higher, better the spatial resolution and smaller the object that can be imaged. Screens have lower spatial resolution compared to direct exposure film. The spatial resolution is 7 lp/mm for fast screens, 15 lp/mm for fine-detail screen, and 50 lp/mm for direct exposure film. Human eye can resolve 10 lp/mm. When the film is used with screens, the light interacts with film by larger area, which is the cause for reduction in spatial resolution. Smaller crystals and thinner phosphor layer improves spatial resolution.

### Handling of Screen

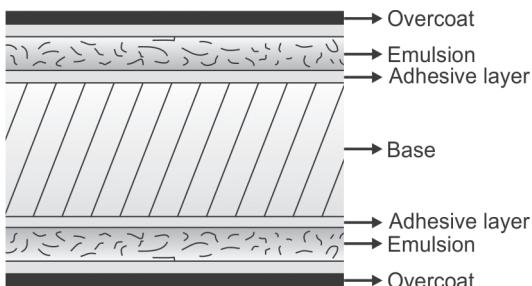
Screen must be handled with care. Any foreign material on the screen, such as paper, blood, scratches, hair, dust, and stains will block light photons and produce area of under exposure, leading to artifact and image degradation. The film should not be made to slide into the cassette, while loading. Its sharp edge may scratch the screen. The film should be removed by tilting the cassette, so that the film fell down on the technologist hand. Finger nails should not be used to take the film from the cassette. The cassette should not be kept open in the darkroom. The screen may be cleaned periodically (monthly) with a solution containing anti-static compounds or soap and water. It should be rinsed and dried after every cleaning. Film-screens should have good contact and this must be checked periodically with a help of a wire mesh.

## STRUCTURE OF X-RAY FILM

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The transmitted X-rays from the patient should be converted into visible image to the human eye for interpretation. The device that does the job is called image receptor. The image receptors are X-ray film, fluorescent screen and solid state device. The medical X-ray film is used for capturing, displaying and storing radiographic images. It consists of a (i) base, (ii) adhesive layer, (iii) emulsion, and (iv) overcoat.

The emulsion is coated on both sides and hence it is called double side emulsion film (Fig. 7.6).



**FIG. 7.6:** X-ray film and its composition

## COMPOSITION

### Base

The base gives a rigid support on which the emulsion is coated. It should be flexible, fracture resistant, and easy to handling without kinking. The base should have dimensional stability, so that it should not produce image distortion. It should have uniform lucency and transparent to light. It should be inert, so that the sensitometric properties of the emulsion are not affected.

Initially, glass and cellulose nitrate were used as bases. Later (1920), the base cellulose triacetate (CTA) and polyester (1960) are being introduced. The polyester is made from polyethylene terephthalate resin. Polyester base is resistant to warping from age, stronger with higher dimensional stability. It is thinner in size ( $175\text{ }\mu\text{m}$ ) and easy to transport in automatic film processor.

The X-ray film base is usually added with a dye, so that the film looks like blue. These films are called blue tinted, which reduces eye strain and fatigue. It gives pleasing appearance to the intermediate densities in the image and increases the diagnostic accuracy.

### Adhesive Layer

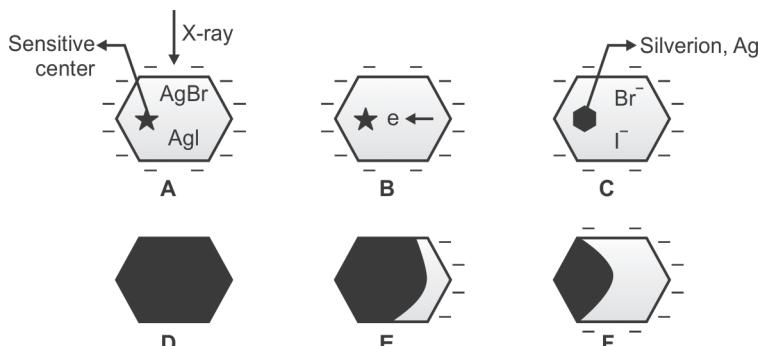
The adhesive layer lies in between the base and emulsion, in the form of thin coat. It uniformly binds the emulsion to the base. It also helps to maintain proper contact between emulsion and base and provide integrity during film processing.

### Emulsion

The emulsion ( $3\text{--}5\text{ }\mu\text{m}$  thick) is coated over the adhesive layer. Emulsion consists of gelatin and silver halide crystal in a uniform manner. Gelatin is transparent to light and porous for chemicals. It gives support for

silver halide, by holding them properly. Among the halides, silver bromide (98%) and silver iodide (2%) is used as crystal in the film. These halides are flat and have high atomic numbers, bromide ( $Z = 35$ ), silver ( $Z = 47$ ), and iodide ( $Z = 53$ ), compared to gelatin ( $Z = 7$ ). The halide crystals are available in tabular, cubic, octahedral, polyhedral or irregular grain shapes. Tabular grain shape (thickness  $0.1 \mu\text{m}$ ) is commonly used in radiography. The crystal formations are done in dark as follows; The metallic silver is dissolved in nitric acid, to form silver nitrate. This is mixed with potassium bromide, to form silver bromide. This is done in the presence of gelatin under given temperature and pressure conditions. The arrangement of atoms in the crystal is cubic and lattice structure that has imperfections. These imperfections provide sensitivity centers, for latent image formation. Direct exposure film has thicker crystals than screen type film. The film speed is controlled by size and concentration of the crystal.

When the film is exposed to X-rays, photon interacts with bromine ( $\text{photon} + \text{Br}^- = \text{Br} + \text{e}^-$ ) and release secondary electrons (Fig. 7.7). These interactions are either photoelectric or Compton type. These electrons migrate to the sensitivity centre and get trapped. Mobile silver atoms ( $\text{Ag}^+$ ) are attracted to the sensitivity centers, where they combine with electrons and become metallic silver ( $\text{Ag}^+ + \text{e}^- = \text{Ag}$ ). The metallic silver atoms give latent image, which is invisible. Basically, the bromine and iodine are present at the surface, whereas silver is inside the crystal. Mostly electrons are provided by bromine and iodine atoms, resulting in collapse of crystal structure. As a result,



**FIG. 7.7:** Latent image formation: (A) X-ray exposure provide electrons, (B) electrons moves to the sensitive center, (C) mobile silver atoms moves to the sensitivity center combine with electron and forms latent image, (D) process repeated, latent image widens, (E) additional silver formation during processing, and (F) final metallic silver image

bromine and iodine are free to move to the gelatin area. No more ionic force is acting in the crystal.

### **Overcoat**

The gelatin is covered by a layer called over coat. It protects the emulsion from scratches, pressure, contamination and handling damages.

## **TYPES OF FILM**

The X-ray film is classified as (i) screen type film, (ii) direct exposure or non-screen type, (iii) mammography film, (iv) laser film, and (v) specialty film. Films are available in variety of sizes. The most common sizes are 7 × 7 in, 8 × 10 in, 10 × 12 in, 14 × 14 in and 14 × 17 in.

### **Screen Type Film**

Screen-film selection depends on contrast, speed, spectral matching, crossover, and safe light. It is available with low and high contrast levels and multiple latitudes. High contrast film contains uniform and smaller size grains and produces black and white image. Whereas low contrast film have large grains with wide range of sizes and gives grey image. Films are available with different speed, and it is controlled by grain size and shape. Screen type film has double side emulsion with double the speed of single side emulsion film. In general, speed refers for a combination of film and two screens. The proper matching of film and screen is required for speed accuracy.

The light from a given screen may expose the base on the opposite emulsion, which is called cross over. Tabular grains reduce cross over significantly. Crossover can be minimized by (i) adding a light absorbing dye, and (ii) intensifying screens that emits shorter wavelengths (blue or UV).

Introduction of rare earth screens require proper spectral matching. The rare earth screens emit UV, blue, green and red light. The film is sensitive to blue and violet, not for green and red. Hence, the film is spectrally sensitized with special types of dyes. If green emitting screen is used, the film should be sensitive to blue and green light. This is called spectral matching and the film is called green sensitive film or orthochromatic film. If there is mismatching, it will reduce the speed and give higher patient dose.

Reciprocity law states that the total exposure is not proportional to time, but product of X-ray intensity and time (mAs). This is true

for direct exposure of film by X-rays. In the case of screen-film combination, the above law fails. Shorter exposure (angiography) and longer exposure times (mammography) has direct relation with optical density. This is called reciprocity law failure.

Safe light provide illumination in the darkroom, keeping the film unexposed. A 15 watt bulb at 1.5 m from the work bench is the correct choice. For a blue sensitive film, an amber light (550 nm) is used. This light will fog the green sensitive film. Therefore, a red filter (660 nm) should be used for green sensitive film. Red filter can be used for both blue and green sensitive film.

### **Direct Exposure Film**

The emulsion of a direct exposure film is thicker and consists of high concentration of AgBr crystals, with single side emulsion. They are mainly used to image thinner body parts, such as hands, feet, etc. It employs higher radiographic techniques and hence increased radiation dose to the patient. It is rarely used to day in medical imaging.

### **Mammography Film**

Mammography film is a single side emulsion type, always used with single intensifying screen. Currently, green emitting terbium-doped gadolinium oxysulfide screens with green sensitive film is used. The surface of the film base opposite the screen is coated with a light absorbing dye, to reduce reflection of light from the screen. The above coating is called antihalation coating and the effect is called halation. This coating is removed during processing, to improve image viewing.

### **Laser Film**

Laser films are used in computed tomography, magnetic resonance imaging and computed or digital radiography. Basically, digital electronic signal from the imaging system modulates a laser signal, proportional to the image signal. The laser beam writes the image on the film in raster fashion. The film is a silver halide film that has been sensitized to red light, which is emitted by the laser. Laser printers are light sensitive and hence require darkness for operation. Different types of lasers are used and offer consistent image quality. It is also available with multiple film sizes and multiple image formats per film.

### **Specialty Film**

Specialty film includes cine films used in angiography and spot films used in fluoroscopy imaging. Cine film is 35 mm and is supplied in

rolls of 100 and 500 feet. Spot films are 70–105 mm width, is used in cameras. They are similar to cine film, but larger in size, hence can be viewed directly. Processing is very critical in the above types of films and requires special processors. It has single sided emulsion and is provided with small notch in one corner of the film. Due to digitization of imaging equipments, the use of above films is declining today.

### **FILM HANDLING AND STORAGE**

X-ray film should be stored and handled properly, otherwise it will produce artifacts. It should not be bend, crease, or rough handled. It is pressure sensitive and sharp objects like finger nails may produce artifact. Film is sensitive to temperature, humidity and should be stored at 20°C. Higher temperature and humidity (60%) may cause fog and reduce image contrast.

Film is sensitive to light and should be stored and handled in dark. Exposure to low level light may increase the fog. Hence, a well sealed darkroom and a light proof storage bin is a must. Ionizing radiations may fog the film and reduce contrast. Film is more sensitive after an exposure than before. In the first exposure, the optical density is raised above the toe. Successive exposure may cause higher optical density. Film should not be stored near radioactive substance and nuclear medicine areas.

Films are supplied in boxes of 50 or 100 sheets. The packing may be by interleaved or non-interleaved method with chemically treated paper. Expiry date is given on the box, which is the self life of the film. Film should not be used after the expiry date, usually 6 months. Aged films may have loss of speed, contrast with increased amount of fog. Film should be stored vertically on the edges. In this, the film will not stick to one another, less likely to warp, and have less pressure artifacts. The storage is made in such a way that oldest film should be used first. Film may be purchased monthly, so that the storage may not exceed more than 30 days.

### **CHARACTERISTICS X-RAY FILM**

The characteristics of X-ray film can be discussed with the following parameters. They are density, characteristic curve, contrast, latitude and emulsion absorption.

## Density

The term density refers to the degree of blackening on the film. When the X-ray film is exposed to X-rays, the metallic silver gives the blackness on the film. That is why X-ray film is said to be a negative recorder. The degree of blackness is directly related to the intensity of radiation exposure. It can be quantified by a term optical density (OD), which is given by the relation:

$$OD = \log_{10} (I_0/I_t)$$

where,  $(I_0/I_t)$  is the inverse of transmittance (T), which is measured by a densitometer. If  $I_0$  is the light intensity measured without film and  $I_t$  is the light transmitted through the film, then  $T = I_t/I_0$ . The useful range of density in diagnostic radiology is 0.25–2.0.

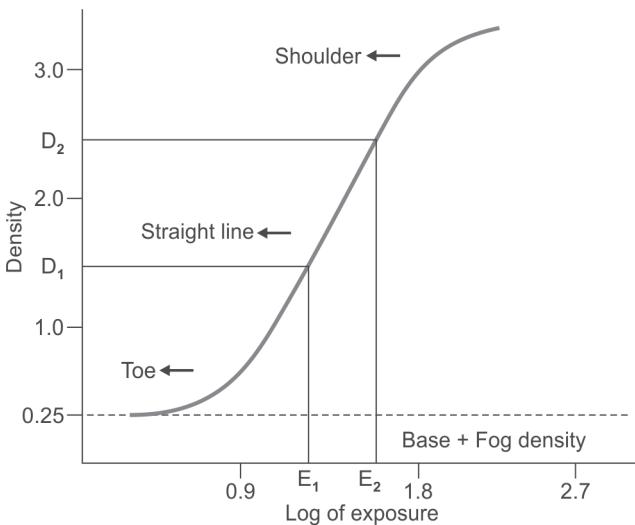
## Characteristic Curve

The relation between radiation exposure and optical density is plotted as a curve, known as the characteristic curve or H and D curve, named after Hurter and Driffield, who first generated these curves in 1890. The film density is plotted on the vertical axis and log of film exposure on the horizontal axis. The curve has sigmoid shape and has three portions, called toe, straight line and shoulder as shown in Figure 7.8. The toe is the low exposure region, and the shoulder is the high exposure region of the curve. Base plus fog level is the film blackening in the absence of any radiation exposure and typically ranges from 0.1 to 0.2 OD units. It refers the background fogging and the tinting (blue) of the base. The maximum film density ranges from 2.5 to 3.0 OD units.

All radiographic techniques should produce a density in the straight line portion. The contrast is related to the slope of the linear portion of the curve. Higher the slope, higher is the image contrast. The parameter which describes the contrast of the film is called average gradient. It is the slope of straight line, connecting two given points in the characteristic curve.

$$\text{Average gradient} = \frac{(D_2 - D_1)}{(\log_{10}E_2 - \log_{10}E_1)}$$

where,  $D_2$  and  $D_1$  are the optical densities on the straight line portion of the curve, resulting from log of exposure  $E_2$  and  $E_1$ . The average gradient value ranges from 2.5 to 3.5. The gradient is the mean slope between two specified densities. A high gradient refers higher radiographic contrast of the film.



**FIG. 7.8:** Characteristic curve of X-ray film

### Speed

The speed refers to the sensitivity of the film-screen combination. Fast film requires less radiation exposure to achieve a given film density and slow film requires more radiation exposure. There are two types of speed in use, one is the absolute speed and the other is the relative speed. The absolute speed of a film is defined as the reciprocal of the exposure in roentgens ( $1/R$ ) that is required to produce a density of 1.0 above the base plus fog density. It is determined from the H and D curve and used only in performance evaluation.

The relative speed is a measure compared with a standard film screen combination. For example, calcium tungstate screen-film combination is called par speed and given a value of 100. An other combination of film-screen having twice the speed of calcium tungstate is given the speed of 200. In this way, the rare earth screen combination is given speed of 400, which is used in general radiography. Angiograms involve short exposures, may require a speed of 600. Bone and extremities require more detail or slow films.

### Latitude

Latitude is the range of exposure levels (mAs) that will produce acceptable range of density (0.25–2.0). The latitude is also called dynamic range, varies inversely with film contrast. A wide latitude film has a low gradient and low contrast, whereas higher contrast film may have lower latitude. Hence, a proper balance has to be made

between contrast and latitude. A low latitude film may require number of retakes, because exact exposure techniques are difficult to decide.

## FILM PROCESSING

Film processing involves a series of organized procedures, namely, development, rinsing, fixing, washing, and drying.

### Developing

Development is a chemical process that produces visible image from the latent image. The solution used for this purpose is called developer. The developer converts the exposed silver ion into metallic silver, by reduction process ( $\text{Ag}^+ + \text{e}^- = \text{Ag}$ ). The developer provides the electrons to the sensitivity center.

A developer solution contains, developing agent, activator, restrainer, preservative, hardener, sequestering agent and solvent. Hydroquinone, phenidone or metol are used as developing agents. Hydroquinone is a reducing agent, slow acting and is responsible for the black shades. Phenidone is a reducing agent, acts rapidly and produces lighter gray shades. Phenidone controls the toe of the curve and hydroquinone control the shoulder of the curve.

Sodium carbonate and sodium hydroxide serve as an activator, which swell gelatin, produce alkalinity and control the pH. Thus, it enhances the action of the developer. Potassium bromide is used as restrainer, which is a antifog agent. It decreases the fog, by protecting the unexposed crystals. Fog is the development of unexposed silver halide grains, that do not contain the latent image. Thus, the restrainer, restrict the action of the developing agents only to those exposed AgBr crystals.

Sodium sulfite is used as preservative. The preservative control the oxidation of the developing agent, by air, thereby increasing the life of the developer. Hydroquinone is more sensitive for aerial oxidation. Preservative helps to maintain proper development rate and also maintains balance among developer components. The oxidation products, of the developing agents decompose the alkaline solution and form colored materials that can stain the emulsion. The preservative dissolves these oxidation products and form colorless sulfonates.

Hardener and sequestering agents are used in automatic processors. Glutaraldehyde acts as hardener, to control emulsion swelling and softening. Chelates is used as sequestering agent, which remove metal

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impurities (aluminium ions) and soluble salts. Water is used as solvent to dissolve all the chemicals.

Development depends on crystal size, developer concentration, time, and temperature. Manufacturer recommended concentration, time and temperature must be employed, to get optimal contrast, speed, and fog. Otherwise, it will reduce image quality.

### **Rinsing**

When the X-ray film is removed from the developer, it should be rinsed, in order to remove the soluble chemicals and oxidation products. Rinsing also partially stops the reaction of the developer and neutralizes the alkalinity of the residual developer. Thus, it reduces fog formation. Rinsing is done with water. Insufficient rinsing will shorten the fixer life and also destroys its hardening action. Hence, stains may appear on the film. Rinsing is not necessary for automatic film processor, where the transport rollers squeeze the film and wipe out residual developer.

### **Fixing**

Fixing is the process of making the image permanent without fading. The solution that does the job is called fixer. It removes the unexposed silver halides without damaging the image, hardens the gelatin emulsion and stops residual development.

The fixing solution consists of activator, fixing agent, hardener, preservative, buffer, sequestering agent, and solvent. Acetic acid is used as activator, which neutralizes the developer and stops its action. Sodium thiosulfate or ammonium thiosulfate salts (Hypo) are used as fixing agents. It dissolves the unexposed, undeveloped silver halides, leaving the developed metallic silver. Excess hypo may cause oxidation and make image disclosure to brown, over a period of time. Silver combine with hypo and forms silver sulfide, which appear yellowish brown.

Potassium alum is used as hardener. It hardens the gelatin emulsion, thereby protecting it from physical injuries. When the undeveloped silver bromide is removed from the film, the emulsion starts shrinking. The hardener enhances the shrinking process, and makes the emulsion too hard. Now, it is suitable for transport through washing and drying.

Sodium sulfite is used as preservative, which maintains the chemical balance. The developer may mix with fixer solution and may cause chemical imbalance. Thus, preservative protects the fixing agent from

decomposition. Acetate is used as buffer, which keep the pH of the fixer as constant. Boric acid and boric salts are used as sequestering agent, which removes metallic ions, such as aluminium impurities. The water is usually used as solvent. It helps to mix the chemicals.

### **Washing**

After fixing, the film must be washed with water, to remove fixing-bath chemicals, especially hypo. Otherwise, it will change the black silver to brown silver sulfide. Washing require running water and it takes about 20 minutes. Incomplete washing permits the hyporetention, which may cause image fading, and make the film brown with age. The temperature of the water must be maintained at 3°C in automatic processor.

### **Drying**

The final step in processing is to dry the radiograph. The film may be dried in dust free open air area, where the temperature is less than 35°C. Hot air drying cabinets are also useful for drying, which is equipped with a fan and heating element to flow hot air. Wetting agents such as photo-flo or alcohol can be used to decrease the drying time. In automatic processor, warm dry air is blown over both surfaces of the film, while it is moving in the drying cabinet.

## **INFLUENCE OF TEMPERATURE AND TIME**

Development is a chemical process, which depends on both temperature and time. Higher temperature increases reduction of silver and enhance the number of silver grains. It is vice versa at lower temperature. Hence, optimum time and temperature (20–22°C) is required. At higher temperature,  $> 24^{\circ}\text{C}$  the emulsion become so soft, and produce chemical fog. To overcome this, development time must be decreased. One quarter of development time is decreased for one degree temperature increase.

At low temperatures,  $< 16^{\circ}\text{C}$  the action of hydroquinone ceases and the resulting radiograph lacks contrast and density. This can be overcome by increasing the developer time. To achieve good development 1/4 minute of development is increased for one degree of temperature decrease. In manual development, radiographic exposure is based on 5 minute development at 29°C, whereas it is 22 s in automatic processor. Processing of films on the above basis is called time-temperature development.

## **REPLENISHER**

Replenisher is used to restore the concentration of developer chemical to the original value. It also compensates the reduction in alkalinity and overcomes the accumulation of bromide. One gallon (4.5 liters) of replenisher should be added for every 40 numbers of 14" x 17" films or their equivalent area. In properly replenished developer, one can develop 125 number of 14" x 17" films per gallon of solution. Replenisher should never be added to the developer, while the films are developing. If it is added, steaks of high density will be produced.

## **DARKROOM**

The darkroom is a film processing room. It is located adjacent to the X-ray room. It must have sufficient space, about 10' x 10' x 10' (cubic feet). The walls of the darkroom must be thick enough to protect against radiation. The floor should be durable, easily cleaned, not slippery and resistant to staining and corrosive substances. Windows should be avoided and air-conditioning is the ideal choice. The darkroom is connected to the X-ray room by means of a pass-box. It has two light tight X-ray proof doors, which are revolving on vertical axis. The exposed and unexposed cassettes may be passed on through the pass-box.

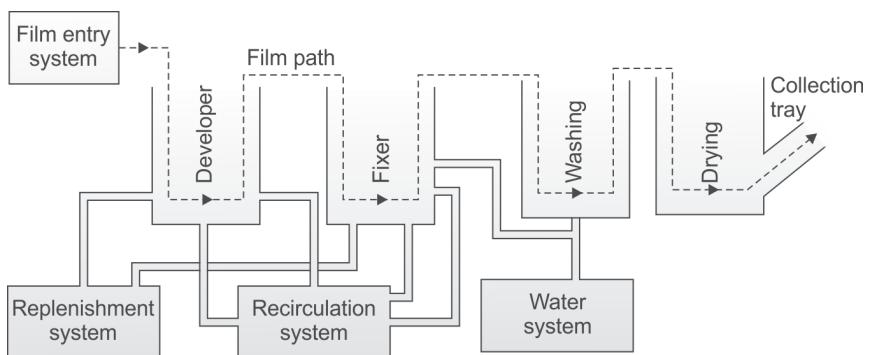
The entrance of the darkroom must be light tight and are provided with interlocked doors. The room is illuminated by a safe light, which will not fog films. Usually, red lamps of < 15 watts are used as safe lights. The working distance between the safe light to the film should not be less than 1.2 m. The effectiveness of the safe light is tested periodically.

The work area of the darkroom is divided as dry side and wet side. The dry side consists of loading bench, cassette components, film bin, storage for reserve film, brackets for film, hangers and was paper basket. The wet side consists of processing chemicals, two stirring paddlers, thermometer, stop clock and processing tanks. The simplest type consists of a three compartment tank, meant for developing, rinsing and fixing. In order to prevent electrical shock, all metallic objects must be earthed.

## **AUTOMATIC FILM PROCESSOR**

Nowadays film processing is done automatically, which provides consistent and uniform quality radiographs. An automatic processor

run the film sequentially through the developer, fixer and washing solutions. The total developing time is only 90 s. The automatic processor consists of a number of separate but interrelated systems. They are transport system, temperature control system, circulation system, replenishment system, and dryer system (Fig. 7.9).



**FIG. 7.9:** Automatic film processor

## TRANSPORT SYSTEM

The transport system consists of feed tray, entrance rollers, micro-switch, roller assembly, transport racks and drive motor. The roller assembly consists of transport rollers, master roller and planetary roller. The film is inserted into the feed tray in the darkroom. The shorter dimension of the film should always be against the side rail, to maintain proper replenishment rate. The transport rollers (1 in dia.) are in pairs, opposite to one another and keep the film in correct path. The master roller (3 in dia.) helps the film to turn around with the help of planetary rollers and guide shoes. A crossover rack helps the film to move from one tank to another. A drive motor with 10–20 rpm transfers power to transport rack and drivers and rollers through belt and pulley or chain and sprocket or gears. The transport system also controls the time to which the film is immersed in the chemical. The micro switch controls the replenishment rate of the processing chemicals.

## TEMPERATURE CONTROL SYSTEM

Temperature of the developer, fixer and water should be maintained precisely. The developer and wash water should be maintained at 35°C and 32°C, respectively. Hence, a heating element controlled by a thermocouple is provided for each tank.

## **CIRCULATION SYSTEM**

The developer and fixer chemicals should be mixed by agitation, to have constant temperature. The circulation system pumps the chemicals continuously, and provides agitation in each tank. In the developer, the circulation system filter the particles of  $100\text{ }\mu\text{m}$  size released by the emulsion. Hence, these particles will not reach the rollers, thereby reducing artifacts. Such a filter system is not required in fixer tank. Water is circulated in the wash tank in an open system, to remove all processing chemicals from the film. Usually, the inlet is provided at the bottom and the outlet is at the top. The water overflows and comes out from the tank at the rate of 12 L/min.

## **REPLENISHMENT SYSTEM**

In every film processing, some amount of developer and fixer is absorbed. As a result, the chemical level in the tank drops and decrease the processing time. The replenishment system monitor this loss and preserves both quality and quantity of the chemicals in the processor. It consists of replenisher tanks, filters, and replenishment pumps. This system helps the developer to maintain its alkalinity and strength. It also helps the fixer solution to maintain its acidity and strength. The replenishment rate is adjusted for each film processing. It is 60–70 mL for developer and 100–110 mL for fixer for every 14 inches film.

## **DRYER SYSTEM**

The dryer system consists of blower, heater, ventilation ducts, drying tubes, and an exhaust system. This system work on negative air pressure and absorbs moisture from the film. The blower sucks the room air and blows on the heating coils (2500 W). The temperature of the air entering system is monitored by a thermocouple. The drying tubes are positioned on both sides of the film and the hot moist air is vented out.

The automatic processor is very much suitable to a busy X-ray department. It reduces the processing time (90 s), and improves the efficiency, work flow and image quality.

## **IMAGE QUALITY**

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The image quality is defined as the ability of the film/detector to record each point in the object as a point on the film. It is used to describe the visibility of diagnostically important detail in the radiograph.

High-quality images are required to make accurate diagnosis. The image quality depends on (i) contrast, (ii) spatial resolution, (iii) noise, (iv) geometric factors, (v) detective quantum efficiency, and (iv) sampling and aliasing.

## CONTRAST

The term contrast refers the difference in density between adjacent areas in the radiograph. The contrast may be a subject contrast and film contrast. The product of subject contrast and film contrast gives the radiographic contrast. Subject contrast is the difference in X-ray intensities transmitted through different parts of the patient. It depends on patient thickness, tissue mass density, effective atomic number, shape of the subject, and photon energy (kVp).

Thicker and thinner body part attenuates radiation differently, and varies the transmitted X-rays. The subject contrast is proportional to the relative number of transmitted X-rays. Tissues of equal thickness, having different mass density contribute to subject contrast. Photo-electric absorption varies with effective atomic number. If the adjacent tissues effective atomic number is different, they contribute subject contrast. Contrast media such as Barium ( $Z = 56$ ), and Iodine ( $Z = 53$ ) will enhance subject contrast.

Shape of the anatomy, which coincides with X-ray beam increases the subject contrast. Shapes that have change in thickness for X-ray path may reduce subject contrast. High kVp gives lower subject contrast, where as low kVp gives higher subject contrast.

The film contrast tells us how the film responds to difference in exposure. The film contrast depends on, characteristic curve, film density, screen or nonscreen exposure, and film processing. High contrast film is made of homogeneous size of silver grains, where as low contrast film is made of heterogeneous grains. Double emulsion films produce greater contrast than single emulsion films. The fog and scatter will reduce film contrast. They produce unwanted film density, which lowers final radiographic contrast.

## RESOLUTION

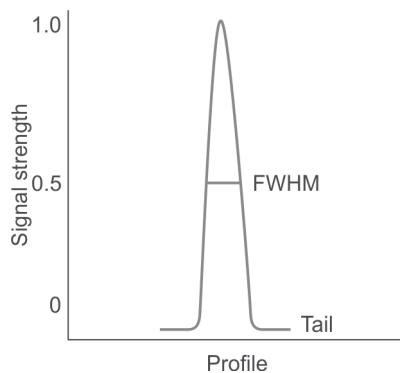
Resolution is the ability to image two closely placed small objects, as two independent images. There are three types of resolution, namely, spatial resolution, contrast resolution and temporal resolution. Spatial resolution refers the ability of the imaging system to record the object in the two special dimensions ( $x, y$ ) of the image. In other words,

it is the ability to image small objects that have high subject contrast, e.g. bone-soft tissue interface. Contrast resolution is the ability to distinguish anatomical structures of similar subject contrast, e.g. liver-spleen. In general, film-screen radiography has excellent spatial resolution. Temporal resolution is the ability of the imaging system to localize the object in time, from frame to frame and follow its movement. Temporal resolution is high for fluoroscopy.

To measure spatial resolution, number of functions are defined. They include point spread function (PSF), line spread function (LSF), edge spread function (ESF) and modulation transfer function (MTF). The image produced for a single point object is called PSF, e.g. imaging a lead sheet that has tiny hole ( $10\text{ }\mu\text{m}$ ) or performing CT scan on a wire, kept perpendicular to the slice plane. Then, the image profile is measured by a densitometer, that gives the PSF (Fig. 7.10). The dimension of the profile is measured at half width, which is called full width half maximum (FWHM). If the separation of two point sources are greater than the FWHM, than only they can be resolved. The presence of scatter radiation will broaden the FWHM, with extended tails, which creates image blur or unsharpness.

Though PSF describes the response of the imaging system, it represents a discrete point in the image surface and it is not suitable for system like film-screen, involving fixed area. Hence, functions like LSF and ESF are recommended. LSF describes the response of an imaging system to a linear stimulus. In this, imaging is done with a slit ( $10\text{ }\mu\text{m}$ , platinum), and 90 degrees image profile is measured, by a densitometer. This can be measured for both vertical and horizontal axis. Similarly, edge spread function (ESF) is measured in fluoroscopy, using a sharp edge.

The easiest way of describing the resolution is line pair (lp), in frequency domain. It refers a bright stripe and a adjacent dark stripe in the image and its unit is lp/mm. This is in analogy to a sound wave (sine) having frequency in cycles/mm. Objects of an image that are separated by small distances (mm), possess high spatial frequency,



**FIG. 7.10:** Point spread function

similar to sound waves. If  $D$  is the size of the object, then spatial frequency ( $F$ ) =  $1/2D$ . For example, if object size is 0.5 mm, then spatial frequency is  $1/(2 \times 0.5) = 1$  lp/mm. The human eye can resolve about 5 lp/mm at a viewing distance of 25 cm. The resolution of screen/film radiography is 5–10 lp/mm.

Modulation transfer function (MTF) is the ratio of the recorded signal frequency (output contrast) to original signal frequency (input contrast) and it is always less than 1. For a imaging system, it is plotted spatial frequency versus MTF, for a given frequency (Fig. 7.11). Lower spatial frequency corresponds to bigger objects that have higher MTF. Higher spatial frequency corresponds to smaller objects that have lower MTF. An imaging system usually consists of several components as in fluoroscopy, and hence each component MTF should be taken into account. The total MTF of a system is the product of the entire component's individual MTF. It is calculated from the measurements of LSF.

Thus, MTF is a useful to quantify the resolution of each component in a imaging system. In day-to-day practice, measurement of spatial resolution is done by using star phantoms in radiography and line pair phantoms in computed tomography.

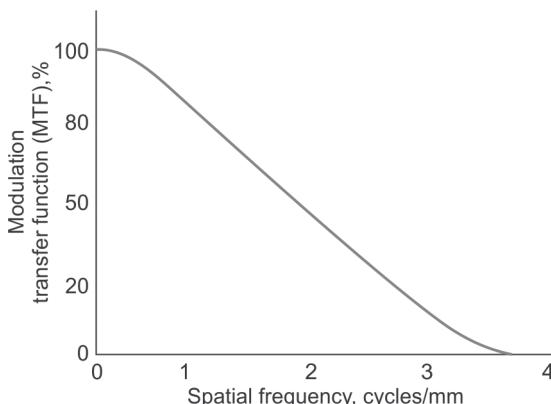


FIG. 7.11: Resolution capability of an imaging system

## NOISE

The noise (mottle) is the random fluctuation of film density about some mean value following uniform exposure. It degrades image quality and limits the ability to visualize low-contrast objects. Noise is mainly made up of screen, film, and quantum noise. The screen noise is caused by nonuniformities in screen construction. Film noise is caused by the grain structure of emulsions. Quantum noise is caused by the

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discrete nature of X-ray photons and it is the most important source of noise in radiography. The adjacent areas of the film receive photons that differ from the mean value, which contribute to quantum mottle.

If  $N$  is the number of X-rays recorded by a pixel in a detector, then noise ( $\sigma$ ) =  $\sqrt{N}$ , where  $\sigma$  is the standard deviation. The relative noise, or coefficient of variation (COV) =  $\sigma/N$ . As the number of photons ( $N$ ) increases, the relative noise decreases. The inverse of the relative noise is called signal-to-noise ratio (SNR), hence

$$\text{SNR} = N/\sigma = N/\sqrt{N} = \sqrt{N}$$

To increase the SNR, the number of photons ( $N$ ) reaching the detector has to be increased. But, it increases the patient dose, hence optimal balance of SNR and radiation dose is a must. The detector does not detect all the incident photons. The screen-film system detects only 60% of the incident X-ray photons. To account this, another parameter called quantum detector efficiency (QDE) is defined. QDE is the ratio between number of photons detected ( $N_d$ ) and the number of photons incident ( $N_i$ ). Therefore, the above equation may be rewritten as:

$$\text{SNR} = \sqrt{N_d} = \sqrt{\text{QDE} \times N_i}$$

The equation does not include other sources of noise such as screen, and CCD in fluoroscopy, etc. Therefore, detective quantum efficiency (DQE) is defined for imaging system. DQE is the ratio of the square of the output SNR to the square of the input SNR:

$$\text{DQE} = \text{output SNR}^2/\text{input SNR}^2$$

Noise of an image has frequency and one can plot noise intensity versus frequency for an imaging system. This kind of plot is called the Wiener spectrum (WS) and it used to quantify noise. The optimal frequency for radiology is 0.2–1 lp/mm and for a screen-film combination, the Wiener spectrum may be related to the MTF as follows:

$$\text{WS} = \frac{G^2}{N} \times \text{MTF}^2$$

where,  $G$  is the film gamma and  $N$  is the mean number of photons per square mm. Thus, DQE and WS are the parameters of interest by which the performance of a imaging system can be certified.

## GEOMETRIC FACTORS

### Magnification

Image quality is affected by geometric factors, such as magnification, distortion and focal spot blur. All radiographic images are magnified and the magnification ( $M$ ) is the ratio between the image size and object size. If SID is the source to image distance and SOD is the source to object distance, then

$$M = \text{SID}/\text{SOD}$$

When the object is closer to the source, the magnification is larger. When the object moves away from the source, magnification decreases. For chest radiography the SOD is about 180 cm, and the magnification is unity. Lesser the magnification means image blur is less and higher the resolution.

### Distortion

Distortion is the result of unequal magnification of different parts of an object. It may be caused by object thickness, object position and object shape. Thick objects produce more distortion than thin objects.

Patient with irregular anatomy may contribute to distortion in a radiograph. If the object plane and imaging plane are not parallel then distortion occurs, due to positioning. The distortion is minimal for object that is positioned at the centre. Object that is positioned lateral to the center may have severe distortion. The objects that are lateral may have unequal magnification than that at the centre. The angle of inclination of the object also influences the degree of distortion.

### Focal Spot Blur

The focal spot ( $F$ ) of an X-ray tube is not a point and have a dimension (0.6–1.8 mm), which produce penumbra at the edge of the field. Penumbra is the region at the edges, where the radiation intensity decreases laterally. It causes blurred region at the edges of the field in a radiograph, which is called focal spot blur ( $f$ ).

$$f = F(M-1)$$

The focal spot blur increases with large focal spot size and higher magnification. It is small on the anode side and large on the cathode side, due to Heel effect. To reduce blur, smaller focal spot size and lesser magnification should be used. To have lesser magnification, the patient-film distance is reduced by keeping them close to each other.

## **OPTIMAL QUALITY IMAGE**

Radiographs with quality images can be produced with proper patient preparation, selection of imaging devices and correct exposure techniques.

The patient anatomy should be placed closer to the receptor/film. The axis of the anatomy should lie parallel to the receptor plane. The central X-ray beam should pass through the center0 of the anatomical region. If multiple anatomy are to be imaged with equal magnification, then all the anatomical structures must be positioned at equal distance from the film. The patient must be immobilized to avoid motion blur. Motion blur can be reduced by the following:

1. Short exposure time.
2. Providing instructions to the patient.
3. Keeping the source to image distance higher.
4. Keeping the object to image distance smaller.

Selection of exposure technique plays an import role in obtaining a good quality image. The exposure time should be always shorter, which will improve image quality. Shorter exposure time reduces motion blur. High frequency generator provides short exposure time than single phase generators.

The kVp controls the radiographic contrast, since it influences both quantity and quality of X-rays. As the kVp increases, higher amount of X-rays are transmitted through the patient. It reaches the film and affects the optical density. Compton interaction also increases, differential absorption is reduced, resulting reduction of subject contrast. The scatter radiation that reaches the film is higher, which increases noise in the image, resulting loss of contrast. As the contrast is low, the latitude is larger and the margin for error is increased. Hence, high kVp reduces contrast and the only advantage is the reduction of patient dose with wide latitude of exposures.

The mA controls optical density in the film, since it influences quantity of X-rays. As the mA increases, the number of X-rays reaching the film increases, resulting in higher optical density. The radiographic noise is lower but the patient dose is higher. Too low mA and high mA shift the optical density away from the straight line portion of the characteristic curve. Thus, it indirectly affects the radiographic contrast.

Therefore, use of high kVp with reduced mA and short exposure time is recommended to obtain a quality radiographic image.

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# Computed and Digital Radiography

## INTRODUCTION

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Screen-film devices are replaced by receptors that can produce digital images. Basically, both computed radiography and digital radiography are capable of producing digital images. The advantages of digital images are: (i) data can be stored and transferred, (ii) it allows post processing, (iii) display of images in computer monitor, and (iv) use of picture archiving and communication systems (PACS) for tele-radiology, etc.

Digital system uses binary numbers, for which the base is 2 and has two digits, namely, 0,1. Whereas we generally use decimal numbers which has a base 10 and uses digits 0 to 9. In a binary, the value of a digit in a position is 2 times than it is in right. In a decimal number, it is 10 times than it is in right. Decimal point is allowed in decimal system, whereas it is not allowed in binary. For example, the value of a binary number 10 is

$$= (1 \times 2^1) + (0 \times 2^0) = 2$$

Similarly, one can convert any binary number into a decimal number by raising powers of 2 in a series and then adding them. Table 8.1 gives the equivalence of binary and decimal numbers.

Digital memory and storage uses the term bit, bytes and words, etc. A bit is a small portion of a disk or tap that can be magnetized for data storage. Bits are grouped in bytes and words and 1 byte (B) is equal to 8 bit. The number of bits in a word may be 16,or 32 or 64, depending upon the computer system. Normally, kilobytes ( $2^{10} = 1024$  byte), mega bytes ( $2^{20} = 1048$  kB) and giga bytes ( $2^{30} = 1073$  MB) are commonly used.

The use of binary numbers in digital image storage is as follows: The image is divided into number of matrix elements called pixels.

**TABLE 8.1** Binary and equivalent decimal numbers

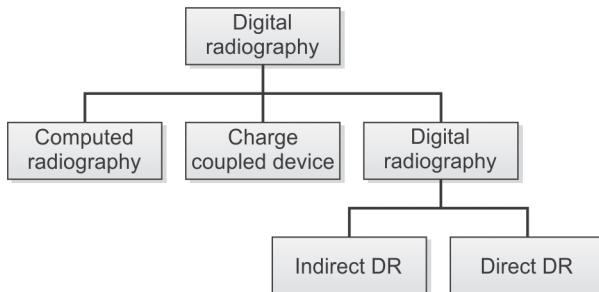
Binary number	Decimal number
0	0
1	1
10	2
11	3
100	4
101	5
110	6
111	7
1000	8
1001	9
1010	10

Each pixel is assigned a value, related to signal intensity. The value stored in the pixel is in binary format and the maximum value that is stored is called bit depth. A pixel having a high value represents dark gray shade, whereas pixel of low value represents white gray shade. This is analogy with film, where high dose gives dark and low dose gives white shade. A single bit can store either black or white, similar to an electrical switch, which can be switched ON or OFF.

In general, computer memory and storage use many bits and each bit have two states, namely, 0 or 1. Similarly 2 bit may have 4 possible states, 3 bits have 8 states and so on. In general, if there are N bits than the possible states are  $2^N$ , e.g. 8 bit may have  $2^8$  states that is equal to 256 gray shades. However, radiography involves large field size and a resolution of about 10 lp/mm, to provide optimal contrast. Therefore, it requires higher bit depth of 8, 12 and 16, respectively. For example, 16 bit depth can provide 65,536 numbers of gray shades. That is why digital radiography needs large storage space in the range of 4–32 MB, which is very much higher compared to a computed tomography scanner.

An ideal digital radiography system should have the following: (i) the physical design should have compatible size with screen-film cassettes, (ii) immediate read out facility, (iii) robust and less costly, (iv) high quantum efficiency with low radiation dose for image capture, (v) spatial and contrast resolution similar to film, and (vi) wide dynamic range and DICOM compatible. Presently, these systems are employed

in the form of (i) computed radiography (CR), (ii) charge-coupled detector (CCD), and (iii) digital radiography (DR). The DR is further classified into indirect digital radiography and direct digital radiography (Fig. 8.1).



**FIG. 8.1:** Different path of getting digital radiography images

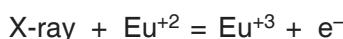
## COMPUTED RADIOGRAPHY

Computed radiography (CR) employs a phosphor, which work on photostimulable luminescence (PSL) principle. When the phosphor is exposed to radiation, it absorbs and store radiation energy. Later, if it is stimulated by a different light source, it gives luminescence. The amount of luminescence is proportional to radiation exposure.

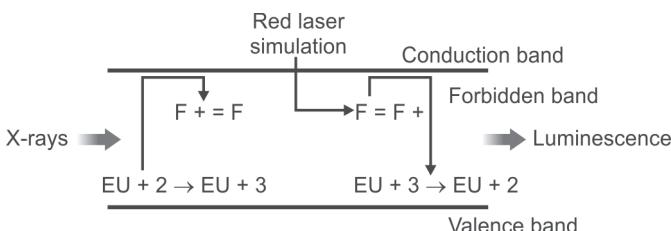
## PHOTOSTIMULABLE PHOSPHOR

The commonly used phosphor is barium fluorohalides: BaFBr (85%) and BaFl (15%): Eu (europium). The europium is called activator and it is present in small quantity, which is responsible for the PSL property. The atomic numbers of BaFBr are 56, 9 and 35 with K-shell binding energy of 37, 5 and 12 keV, respectively. The radiation interaction are mostly by Compton and photoelectric with outer shell electrons.

The activator creates defects in the crystals (F-center), which can trap electrons. When the phosphor is exposed to radiation, the divalent europium atoms ( $\text{Eu}^{+2}$ ) get oxidized into trivalent  $\text{Eu}^{+3}$  with release of electrons in the valence band (Fig. 8.2). These electrons move from the valence band to conduction band, later they are trapped at the F-centers in the forbidden zone. The electrons can stay in these centers for longer period of time. Thus, billions of electrons are trapped in the F centers. The number of electrons per unit area is proportional to the absorbed radiation energy.



Over the time, these electrons may return to the ground state on their own. However, exposing the phosphor to a red light source may accelerate the electron return process. That is why, it is called photo-stimulable phosphor. When the imaging plate is scanned by red laser light, the F center absorbs energy and transfer the same to the electrons. The electrons reach the conduction band, where they become mobile again. They move to the valence band, with emission of blue-green light. The electron joins with  $\text{Eu}^{+3}$  and is converted into  $\text{Eu}^{+2}$ . The blue-green light energy is greater than that of laser light energy.



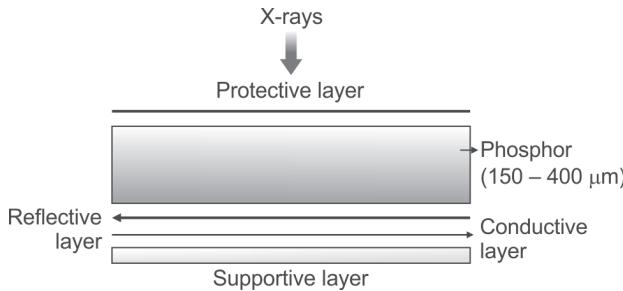
**FIG. 8.2:** Principle of photostimulable phosphor

The phosphor will not give up all its trapped electrons in the first stage of laser light. It will retain some amount of trapped electrons. Hence, it is to be exposed to very bright light source, which moves all the trapped electrons to the valence band, thus emptying the F centers. Now, the phosphor can be used for another radiation exposure.

## IMAGING PLATE

The phosphor can be made as flexible screen, which is enclosed in a rugged cassette and is called imaging plate (0.5 mm). Imaging plate was first introduced by Fuji, Japan, in 1983 and is similar to that of a screen-film cassette. The cross section of a imaging plate is shown in Figure 8.3. In the imaging plate, the PSP particles are randomly present through out the binder and it can be handled similar to that of screen-film cassette.

The cassette are available in various sizes:  $14 \times 17$  inches,  $14 \times 14$  inches,  $10 \times 12$  inches, and  $8 \times 10$  inches' with a pixel range of  $200 \times 200 - 100 \times 100 \mu\text{m}$ . It is available in variety for general radiography, mammography, etc. The matrix sizes available are  $1760 \times 2140$  for normal resolution and  $2000 \times 2510$  for high resolution. The spatial frequency is 2–3 lp/mm for standard radiographic work and 10 lp/mm, for mammography work.

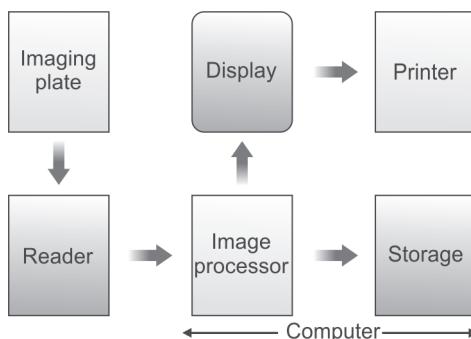


**FIG. 8.3:** Cross section of a imaging plate

## PHOSPHOR READER

The CR system consists of imaging plates of various sizes, a reader, computer and a printer in addition to X-ray unit (Fig. 8.4). The computer offers image processing, storage and image display facility. The imaging plate is used instead of screen-film cassette. It can be re-used again and again by erasing old image data. The following is the procedure involved in computed radiography:

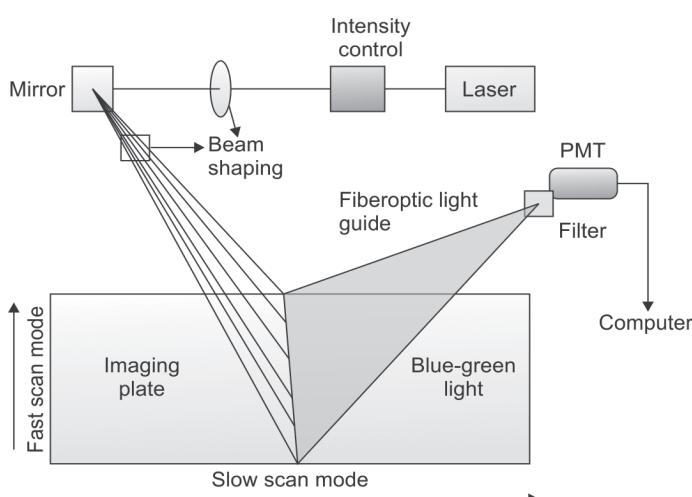
- CR cassette is exposed to X-ray beam,
- Cassette is Inserted into the reader,
- Imaging plate is removed from the cassette,
- Scanned by the He–Ne laser beam (633 nm), blue-green or UV, light (390 nm) is released from that location (x, y),
- PMT is used to collect this light, and gives an electronic signal
- Electronic signal is digitized and stored in memory,
- The plate is exposed to bright white light, to erase the residual energy, for another use.



**FIG. 8.4:** Block diagram of computed radiography system

The reader is most critical part of CR imaging system. The CR consists of entry system for imaging plate, laser light source and a photo multiplier tube (PMT) as shown in Figure 8.5. After the radiation exposure, the CR cassette is inserted into the reader, where the imaging plate is removed and fitted to a drive mechanism. The drive mechanism moves the plate with constant velocity along the y axis. This is usually done with slow motion and is called slow scan mode.

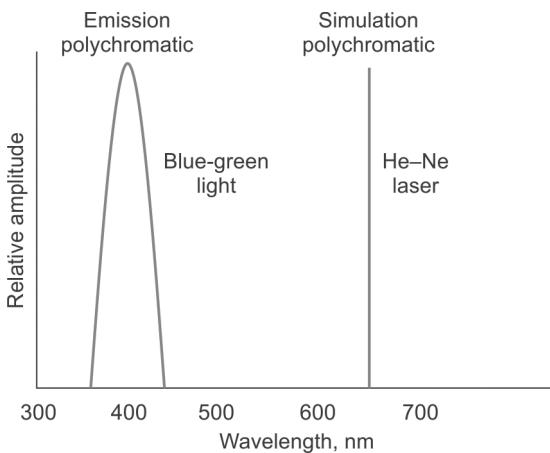
A rotating and multifaceted mirror reflects the red light from a laser light source (He-Ne laser, 633 nm). This light is deflected back and forth across the phosphor plate in the horizontal, X-direction, which releases visible, blue-green light of 390 nm (Fig. 8.6). This means that the trapped energy due to radiation exposure is released from that spatial location (x, y). This is done in fast scan mode. The slow and fast scan modes are controlled by the CR computer.



**FIG. 8.5:** Computed radiography reader

The above light is collected by the PMT or charge coupled device (CCD), through the fiberoptic light guide. The PMT amplifies the signal and gives the output electronic signal, which is an analog signal varying with time. This is fed into a computer, where it is processed for amplitude, scale and compression. Then, the signal is digitized with sampling and quantification and finally stored in a hard disc. Sampling and quantification are the import process in the analog-digital conversion. Sampling refers to the time between samples and quantification refers to the value of each sample.

For every spatial location  $x, y$ , a gray scale value is obtained. The wavelength of laser light and blue-green light is different. The scattered laser light may reach the PMT, spoil the signal and creates noise. To avoid this, an optical filter is mounted in front of the PMT. This filter attenuates the scattered laser light, and transmits the blue-green light emitted by the phosphor, thereby increasing the signal-noise ratio.



**FIG. 8.6:** CR spectrum: Simulation and emission of light

In some other systems, the cassette is inserted vertically and imaging plate is withdrawn downwards, during which it is scanned by a horizontal laser. In this, the imaging plate is not completely removed from the cassette, which avoid roller damage. The laser scanning is done right angle to the grid lines, which also avoid aliasing artifacts.

The laser beam size is very important and it should be kept less than  $100 \mu\text{m}$  at the mirror level. The laser beam shape, size, speed and intensity must be kept constant at the imaging plate level. This is achieved by means of a beam shaping devices. A reader can process about 70 cassettes in one hour and it take about 110 s to process each cassette.

## IMAGE DISPLAY

The digital image that is obtained from analog digital converter is fed into a image processor, where smoothing and edge enhancement, etc. are carried out. Finally, image is displayed either in video display or hard copy film. The array processor feeds the image information to

a digital-analog converter. The analog signal modulates the laser beam, which scans the film and image is recorded. This is a different He-Ne laser, only used for making hard copy. The film has one side emulsion with peak sensitivity at 633 nm, to match the laser. Multiple images may be printed in a single film.

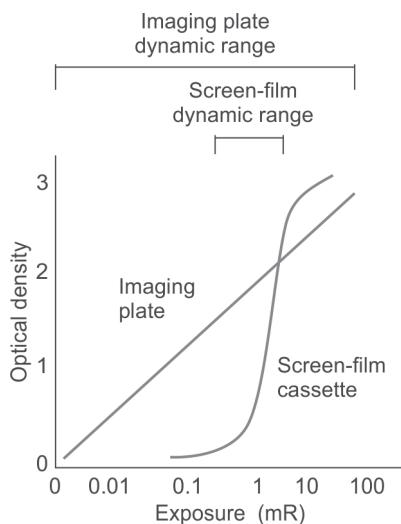
## IMAGE CHARACTERISTICS

The main advantage of imaging plate is its wide dynamic range and it has wide latitude towards radiation exposure (Fig. 8.7). The technologist has freedom to select his exposure techniques. But, technologist cannot understand his error, since it is adjusted in the image post processing. However, the retake rate is lesser compared to screen-film radiography.

Images produced with low radiation exposure involve higher quantum noise. Whereas images produced with high radiation exposure involves low noise, but high patient dose. The main source of noise is the scatter radiation. The drive mechanism, laser and computer devices also contribute to noise.

The CR systems are faster compared to 400 speed screen-film system. Hence, images can be produced with lower patient dose. In screen-film radiography, kVp control contrast and mA control optical density, this concept is no more valid in computed radiography. Because CR image contrast is constant, irrespective of exposure techniques. Therefore, high kVp and lower mAs can be used to produce CR images, which reduce patient dose further.

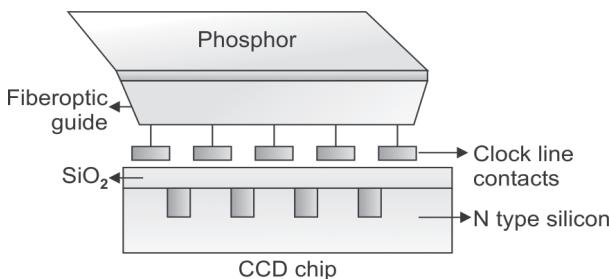
The spatial resolution is less (5.5–3.5 lp/mm) compared to screen-film systems (8–12 lp/mm) and it is preferred in portable radiography. It is very difficult to access detector dose and some vendors came with detector dose indicators (DDI). Hence, these systems require elaborate quality assurance.



**FIG. 8.7:** Comparison of dynamic range of imaging plate and screen-film systems

## CHARGE-COUPLED DEVICES

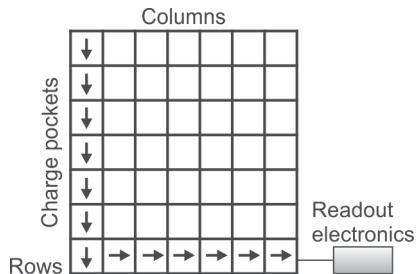
Charge-coupled devices (CCD) form images from visible light. It is usually used with intensifying screens and image intensifier tubes (Fig. 8.8). Basically, CCD chip is an integrated circuit, made up of amorphous silicon. Its surface is edged with pixel electronics, e.g.  $2.5 \times 2.5$  cm CCD may contain  $1024 \times 1024$  pixels in its surface. The silicon surface is photoconductive. If it is exposed to visible light, electrons are liberated and build up in the pixel. Higher the light intensity, larger the electrons, that are liberated. The electrons are kept in the pixel by electronic barriers on each side of the pixel. Thus, each pixel act as capacitor and collect charge, proportional to light.



**FIG. 8.8:** Design of charge coupled device

Electronic charge in each pixel is read out along column wise. The electron in each pixel is shifted to another pixel, by adjusting the voltage barriers of each pixel (Fig. 8.9). Thus, the charge pocket in one column moves in unison and finally reaches the pixel in the bottom row. The bottom row is read out pixel by pixel and the charge is shifted to the read out electronics, which produces an electronic signal. This signal is digitized by a analog-digital converter and the digital signal is used to construct image matrix with bit depth of 8–12.

Similarly, next column charges are shifted to the bottom row that gives another signal. This process is repeated until all the pixels in the detector are read out completely. The read out is faster and it is at the rate



**FIG. 8.9:** Movement of charge pockets column by column through bottom row

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of 30 frames per second. The CCD geometry is uniform and distortion free. It has wide dynamic range with low electronic noise.

### **APPLICATIONS OF CCD**

CCD produces high quality images and have application in, dental radiography, mammography, fluoroscopy and cineradiography. In dental radiography, intensifying screen is coupled with CCD and the field of view (FOV) is too small ( $25 \times 50$  mm). The light emitted by the screen is collected by the CCD efficiently. Since the coupling is too good only little light is wasted.

In digital biopsy mammography, the FOV is higher than the area of the CCD and hence, fiberoptic taper is used between the intensifying screen and CCD. The fiberoptic taper acts as a lens and focus the light emitted by the screen on the CCD surface. The input and output surface of the fiberoptic taper is  $50 \times 50$  mm and  $25 \times 25$  mm, respectively. The loss of light is not significant in this system.

In chest radiography, the FOV is larger ( $35 \times 43$  cm) than the CCD surface and the amount of light lost is higher (99.7%). The amount of light lost is proportional to the demagnification factor required to couple the input area to the output area. As less number of photons are used to construct an image, a secondary quantum sink will occur. This will increase the noise and reduce the image quality. This will also result in higher radiation dose to the patient.

### **DIGITAL RADIOGRAPHY SYSTEMS**

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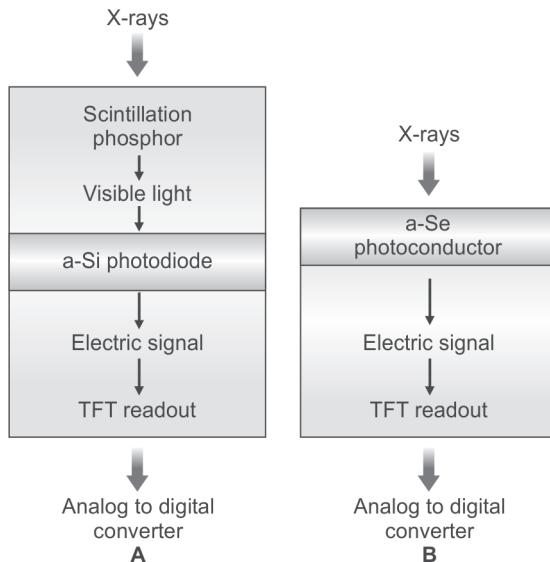
Digital radiography systems consists of large area, flat panel, solid state detectors with integrated thin film transistor (TFT) readout and having fast access with best image quality. It should have higher spatial resolution, contrast resolution and dose efficiency. In general, they are available in two configurations, namely,

1. Indirect detection flat panel systems,
2. Direct detection flat panel systems,

In indirect systems, the X-rays are converted into light by a phosphor and then light is converted into electric signal. In direct systems the X-rays are converted directly into an electric signal (Fig. 8.10).

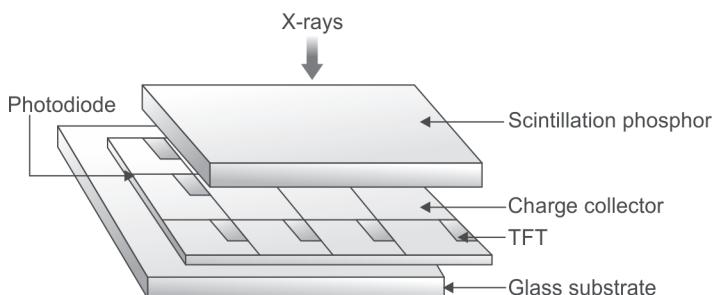
### **INDIRECT DETECTION FLAT PANEL SYSTEMS**

Indirect detection flat panel systems consist of a Scintillation phosphor, amorphous silicon photo diode (a-Si) and flat TFT arrays. The scintillation



**FIG. 8.10:** Principle of digital radiography: (A) Indirect and (B) Direct detection flat panel systems

crystal used is CsI:Tl or Gd<sub>2</sub>O<sub>2</sub>S:Tb that converts the incident X-rays into light. It works similar to that of an intensifying screen in a cassette. The detector base is the glass substrate, on to which light sensitive a-Si with a capacitor and TFT is embedded in the form of pixels. The top most component is the scintillation phosphor (Fig. 8.11). Amorphous silicon is a fluid that can be painted on a given surface. The entire assembly is put in a protective enclosure with external cable connections. There is no need of fiberoptic guide as in the case of CCD for image demagnification.



**FIG. 8.11:** Indirect detection flat panel system

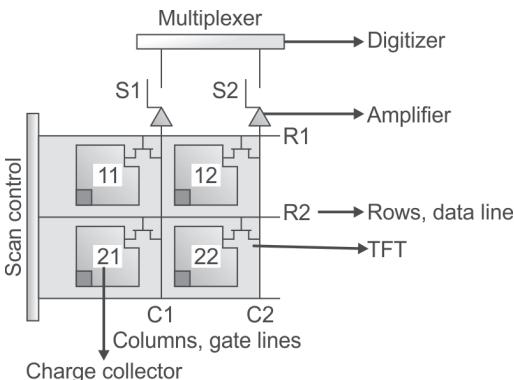
The thin film transistor (TFT) has three connections, namely, gate, source and drain respectively (Fig. 8.12). The source is the capacitor, drain is connected to the read out line (vertical column) and the gate is connected to the horizontal (rows) line. TFT is basically an electronic switch that can be made ON and OFF. When negative voltage is applied to the gate, the TFT is said to be OFF and if positive voltage is applied to the gate, it is said to be ON.

Initially, the capacitor of each detector element that stores the charge is earthed, so that all the residual charges are passed on to the ground. When exposed to X-rays, the scintillation emits visible light, which in turn exposes the light sensitive photo diode (a-Si). The photodiode release electrons, so that charge build up in each detector element, which is stored by the capacitor. Later, the charge in each detector element is read out by the electronics as follows.

During the X-ray exposure, negative voltage is applied to the gate and all the transistor switches are in OFF position. The charge accumulated in each detector element is stored in the capacitor. During the read out process, positive voltage is applied to the gate, one row at a time. Thus, the switches of detector elements in a given row are made ON. This will connect vertical wires C1, C2, to the digitizer through switches S1, and S2. The multiplexer select the column sequentially (one column at a time) and the charge is amplified and allowed to move to the digitizer. Thus, the gate selects a row and multiplexer selects a column and the charge in each detector element is read out sequentially. Finally, the signal is digitized and stored for image analysis.

### Phosphor Materials

Two type of phosphor materials are commonly used, namely, terbium doped gadolinium oxisulfide ( $\text{Gd}_2\text{O}_2\text{S:Tb}$ ), and thallium doped cesium iodide ( $\text{CsI:Tl}$ ). The  $\text{Gd}_2\text{O}_2\text{S:Tb}$  ( $Z = 64$ ) is a unstructured crystal produced in a uniform layer held in a binder and it is borrowed from screen-



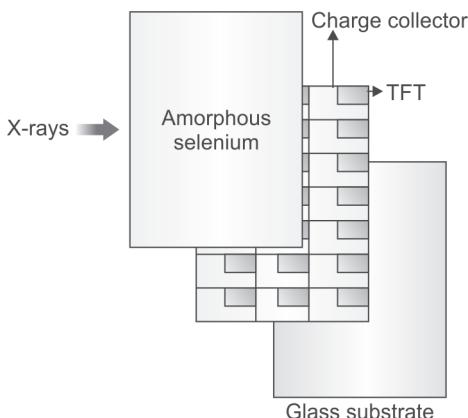
**FIG. 8.12:** The TFT read out process by scan control

film technology. These systems lose light energy by scatter accounting lateral light spread. As a result, X-ray photon interaction spread to adjacent pixels, which reduces the spatial resolution.

CsI:Ti ( $Z = 55$ ) is obtained from the image intensifier technology. It consists of discrete monoclinic needles of  $5\text{--}10 \mu\text{m}$  wide and  $600 \mu\text{m}$  long. These crystals are hygroscopic and quickly degrade if not completely sealed. Since it pushes the light in the forward direction, light spread is reduced. This facilitates design of thicker phosphor material which increases X-ray photon interaction and quantum efficiency.

### DIRECT DETECTION FLAT PANEL SYSTEMS

Direct detection flat panel systems use X-ray photoconductor material like amorphous selenium ( $a\text{-Se}$ ,  $Z = 34$ ), which directly converts X-rays into electrical signal (Fig. 8.13). There is no intermediate material like scintillation phosphor which converts X-rays into light, as in the case of indirect detector flat panel systems. Since selenium is in amorphous form, large area plates can be made by vapor deposition, which is a cost effective and reproducible technology. It has good X-ray detection properties and high spatial resolution. Selenium is a photo conductor, and it alters its electrical conductivity, when exposed to X-rays. The altered electrical signal is proportional to the intensity of X-rays.



**FIG. 8.13:** Direct detection flat panel system

Initially, 5 kV bias voltage is applied to the surface of the selenium. Later, when it is exposed to X-rays, it emits electrons, which discharge part of the applied voltage. The amount of discharge is proportional to the radiation intensity, resulting in latent charge image. These charges are stored in the capacitor, and the pattern of charge is read out by scan control lines, similar to that of indirect systems. Finally, the signal is amplified, digitized for image analysis. Selenium is susceptible to humidity and temperature variations and requires protection from environment.

## COMPARISON OF DETECTOR SYSTEMS

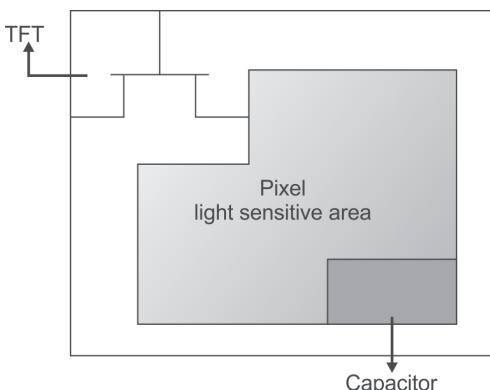
DR systems should have high signal to noise ratio and provide connectivity to DICOM, hospital information system (HIS) and radiological information system (RIS). The probability of photon interaction with detector material is given by the quantum efficiency and it is higher for gadolinium oxy sulfide. This can be increased by increasing the detector thickness and materials having high attenuation coefficient. Both direct and indirect detector system have wide dynamic range compared to screen-film system. However in practice, this wide range is restrictively used, because low exposure gives noisy image and high exposure increase patient dose.

The light collection efficiency of each detector element depends on the fractional area that is sensitive to light, which is defined by the fill factor as follows:

$$\text{Fill factor} = \frac{\text{Light sensitive area}}{\text{area of detector element}}$$

In DR, detector is occupied by conductors, capacitors and TFT, and only partial area of the detector is sensitive to X-rays (Fig. 8.14). Hence, the fill factor is always less than 100% and this will vary depends upon the detector system. The fill factor is higher for direct a-Se system compared to indirect DR systems.

Small detector elements give higher spatial resolution, but reduce the fill factor. Low fill factor reduces the signal to noise ratio, resulting in poor contrast resolution. Hence, there is trade off between spatial resolution and contrast resolution for a given detector. The specifications of various detector materials are given in Table 8.2.



**FIG. 8.14:** Fill factor is the ratio of the light sensitive area to detector element

## COMPARISON OF IMAGE QUALITY

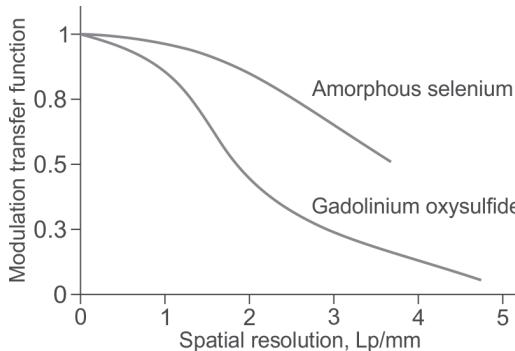
Digital radiography offers on line imaging and the image is read out from the receptor quickly. There is no need to remove the receptor, which increases patient through put. The images can be transmitted electronically and identical copy of the image can be made. Use of

**TABLE 8.2** Specifications of DR detector materials

Parameter	Gd <sub>2</sub> O <sub>2</sub> S:Tb	CsI:TI	a-Se
Active area	22.5 cm × 28.7 cm	43 cm × 43 cm	35 cm × 43 cm
Thickness	500 µm	550–600 µm	500–1000 µm
Element array	2256 × 2878	2688 × 2688	2560 × 3072
Pixel pitch	100 µm	143 µm	—
Element pitch	140 µm	173 µm	139 × 139 µm
Fill factor	52%	68%	86–100%
Spatial resolution	5 Lp/mm	3.5 Lp/mm	3.6 Lp/mm
Pixel depth	12 bit	12 bit	14 bit
Display time	3 s	10 s	7 s

computer for image archiving, and post adjustment of contrast is also possible. This will result in improved interpretation and improved diagnosis. The various factors that influence the image quality are (i) spatial resolution, (ii) noise, and (iii) detective quantum efficiency.

The spatial resolution is determined by the pixel spacing in the detector. The frequency that limits the resolution is the Nyquist frequency, which is inverse of twice the pixel spacing. The spatial resolution of digital radiography systems is lower compared to screen-film system. However, they have better contrast resolution than film. The MTF of the direct detector digital radiography system is high, up to the Nyquist frequency (Fig. 8.15). Higher Nyquist frequency and MTF facilitates visualization of finer diagnostic details.

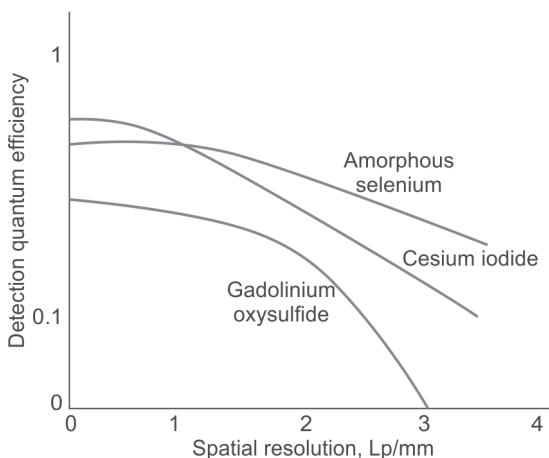


**FIG. 8.15:** MTF for direct and indirect detector systems as a function of spatial resolution

Noise in the image quality depends on the quantum efficiency of the phosphor, charge collection by the capacitor and noise free read

out of the stored signal. The indirect detector system uses light that undergoes scattering, resulting in reduction of signal-to-noise ratio.

Detective quantum efficiency (DQE) of an ideal imaging system is 100%. However, in practice it is less than 100%, due to inefficiency in incident X-ray detection and internal source of noise. The DQE of a-Se is higher (50%) compared to CsI and Gd<sub>2</sub>O<sub>2</sub>S (Fig. 8.16). The DQE decreases at high spatial frequency due to increased noise in the image.



**FIG. 8.16:** DQE of various detector materials as a function of spatial frequency at 70 kVp

The indirect detector system suffers due to light scattering, and there is trade off between X-ray absorption and MTF. Thinner scintillation crystal gives high MTF, but reduces X-ray absorption. Thicker crystal increases the X-ray absorption, but reduces the MTF due to increase of light scatter. In the case of direct detector system, the thickness can be increased, so that X-ray absorption can be increased without loss of MTF.

### Radiation Exposure

Digital radiography likely to reduce radiation exposure to the patients, compared to screen-film systems. This is possible without loss of image quality. Reduction of number of retakes is the main cause of dose reduction in DR system. In addition, it has wider dynamic range and superior quantum detection efficiency that also helps in dose reduction.

## PICTURE ARCHIVING AND COMMUNICATION SYSTEM

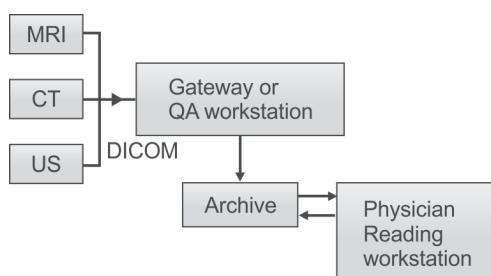
The picture archiving and communication system (PACS) is a provision, used in medical imaging technology. It provides cost effective and easy access to images from multiple imaging tools. Radiological images and reports are transmitted digitally through PACS. It eliminates manual filing, retrieving and transport of films. DICOM (digital imaging and communications in medicine) is the universal format for image storage and transfer, which is used in PACS.

Basically, PACS require four components namely, (i) imaging device, (ii) network system, (iii) workstation, and (iv) storage. It delivers the images timely and provide easy access to images and interpretations. It avoids traditional film based image retrieval, distribution and display.

The PACS finds variety of uses that includes: (i) replacing hard copy such as films, (ii) providing remote access, including distance education and teleradiology, (iii) providing electronic image integration platform with easy access to HIS (hospital information system), RIS (radiology information system), and (iv) helping radiology workflow management such as patient examinations.

The imaging tools include CT, ultrasound, MRI and PET, etc. The images from the modalities are sent to the quality assurance workstation, called PACS gateway (Fig. 8.17). It checks the patient demographics as well as attributes of the study. If the study information is correct, images are passed to the archive for storage.

Then, the radiologists review the images through their workstations and make the final report. The workstation and archive is a bidirectional transmission. PACS uses web based interfaces to use internet or wide area network (WAN) as their way of communication, via VPN (virtual private network) or SSL (secure sockets layer). The client side software includes Activex, Javascript and Java Applet. Very good backup for



**FIG. 8.17:** Picture archiving and communication system, block diagram

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patient images is required, in case of loss of images from PACS. Hence, the images are automatically sending their copies to a separate computer for storage.

### **TELERADIOLOGY**

Teleradiology (TR) is the transmission of patient images from one location to another location. The images include X-ray, CT, ultrasound and MRI, etc. The main purpose is to share the images or study with other radiologists and physicians. Since the number of radiologists is lesser than the imaging procedures, teleradiology fill the shortage of radiologists.

Teleradiology improves patient care, by allowing radiologist services, who is physically not present at that location. This is highly true in specialist such as MR radiologist or Neuroradiologist, or Periodic radiologist, etc., who are available only in urban cities. Teleradiology allows round the clock service of the specialists without interruption.

Teleradiology uses internet, telephone, wide area network (WAN) and local area network (LAN). Specialized software is used to transmit images. Advanced technologies such as graphic processing, voice recognition, and image compressions are also used in teleradiology.

# 9

# Fluoroscopy Imaging

Fluoroscopy is an imaging technique used to obtain real-time moving images of the internal structures of a patient through the use of a fluoroscope (Thomas A Edison, 1896). Whereas in radiography, images are made with transmission of X-rays through the body, with film as detector. Fluoroscopy is basically a dynamic imaging, where the radiologists view images of moving organs continuously, while the X-ray beam is ON. Real-time imaging requires 30 frames per second similar to that of television technology. This will facilitate to follow motion of an organ over a period of time. Hence, fluoroscopy imaging possesses higher temporal resolution. Since fluoroscopy is a dynamic process, the radiologists must adapt the moving images, even though images are dull. This requires the knowledge of image illumination and visual physiology.

Fluoroscopy has wide variety of applications, such as photospot imaging, spot film acquisition, digital subtraction angiography, endoscopic examination, lithotripsy and cine-radiography. Fluoroscopy can be discussed in two headings, namely, (i) direct vision fluoroscopy and (ii) image intensifier fluoroscopy.

## **VISUAL PHYSIOLOGY**

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Human eye consists of cornea, lens and retina. There is a structure called iris between the cornea and lens, which control the light that falls on the eye. Light from an object passes through cornea, lens and then retina. In bright light, the iris contracts, allows less light to the eye. In low light, the iris expands and allows large light.

Human vision is due to rods and cones in the retina of the eye. Basically, rods and cones are very small structures and there are about  $10^5$  numbers per square mm in the retina. Cones are present at the center of the retina (fovea centralis), whereas rods are present at the

periphery. Rods are sensitive to low light levels, whereas cones responds to intense light. Thus, cones help for day light vision (photopic vision) and rods are responsible for night vision (scotopic vision). Cones detect small objects better and also differentiate brightness levels. It is sensitive to wide range of wavelength of light and capable of detecting color. Rods unable to detect color and hence become color blind. In scotopic vision, eyes are sensitive to green light (555 nm). The transformation of scotopic vision into photopic vision is called dark adaptation, which takes place in direct vision fluoroscopy.

## **DIRECT VISION FLUOROSCOPY** ---

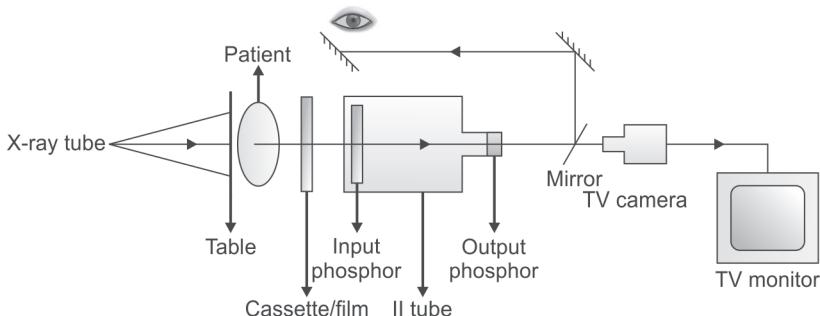
In direct vision fluoroscopy, the X-rays that are transmitted through the patient are passed on to a scintillation phosphor screen ( $ZnS$ ), which gives an image as faint scintillations. The radiologists used to view the image directly with red goggles in a darkroom. The thick phosphor converts the X-rays into light proportionally, but the brightness is too low. The screen was backed with lead glass to reduce radiation to the radiologist. This type of fluoroscopy had been in use until 1950, which is called as conventional fluoroscopy. It was discontinued due to the following reasons.

The light output of the fluorescent screen is very poor, for a given exposure rate. The light conversion efficiency of the screen is also very low and the spatial resolution is very poor. The radiologists can see only a small percentage of the light from the screen, due to narrow viewing angle ( $6^\circ$ ). The contrast of the fluoroscopic image is  $1/10$ th of radiographic image, because the visual acuity of the eye is 10 times lesser at low light level. Since the images are so faint, it has to be viewed under dark conditions with red goggles. However, it requires 10–20 minutes, to adopt full darkness. In addition, the patient may receive higher dose of radiation and hence this modality is not in use to day.

## **IMAGE INTENSIFIER FLUOROSCOPY** ---

Modern systems use image intensifiers (II) and a closed circuit TV system. When X-rays passes through human body it is absorbed differentially. This results in a variation of the transmitted radiation. This transmitted radiation is made to fall on a phosphor through a image intensifier. A collimator limits the size of the X-ray beam automatically

to the proper field of view (FOV). As a result, a visible image is formed on the fluorescent screen. This image can be viewed directly or indirectly by the radiologist. The image can be observed by a video camera and presented on a television monitor (Fig. 9.1)



**FIG. 9.1:** Fluoroscopy imaging with image intensifier and TV system

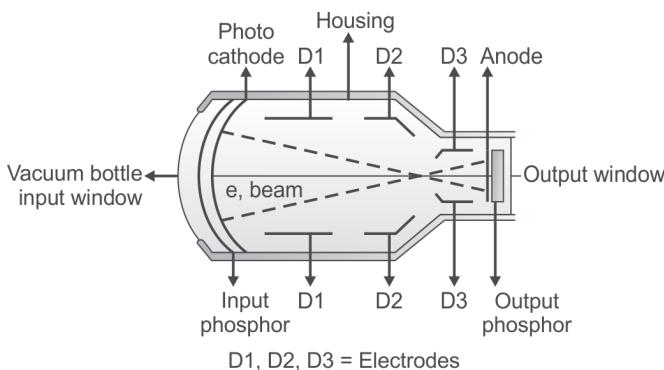
Fluoroscopy must have maximum image detail, for which greater image brightness is required. The image intensifier fulfills the above requirement. Image brightness mostly depends on the anatomy, kVp and mA. Hence, by controlling the kVp and mA, the image brightness can be varied in fluoroscopy. Fluoroscopy examination requires several hours, and the radiation dose to the patient is high. In order to reduce the patient dose, the exposure rate (200) is much lower in fluoroscopy. Fluoroscopic system uses low current (1–5 mA), output and typically produces 30 images every second. This shows that only fewer X-ray photons are used in forming a single fluoroscopic image. Therefore, fluoroscopic images are statistically inferior to radiographic images. Hence, there is a need of high gain detector system in fluoroscopy.

## IMAGE INTENSIFIER

In the beginning, fluoroscopy was performed by viewing the live image, produced by X-rays on a thick intensifying screen. The room must be completely dark so that faint glow of the screen could be seen. The radiologists also wear red goggles for dark adaptation. The difficulties involved in dark adaptation and working in dark, lead to the development of image intensifiers (II). The II convert the X-ray beam incident upon it to a visible light image.

The image intensifier is an evacuated glass envelope (vacuum bottle), which contains four basic elements. They are (i) input screen, (ii) focusing electrodes, (iii) anode, and (iv) output screen as shown in the Figure 9.2.

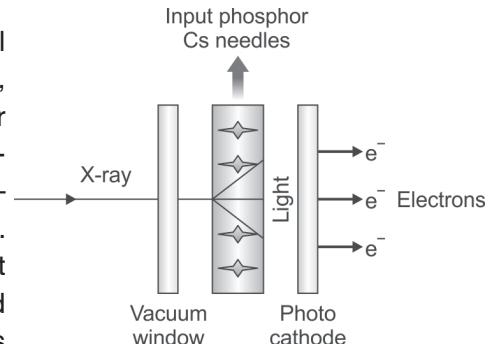
The patient side of the vacuum bottle has the Al window (1 mm), which is a curved one to withstand air pressure. The evacuated glass envelope limits the size of the II and the diameter ranges from 23 to 57 cm. The field size can be reduced electronically using electrostatic focusing. The glass envelope is mounted inside a metal container, which will avoid damage and rough handling.



**FIG. 9.2:** Image intensifier and its components

### Input Screen

The input phosphor follows the Al window and has three layers, namely, a curved substrate layer (0.5 mm Al) to support the phosphor, CsI input phosphor (200–400  $\mu\text{m}$ ) and a photocathode (Fig. 9.3). X-rays from the patient passes through the Al window and falls on the input phosphor. This phosphor absorbs the X-rays and converts the X-ray energy into visible light. This is similar to the action of intensifying screens in cassette. The photocathode is coated on the inner side of the phosphor, which converts the visible light into electrons. It is thin metal layer and commonly used photocathode is antimony and cesium ( $\text{Sb}_2\text{Cs}_3$ ) compounds. Many light photons emit one electron. The number of electron that is emitted is directly proportional to the light intensity, which in turn is proportional to the intensity of input X-rays. For example, a 60 kV X-ray photon can emit about 8000 light photons, followed by the emission of 400 electrons.

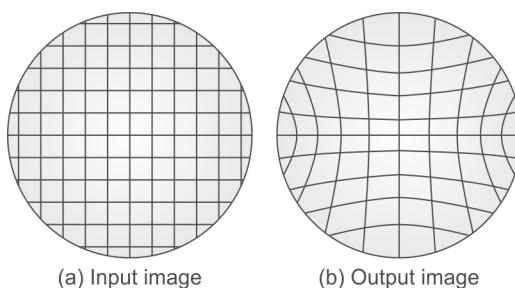


**FIG. 9.3:** The input screen of the image intensifier tube

CsI phosphor is commonly used as input phosphor, because of its special property. It is a vertically grown needle like crystal ( $5 \mu\text{m}$ , dia.), tightly packed, that can push the light in the forward direction. It also reduces lateral spread of light, which can minimize the unsharpness. The K-edges of cesium (36 keV) and iodine (33 keV) falls under the bremsstrahlung spectrum of fluoroscopy X-rays, which increases its absorption efficiency (60%). This will reduce patient dose. The input screen is maintained at a high negative potential compared to anode, to accelerate the electrons. The screen size may vary from 150–400 mm in diameter, depending upon the clinical application.

### Focusing Electrodes

There are 3 electrodes (D1, D2, and D3) between the input screen and anode. They are basically metal rings, which are given positive voltage, with respect to the photocathode. They accelerate the electrons and focus them on the output screen. The image intensifier is about 50 cm long, and a potential difference of about 25,000 V is maintained between the photocathode and anode, to accelerate the electrons. The electrons arrives the anode with high velocity and energy and contain the image of the input phosphor. Thus, the electrons gain energy and forms a minified, inverted image at the output screen. The curved nature II tube enhances the electron focusing, but results in pincushion distortion. The resultant image at the output phosphor appear as distorted (Fig. 9.4).



**FIG.9.4:** Pincushion distortion

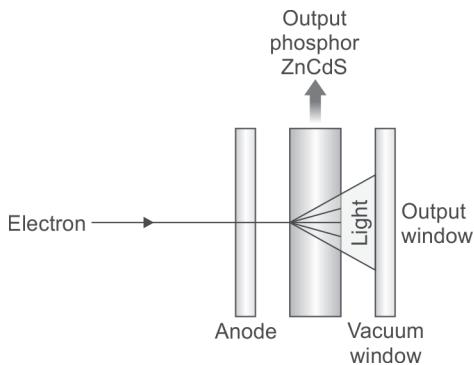
### Anode

The anode is a circular plate with a hole in the middle to permit the flow of electrons. It is made with a thin coating of Al ( $0.2 \mu\text{m}$ ) on the inner side of the output screen. It is electrically conductive and its potential is 25 kV higher than the input screen. The anode

receives the accelerated electrons, so that they deposit their energy in the output phosphor.

### Output Screen

The output phosphor most widely used is zinc cadmium sulfide ( $Zn\text{cds}$ : Ag) compound doped with silver. The output phosphor is small and it is about 25–35 mm in diameter and thin (4–8  $\mu\text{m}$ ) enough to preserve resolution (Fig. 9.5). It absorbs the electrons and emits large amount of green light, for which the video camera is very sensitive. Each electron may produce about 1000 light photons. The ratio between the number of light photons at the output phosphor and the number of X-rays in the input phosphor is called flux gain. The back side of the screen is covered with Al (0.5  $\mu\text{m}$ ) to prevent backward light emission. The total light emission of the output screen is proportional to the input X-ray intensity.



**FIG. 9.5.** Output screen of the image intensifier tube

### Image Intensifier Performance

The performance of the II tube is basically due to its increased illumination of the image, which is caused by multiplication of light photons at the output phosphor and image minification. The ability to increase the illumination depends on the brightness gain of the II tube. Hence, the terms brightness gain and conversion factor are used to estimate the performance of a II tube. Brightness gain is the ratio of the brightness of the output phosphor to that of the input phosphor.

$$\text{i.e. Brightness gain} = \frac{\text{Brightness of the output phosphor}}{\text{Brightness of the input phosphor}}$$

It is made up of two factors, namely, flux gain and minification gain. Flux gain is the ratio of the output screen light photons to the input screen light photons and it is of the order of 50.

$$\text{i.e. Flux gain} = \frac{\text{Number of output light photons}}{\text{Number of input X-ray photons}}$$

Minification gain describes the intensification caused by the smaller size of the output screen. It is equal to the square of the ratio of the input and output screens. If  $d_1$  and  $d_2$  are the area of the input and output screen then,

$$\text{Minification gain: } (d_1/d_2)^2$$

If  $d_1$  and  $d_2$  are 300 and 30 mm respectively, then the minification gain is  $(300 \div 30)^2 = 100$ . The overall brightness gain is the product of flux gain and minification gain. In the above example, it is of the order of 5000 ( $50 \times 100$ ).

Brightness gain is not a measurable quantity, and hence the term conversion factor is introduced to evaluate the performance of the II tube. Conversion factor is defined as the ratio of the brightness (luminance,  $\text{cd}/\text{m}^2$ ) of the output phosphor to the input exposure rate ( $\mu\text{Gy/sec}$ ).

$$\text{i.e. Conversion factor} = \frac{\text{Brightness of the output phosphor}}{\text{Input exposure rate}}$$

The typical values are 25–30  $(\text{cd}/\text{m}^2).(\mu\text{Gy/sec})^{-1}$ . It is in the order of 50–300 and the corresponding brightness gain range is 5000–30,000. Internal scatter of X-rays, and electrons may reduce contrast of the II tube, and it is called veiling glare. The gain and conversion factor deteriorates over a period of time and hence require quality assurance.

### Magnification Mode

Changing the voltages of the intermediate electrodes, electron crossover point of focal point can be moved nearer the patient. This will reduce the FOV, irradiate only smaller volume of tissue and the image appear magnified, since it fills the entire screen on the monitor. This will minimize scatter and increase the image contrast. Thus, magnified images can be obtained on the output screen, with higher spatial and contrast resolution with higher patient dose. However, the magnification mode will reduce the minification gain, resulting in lesser brightness of the image. To restore the brightness, higher exposure factors (increase of mA) are used in magnification mode. This will increase the patient's skin radiation dose. The magnification factor is directly proportional to the diameters. If an II tube of 25 cm diameter is operated in 12 cm mode, then the image magnification is  $25 \div 12 = 2.1$ .

In general, II tubes are provided with 1–4 magnified fields, which are called multi field image intensifiers. For example, a II tube of 25/17/12 cm

is trifield tube, which can be operated with 25 cm, 17 cm and 12 cm FOV. When it is operated in 17 cm FOV, the focal point moves towards the patient and operates in magnification mode (Fig. 9.6).

## TELEVISION SYSTEM

The television (TV) system consists of a video camera, monitor and a optical coupling. The video

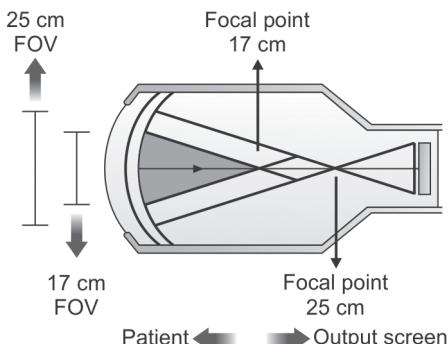
camera converts light image from the output phosphor into electrical signal, which is proportional to the light falling on the camera plate. This can be viewed on A TV monitor, which enlarges the image to the original size with real-time image display. The advantage of TV monitor display includes (i) the brightness level can be adjusted, (ii) multiple observers can see the display at the same time, and (iii) image storage in the electronic format. There are two methods of producing electrical signal, namely, (i) thermionic television camera and (ii) charge coupled device. However, the TV camera is a vacuum device that produces only analog video signal.

### Thermionic Television Camera

This type of camera consists of cylindrical housing that contains the camera tube and electromagnetic coils to steer the electron beams. The tube consists of electron gun, grids and the target, which serve as anode (Fig. 9.7). The gun produces electrons by thermionic emission, which are passed through the grid and get accelerated towards the anode. The electrons are further accelerated and focused by the electrostatic grids. The size of the electron beam and position is controlled by the electromagnetic coils, namely, deflection coils, focusing coils and alignment coils.

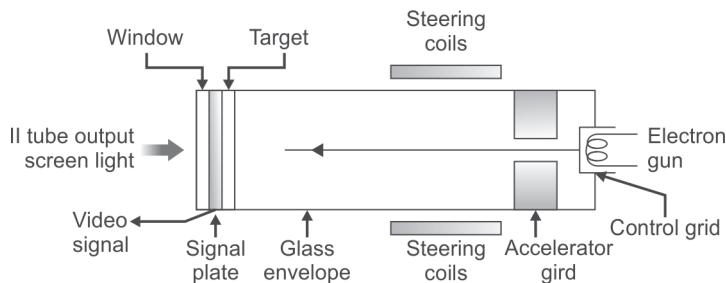
The target assembly consists of a thin window, signal plate and a target. The inner side of the window is coated with thin layer of metal or graphite. This graphite layer transmit light as well conducts electricity, is called signal plate. The inner side of the signal plate is coated with photo-conductive layer of antimony trisulfide, and act as a target.

When the II tube light falls on the window, it is transmitted to the target through the signal plate. If the electron beam strikes the same



**FIG. 9.6:** Magnification of II tube changes the focal point towards the patient

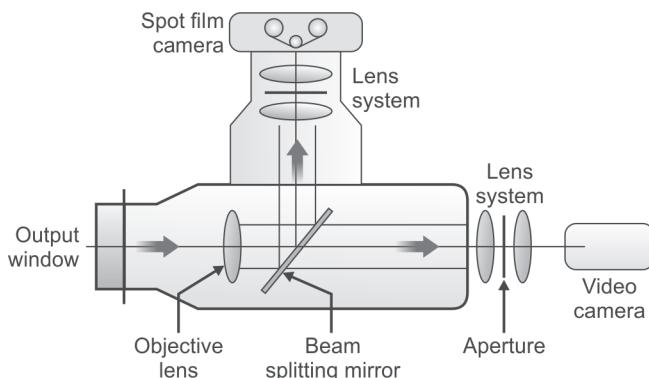
area of the target at the same time, electrons moves from the target, via signal plate and outside, by forming video signal. Whenever there is no light in the target area or in dark, no video signal is produced. Thus, the video signal is proportional to the intensity of light from the II tube.



**FIG. 9.7:** Vidicon television camera and its parts

### Image Intensifier and Camera Coupling

The optical system connects the II tube and camera, so that there is no loss of light. The output phosphor diameter should be equal to the window of the television camera tube. There are two ways of optical coupling, namely, lens system and fiberoptic system. The lens system is a traditional method, bigger in size, suitable for cine or photospot camera (Fig. 9.8). The objective lens receives the light from the output phosphor and converts into a parallel beam. Whenever image is recorded on film, the mirror split the beam, and sends a part of a beam to



**FIG. 9.8:** Image intensifier and the video camera is coupled with a lens system

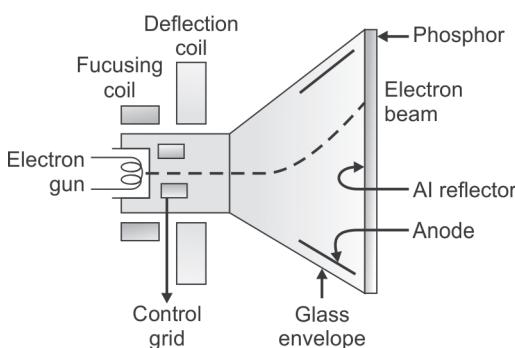
the video camera. The remaining portion will be sent to the film camera. This will enable to view images during recording. The beam splitting mirror is retracted from the beam, when the film camera is not in use. Both television camera and film camera are coupled through lenses. The lens and mirror position needs to be very precise to produce good images.

The fiberoptic system is efficient in light collection, which improves geometrical integrity. It is a simplest method with bundle of fiberoptics, which is few mm thick and contains 1000 of glass fibers per square mm. It is easy to move the II tube tower and rugged coupling can be made with fiberoptics. The disadvantage is that it cannot accommodate additional cameras like photospot or cine camera.

### Television Monitor

The video signal from the camera is amplified and passed on to the monitor by means of cable. The main part of the monitor is the picture tube or cathode ray tube (CRT). It consists of a electron gun, control grid and anode assembly in a glass envelope and external coils to steer and focus electron beam (Fig. 9.9). The anode assembly consists of a fluorescent screen and graphite lining. The phosphor is backed with thin layer of Al, which transmits electrons and reflects light.

The video signal received by the picture tube is modulated, proportional to the light intensity received by the camera. The electron beam intensity is varied by the control grid, proportional to the modulated video signal. The electron beam strike the fluorescent screen, and produce light. Thus, the video signal modulates the electron beam of the picture tube and transforms the electron beam into visible image at the fluorescent screen.



**FIG. 9.9:** Cathode ray tube and its parts

The movement of the electron beam follow raster pattern on the screen of the picture tube. That is, the electron beam begins in the upper left corner of the screen and moves to the upper right corner, creating varying intensity of light as it moves. This is called an active trace. Then, the electron beam is turned off and return to the left side of the screen, known as horizontal retrace. A series of active traces and horizontal retraces are repeated until the electron beam is at the bottom of the screen. Thus, one television field is completed.

The electron beam then performs second television field with active trace and horizontal retraces. The only difference is that the present active trace is in between the adjacent active traces of the first television field. This movement of the electron beam is called interlace and two interlaced television fields form a single television frame. If the power supply frequency is 50 Hz then, there are 50 television fields per second. This may give 25 television frames per second and each frame is of 40 ms duration. In practice, 312.5 lines are per television field is used and thus one frame may have 625 lines. Human eye cannot detect flickering, if the frame rate is  $> 20$  frames per second.

The vertical resolution of the monitor depends on the number of scan lines. The horizontal resolution depends on the band pass, which is expressed in Hz. Band pass describes the number of times the electron beam is modulated in one second. For example, band pass of 1000 Hz refers that the electron beam is modulated 1000 times in each second. Higher the band pass, better will be the horizontal resolution. Usually, the fluoroscopy monitors will have a band pass of 4.5 MHz. In general, the spatial resolution of the monitor is lower than that of II tube (1 lp/mm vs 5 lp/mm). Thus, TV monitor degrades the image quality and hence image is recorded in a film with photographic camera.

## LIMITATIONS OF IMAGE INTENSIFIER SYSTEMS

1. The unit is large in size, difficult to position during the procedure.
2. If there is air leak in the II tube, the vacuum is destroyed, resulting in degraded image quality.
3. If the voltage of the electrodes is not correctly adjusted, the electrons will not reach the focal point (defocusing effect). It may result in blurred image with loss of spatial resolution.
4. Variation of magnetic and stray electromagnetic fields from power supply may also give defocusing effect.

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5. The television system depends upon the raster lines and bandwidth frequency for spatial resolution. Its resolution is always lesser than the II tube output screen.

### **X-RAY GENERATOR**

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The X-ray generator for fluoroscopy consists of X-ray tube, high frequency generator, image intensifier, video camera, monitor for real-time display and film camera for hard copy. The X-ray tube should have maximum thermal capacity and heat dissipation. Usually, dual focal spot sizes of 0.3 mm and 1.0 mm are used with anode angle of 8–12 degrees. The focal spot size is not a limiting factor for resolution, since II tube has upper resolution of 4 lp/mm. However, large focal spot under magnification mode create image unsharpness and reduce resolution. High frequency power supply of 15–80 kW is recommended. This will provide maximum power of 450–2000 W for continuous operation. The heat storage range is 0.4–1 MJ. The advantage of high frequency generator is reduced patient dose, due to lesser ripple factor, and smaller in size.

The typical equipments used today are C-arm geometry with following configurations: (i) X-ray tube plus II tube with fiberoptic coupling of CCD, (ii) cantilevered C arm having an X-ray tube with flat panel detector system with DSA, and (iii) cardiovascular system having an X-ray tube with flat panel system. The first type is used for orthopedic intervention, foreign body localization, cholangiography, cystography, pyelography and cardiac pace maker implantation. The second type finds application in vascular intervention studies. The third type is used exclusively for cardiac angiography.

### **PULSED FLUOROSCOPY**

Pulsed fluoroscopy is performed with grid controlled X-ray tube with high frequency generator. Large number of photons are produced within short time interval of each pulse without increasing the kilovoltage. Thus, it offers great degree of dose reduction to the patient with improved temporal resolution. The production of pulses and the video system need to be synchronized. The pulses may be saw tooth or square wave pulse and is generated before the start of each readout cycle. The duration of pulse should be lesser than that of read out cycle. In practice, the pulse gets distorted due to the capacitance of the cable

and hence the pulse have longer rise and decay time. This may cause production of low energy X-ray, which will increase the patient dose.

## **AUTOMATIC BRIGHTNESS CONTROL**

In fluoroscopy imaging, if image intensifier moves from a thinner to thicker region of the patient, higher the amount of attenuation of X-rays. This will reduce the brightness of the output screen. Automatic brightness control (ABC) is a mechanism, which can keep the brightness of the image constant at the monitor. It is basically a feedback circuit, which measure the light intensity of the output screen or videocamera signal. A photomultiplier or a photodiode is used to monitor the light output of the II tube. The corresponding changes will be feedback to the generator for adjustment. The generator will regulate the X-ray exposure rate incident on the input phosphor of the II tube, by changing the kV or mA automatically.

In general, the brightness of the central area of the output screen is taken into account for adjustment. Brightness can be adjusted by both kV and mA, which has influence on contrast and patient dose, respectively. The three methods of adjustments are (i) change of kV at constant mA, (ii) change of mA at constant kV, and (iii) change of both kV and mA. If the II tube moves from thinner to thicker part of the patient, increase of kV may result in lower dose with lesser contrast. But, increase of mA may result in better contrast with higher dose.

Alternatively, the brightness of the image on the monitor can be adjusted by varying the gain of the TV system, which is called automatic gain control (AGC). This will result in increased image noise and unwanted radiation dose. Generally, mA is adjusted as a first step to obtain the input dose rate of II tube. If the current limit is reached, then kV is adjusted to get the input dose rate. Even it is not sufficient then, AGC option is used to adjust the video display brightness.

## **IMAGE RECORDING**

Fluoroscopy images can be recorded by means of (i) spot film cassette, (ii) rapid sequence cut film, (iii) cine fluoroscopy, and (iv) videotape or digital recording.

The spot-film cassette is positioned in between the patient and II tube. Normally, this device is covered by lead, so that it cannot be exposed unintentionally. When the exposure is desired, the control moves

the spot film cassette sideways, and position in the path of the X-rays. Low fluoroscopic mA (3 mA) is changed into high radiographic mA (300 mA). The area of the film can be selected by masking with lead diaphragms so that multiple images can be recorded. Thus, 1–6 images can be recorded in a single film. During the cassette movement and recording, image cannot be viewed on the monitor. The spatial resolution is about 5 lp/mm and the radiation dose to the patient is higher.

Rapid sequence cut film camera is similar to a movie camera and it exposes only one frame during activation. It receives image from the output phosphor of the I<sub>1</sub> tube through the split mirror and involves lesser patient dose. It will not interrupt the fluoroscopic examination, while recording and reduce heat loading of the X-ray tube. This camera uses 70 mm and 105 mm film sizes at the rate of 10 images per second. Larger the film size, better the image quality with increased patient dose. The patient dose is 3 times lesser compared with spot film cassette. It gives good image quality with reduction of spatial resolution. However, spot film cassette gives life size images, whereas above camera image is minified.

Cine fluoroscopy records moving images on a 35 mm cinematographic film with frame rates of 12.5, 25, 50, 100 and 150 frames per second. After processing, the images can be viewed on cine mode. Heart examinations are performed with cine fluoroscopy. It employs double imaging system consists of both anterior and lateral plane, in which the pulsed radiation exposures are synchronized with film transport facility. Grid controlled X-ray tube with small focus is used. The images have good contrast with reduced geometric blurring.

Videotape recording of image is possible directly from the television system. It is economy, but the quality of images is very poor. Alternatively, videomagnetic disk can be used for image recording. It allows instant play back of stored images. Images can be selected after the examination for film recording, before the disc is erased. If the output of the videocamera is digital, it can be stored in a  $1024 \times 1024$  matrix. Later, images are selected for film format recording.

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## DIGITAL FLUOROSCOPY

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The signal from the video camera can be converted into digital format and fed into the computer. The computer will display high resolution digital images, that can be viewed like a movie. Digital fluoroscopy

has faster image acquisition and storage and image manipulation. Its advantages includes (i) image storage facility, (ii) image processing such as noise reduction, and edge enhancement, (iii) black and white reversal, i.e. display of – and + images, (iv) geometrical inversion, i.e. left to right and top to bottom, (v) dose reduction, (vi) wide dynamic range, (vii) dynamic imaging, and (viii) filmless imaging. The images are compatible to picture archiving and communication systems (PACS). In digital fluoroscopy, either charge coupled device or flat panel detectors are used.

### CHARGE-COUPLED DEVICE

The charge-coupled device (CCD) is used to get digital format of the X tube light. It is a solid state imaging sensor without electron gun and deflection coils and evacuation, etc. The input screen of CCD camera is a semiconductor photo sensitive surface, which is divided into thousands of individual photodiodes and arranged in rows and columns. Generally, amorphous silicon (thin) is used as input screen, which is divided into number of pixels ( $1024 \times 1024$ ). Each pixel acts as capacitor and collect charge, proportional to the intensity of light from the X tube output screen.

The collected charge on each pixel is read out row by row. The stored information in each row is shifted electronically and read out as video waveform. The camera is operated under 12 V supply and a lens system is used to focus the light on the input screen. The read out can be done quickly at the rate of 30 frames per second. The CCD camera is having 12 bit image depth, so that the dynamic range is higher. Detail description of CCD and the read out process are given in Chapter 8.

### FLAT PANEL DETECTOR

Instead of X tube and CCD camera, flat panel detectors (FPD) are used in digital radiography, which are also used in fluoroscopy. It consists of array of individual detector elements (DEL), of varying size of 200–140  $\mu\text{m}$ . The size of array ranges from  $25 \times 25$  to  $40 \times 40$  cm. A typical FPD may contain 1.5–5 million DEL's. The commonly used FPD is the indirect solid state system. It is a CsI phosphor with amorphous silicon and thin film transistor. It converts X-ray into light and then into electronic signal. CsI phosphor emits light proportional to X-ray intensity. This light falls on the amorphous silicon photodiode and electrons are

released. The conduction of photodiode varies in proportion to light intensity. The electrons reach the fully charged capacitor and neutralize some of the charges. The remaining charge in the capacitor is drained out and sent to the read out electronics, which measure the change in charge caused by the X-rays. This is repeated many times per second. By reading the DEL row by row, the electronic image distribution of X-rays that are incident on the FPD can be formed. Thus, FPD forms image without TV camera.

CsI phosphor's detective quantum efficiency is 65% and having advantages of increased dynamic range and improved spatial resolution. Hence, flat panel's detect greater range of signals and its contrast resolution is about 3 lp/mm for a pixel size of 150  $\mu\text{m}$ . Image intensifier's maximum spatial resolution is 1.2 lp/mm for a given largest FOV, which is equal to a pixel size of 400  $\mu\text{m}$ . In the case of magnification, flat panel do not improve spatial resolution, since the pixel size remains the same. Alternatively, the image intensifier improves the resolution with magnification.

### **Advantages**

1. They do not exhibit pincushion and S distortion.
2. Vignetting effect is absent and provides excellent uniformity.
3. There is no defocusing effect.
4. It is small in size, easy to position.
5. It does not require TV camera, hence electronic noise is reduced.

### **Limitations**

1. It is difficult to manufacture a flat panel detector without defective detector elements.
2. Software interpolate values for defective detector element, which may create artifacts.
3. Flat panel is sensitive to temperature and mechanical shock.
4. Its resolution is limited by the size of the detective element.

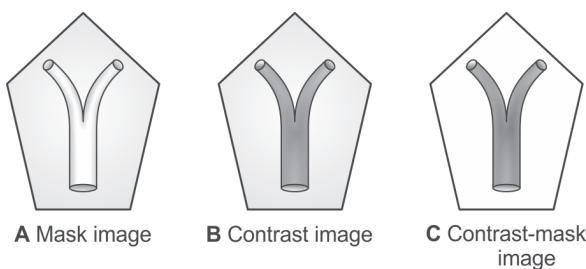
## **DIGITAL SUBTRACTION ANGIOGRAPHY**

Digital subtraction angiography (DSA) is a special method of fluoroscopy, which gives image of the vessels that are filled with contrast. Digital images are obtained before and after injection of contrast medium, to differentiate vascular pathology from surrounding anatomy. It is a noninvasive procedure, that provides improved image quality with lesser use of contrast medium. The principle of DSA imaging is as follows:

**Step 1:** Image of a particular anatomical region is recorded. This is called mask image (A), which shows normal anatomy of the region. It requires 2 frames, the first is used to stabilize the system (technical factors) and the second frame is stored as mask image in the computer (Fig. 9.10).

**Step 2:** The patient is injected with contrast to fill the vessels and the image is taken. It is called contrast image (B), which shows contrast filled vessels, superimposed on the anatomy.

**Step 3:** The mask image is subtracted from the contrast image (B-A), by pixel by pixel basis and stored as contrast-mask image (C). This image reveals only vessels filled with contrast medium.



**FIG. 9.10:** Principle of digital subtraction angiography: (A) mask image, (B) contrast image, and (C) contrast-mask image

The final image is viewed in real time. There should not be any movement of the patient during the above procedure and the images are obtained rapidly. Movement between frames may lead to miss registration of images especially at the bone edges. This can be minimized by the movement of mask, which is called pixel shifting. Pixel shifting can be done manually or automatically.

To reduce the scatter, special algorithm is used. Taking successive frames reduces quantum mottle and increases signal to noise ratio. However, subtracted images reduce the signal to noise ratio and the image looks noisier. Hence, higher mA is used in DSA.

If the length of the anatomy is greater than the FOV (e.g. leg), then multiple images are to be taken to cover the entire anatomy. Each time the contrast is injected to the region. However, modern machine provides software that acquires several mask images and contrast images over the full length of the anatomy. This is facilitated by the table movement and its position for each image. Appropriate mask image

for a given table position is subtracted from the contrast image. There is a chance of movement during the above procedure.

### **DSA Equipment**

Digital subtraction angiography equipment is basically a fluoroscopy X-ray machine with I<sub>1</sub> tube or CCD camera or flat panel DR system. The signal as image frames are acquired, digitized with ADC and stored in random access memory (RAM). There are two memories, one for the mask image and the other for the contrast image. The image content is subtracted in a arithmetic unit, processed and converted back into a analog signal by DAC. Then, the signal is displayed in a high resolution monitor. Images can be stored on a magnetic hard disk and computer needs large memory of the order of 512 MB–1GB. The acquisition and processing is controlled by the CPU of the computer.

### **Logarithmic Subtraction**

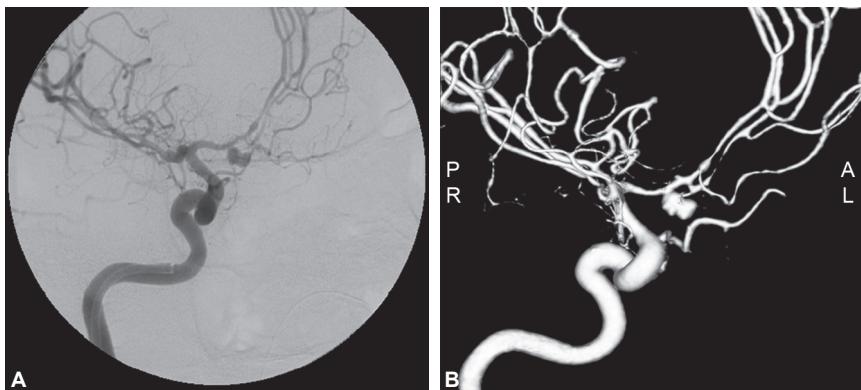
The attenuation of X-rays is exponential and subtraction should fulfill the following conditions: (i) the subtracted image signal must be linear to the concentration of the contrast medium, and (ii) subtracted image should be independent of overlying tissues. Hence, direct linear subtraction of images are not used, instead logarithmic subtraction is used. The mask image and the contrast image are converted into their logarithms initially and then subtracted. Logarithmic subtraction ensures a vessel of uniform diameter with uniform contrast; even it passes through regions of varying thickness. It does not retain stationary anatomical structures, which may obscure small signals. Finally, the signal is converted into intensity levels for display, which is independent of patient size. The intensity depends on the product of vessel thickness and attenuation coefficient of the contrast medium.

### **Rotational Angiography**

In rotational angiography, mask images are taken at several angles, by rotating the X-ray tube and I<sub>1</sub> tube for a 90° rotation. Similarly, post-contrast images are taken for each angle and subtracted suitably. This may be used for 3D reconstruction of vessels, which can be viewed at any angle (Fig. 9.11). This is a useful technique when there is a superposition of vessels that are lying one over the other.

### **Dual Energy Subtraction Technique**

In diagnostic radiology, the linear attenuation coefficient decreases with increase of energy. The decrease is greater in bone than in soft tissue.



**FIG. 9.11.** Rotational angiogram: Cerebral angiogram, endovascular coiling anterior communicating aneurysm: (A) pre- and (B) postprocessed image



**FIG. 9.12:** (A) Normal chest radiograph, (B) Bone subtracted and (C) Soft tissue subtracted

This concept is used to obtain dual energy subtraction. In this technique, images of the same anatomical region are taken rapidly at low and high kV. When the low kV image is subtracted from the high kV image, soft tissue details get canceled, and gives only bone details (Fig. 9.12). Similarly, subtraction of high kV images from the low kV image, cancels details of bone anatomy and highlights only soft tissue. This technique is insensitive to patient motion, and remove effects due to involuntary motion of bowel gas.

However in practice, hybrid subtraction sequence is used in two steps. In step 1, mask image is taken with low and high kV, then low kV-high kV subtraction is obtained. The final image contains only bony details. In step 2, low and high kV images are obtained with contrast. Low kV contrast-high kV contrast subtraction is done. The final image consists of bony and vessel details. The bony image (step 1) is subtracted

from bony + vessel image (step 2) that gives only vessel details. Hybrid subtraction combines dual energy subtraction and temporal subtraction. It eliminates overlying bone and movement related misregistration issues. It also eliminates artifacts related to soft tissue motion. However, there is an increased noise with higher radiation exposure to the patient.

## **IMAGE QUALITY**

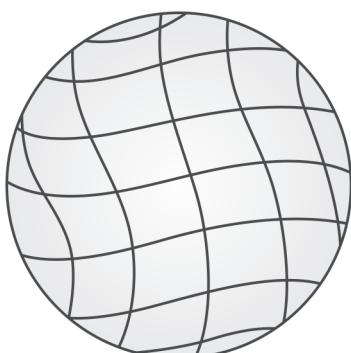
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The fluoroscopic image quality can be described by spatial resolution, noise, veiling glare and geometrical distortion. The typical resolution that may be obtained is 1.2 lp/mm. The Video camera and TV monitor degrade the image quality in fluoroscopy. Contrast resolution of fluoroscopy is low compared with radiography because low exposure levels produce images with relatively low signal-to-noise ratio (SNR). Excellent temporal resolution of fluoroscopy is its strength and its reason for existence.

Noise influences the contrast resolution in fluoroscopy. Noise can be reduced by increasing the input dose rate, increasing the mA, resulting in higher dose to the patient. Alternatively, noise can be reduced by frame averaging. In this, successive images are added together pixel by pixel and the average value is displayed as image. There should be no movement between the frames, otherwise blurring may occur in the image.

Veiling glare is due to scattering in the I<sub>1</sub> tube, which reduces image contrast. It causes the central area of the image brighter than the periphery. Scattering of light at the input phosphor and output window and electron scattering in the I<sub>1</sub> tube, accounts for the above effect. Larger the I<sub>1</sub> tube size, greater the veiling glare.

Geometrical distortion is made up of pincushion distortion and S-type distortion (Fig. 9.13). Pincushion distortion is due to curved geometry of the input screen. The effect of magnetic fields on the electron path causes the S-type distortion. But, both of the distortions are not visible and less significant during fluoroscopy imaging.



**FIG. 9.13:** S-type distortion

## PATIENT RADIATION DOSE

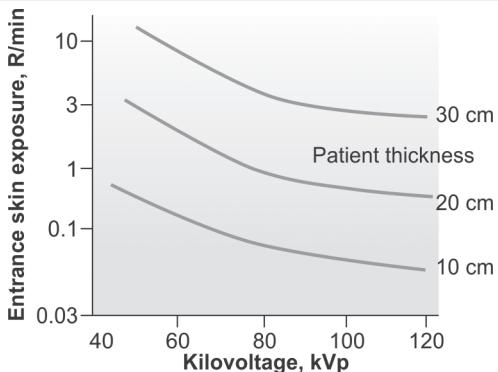
Fluoroscopy imaging time is higher and it may vary from 1–3 minutes for a examination to 15 minutes for a interventional procedure. Therefore, the radiation dose is higher for both patient and the operator. Hence, the skin entrance dose should not exceed 50 mGy/min to the patient. It will vary with kVp and patient thickness. Higher the kVp and lesser the patient thickness, lesser the radiation dose to the patient (Fig. 9.14). A typical fluoroscopy examination may give a dose of 10–30 mGy/min. A procedure with 1 minute screening, at the rate of 50 mGy/min dose rate is equal to 15 pelvis radiographs. Dose reduction to the patient may be achieved by the following:

1. Heavy X-ray beam filtration.
2. Use of low frame rate pulsed fluoroscopy.
3. Use of lower-dose ABC options.
4. Last-frame-hold features.
5. Application of high kV.
6. Frame averaging.
7. Using the largest field of view suitable to a given clinical study.

The X-ray beam passing through the II tube is heavily filtered by 1.5 Al window. The fluoroscopy dose rate depends upon the quantum detection efficiency (QDE) of the II tube, which is kVp dependent. For II tube with CsI output window, the QDE is about 65% and it is maximum at 60 kVp. Hence, use of higher kVp > 60 is recommended. This will reduce patient dose drastically.

The occupational exposure of physicians, nurses, technologists, and other personnel who routinely work in fluoroscopic suites can be high. Hence, they should adhere the following:

1. Lead aprons should be worn when the X-ray beam is on.
2. Portable lead glass shields should be available for additional protection to staff members observing or otherwise participating in the procedure.
3. Reducing total fluoroscopy time is beneficial to everyone.



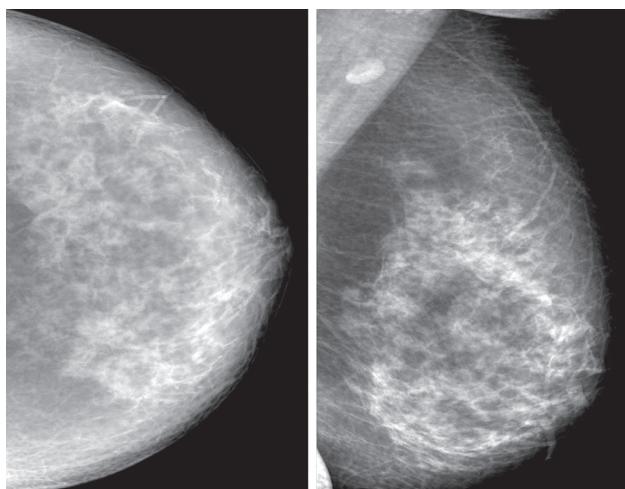
**FIG. 9.14:** Variation of entrance skin exposure with kVp and patient thickness

# 10

# Mammography

The X-ray imaging of breast organ is called mammography. It is basically a soft tissue radiography in which the tissues have similar effective atomic number and similar mass density. However, it will enhance the differential absorption in soft tissues. It helps to detect breast cancer early and reduces the mortality rates for women. Breast cancer is the leading cancer in women and in India one in 22 women may have the chance of getting breast cancer in her life time. There are two types of mammography, namely, (i) diagnostic mammography and (ii) screening mammography. The former is performed on high-risk patients or patients with symptoms. The later is performed in asymptomatic patients. Mammography is a safe and effective procedure for women of age > 40.

Screening mammography normally includes a craniocaudal and mediolateral oblique view of each breast (Fig.10.1). It should reveal



**FIG.10.1:** Breast mammogram: (A) Craniocaudal view,  
(B) Mediolateral oblique view

cancer masses with irregular margins, cluster of microcalcifications and architectural distortions of breast structures. In addition to the normal views, additional examinations, such as magnification view, spot compression view and stereotactic biopsy study may be required. Some time ultrasound imaging, MR imaging and nuclear imaging may enhance the diagnostic accuracy of the breast imaging.

## BREAST ANATOMY

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The breast consists of three tissues, namely, fibrous tissue, glandular tissue, and adipose (fat) tissue. In premenopausal women, the fibrous and glandular tissues are characterized with ducts, glands and connective tissues that are surrounded by fat layer. But, degeneration of fibroglandular tissue and increase of adipose tissue is found in post menopausal breast. In addition, young age breast is dense and hard to image due to glandular tissue, but older age breast is more fatty and easy to image.

The most sensitive tissue to radiation is glandular tissue. Adipose tissue requires lesser radiation exposure. Malignant breast appears as distorted ductal and connective tissues. Majority of breast cancer is due to ductal tissue and related microcalcifications, which appear as small grains of varying sizes ( $\mu\text{m}$ ). The incidence of breast cancer is higher in the upper lateral quadrant of the breast.

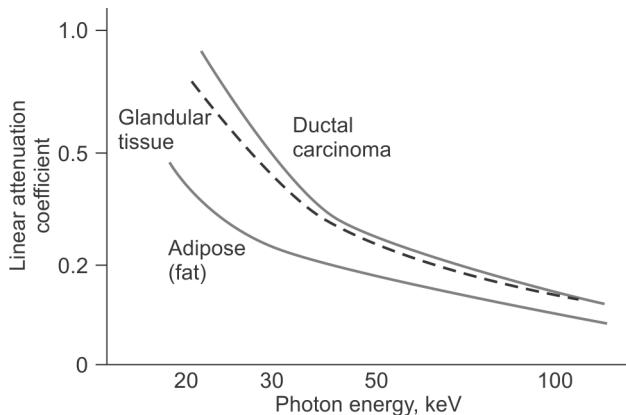
The mass density and atomic number of the breast components are similar. It is necessary to image blood vessels, ducts and microcalcifications as small as  $< \mu\text{m}$ . Conventional radiography is not useful because Compton interaction dominates in soft tissue and differential absorption is very minimal in soft tissue (Fig. 10.2). Hence, low kVp must be used to visualize the above structures (soft tissues). Low energy radiation maximizes photoelectric effect, increase the differential absorption and differentiate the attenuation coefficient of various breast tissues. Though low kV is useful, it will reduce penetration of X-ray beam and hence, high mA is required.

Hence, dedicated mammography unit is essential for good quality breast image with less patient dose.

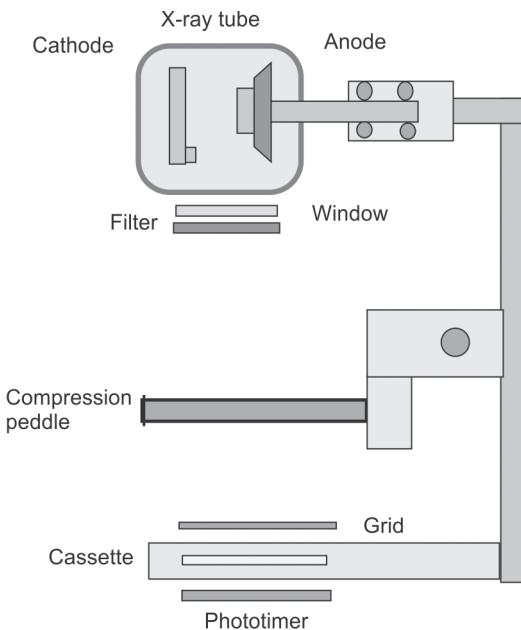
## MAMMOGRAPHY EQUIPMENT

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Mammography equipment comprises of high frequency generator molybdenum target, filter, grid, compression device, and automatic



**FIG. 10.2:** Variation of attenuation coefficient of breast tissues at low photon energy



**FIG. 10.3:** Mammography X-ray equipment

exposure control system (Fig. 10.3). High frequency generator works on single phase and provides rectified smoothed voltage supply to the tube with ripple factor of about 1%. They are smaller in size with good reproducibility and capable of providing up to 600 mAs. It employs small focal spots, low kVp technique, low grid ratio, and special screen-films.

## TARGET

The X-ray tubes are designed with tungsten ( $W$ ,  $Z = 74$ ), molybdenum ( $Mo$ ,  $Z = 42$ ), and rhodium ( $Rh$ ,  $Z = 45$ ) targets. Tungsten target X-ray tube is operated under 30 kVp with 0.5 mm Al filter. It gives bremsstrahlung and 12 keV L-shell characteristic X-rays. Bremsstrahlung X-ray is useful for mammogram and the characteristic X-ray too low for image formation and gives only patient dose. The useful energy range for mammogram is 17–24 keV, whereas tungsten target also have additional energy below and above the useful range.

Molybdenum target is used with 30  $\mu m$  Mo filter or Rh and produce strong K-shell characteristic X-rays. This energy range is more suitable for mammography. The spectral variation between tungsten and molybdenum is due to their atomic number. Rhodium target filtered with rhodium filter (50  $\mu m$ ) gives similar spectrum of molybdenum and gives K-shell characteristic X-rays of energy 20.2 and 22.8 keV. Since the atomic number of rhodium is slightly higher, its bremsstrahlung spectrum is slightly bigger than that of molybdenum.

Molybdenum and rhodium characteristic X-rays arise from K-shell, and the energies lie in the useful range of mammography. Their bremsstrahlung spectrum is smaller than that of tungsten. Hence, most of the X-ray tubes are designed with Mo target with Mo filter or Mo target with Rh filter or Rh target with Rh filter.

The anode is mounted on a molybdenum stem, which is attached to a bearing with rotor and stator assembly. In mammography, in addition to the anode tilt, tube tilt is also incorporated. Hence, the term effective anode angle is used. It is defined as the anode angle relative to the horizontal tube mount. The effective anode angle is about 22–24°, for a source to image distance of 65 cm. This can be achieved either by 0° anode angle and 24° tube tilt or 16° anode tilt and 6° tube tilt.

## FILAMENT

The filament is positioned within a focusing cup, which offer two focal spot sizes, namely, 0.4 mm and 0.1 mm, respectively. The filament types are either double wound filament to increase electron density or flat ribbon filament to give more focused uniform beam or circular filament to create camel hump profile (two focal spots). Focal spot size is very critical in mammography, where high spatial resolution is required.

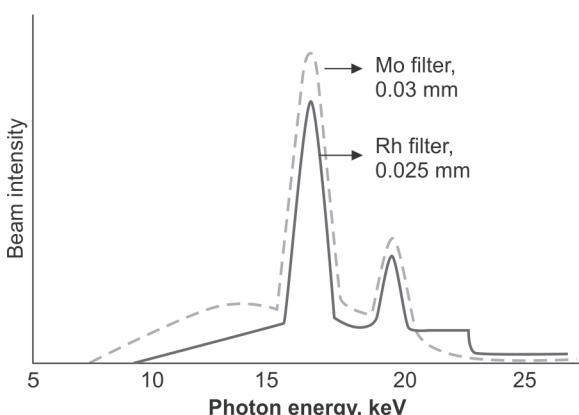
Focal spot size can be selected either by a negative bias on the cathode or by having two separate filaments with two focal spots. Smaller focal spot is required to image micro calcifications. Small focal spot also minimizes geometric blurring and give high spatial resolution. Smaller focal spot size is used with small anode angle, which permit the use of higher mA factors. Usually, the cathode is positioned towards the chest wall. This will make the patient positioning easy and take care of the heel effect.

## FILTERS

Thin beryllium ( $Z = 4$ , 1 mm thick) window or borosilicate glass window is used to reduce inherent filtration, since it offers low attenuation. This will facilitate the inherent filtration in the order of 0.1 mm Al equivalent. In addition, filters are used to remove unwanted high energy bremsstrahlung X-rays. However, the total beam filtration should not be lesser than 0.5 mm Al equivalent. For a tungsten target X-ray tube, Mo or Rh filter is recommended. Molybdenum (0.03 mm) or rhodium filters (0.025 mm) are used for molybdenum anode X-ray tube (Fig.10.4).

In the case of rhodium target X-ray tube, rhodium filter of 0.025 mm is used. It gives high quality X-rays with higher penetration. This combination is suitable for thicker and dense breast imaging. Nowadays bi-angle and double track anode X-ray tubes are used. One track is meant for Mo and other is for Rh.

Generally, filter material is same as target material. This will allow its K characteristic X-rays to reach the breast, and suppress the low



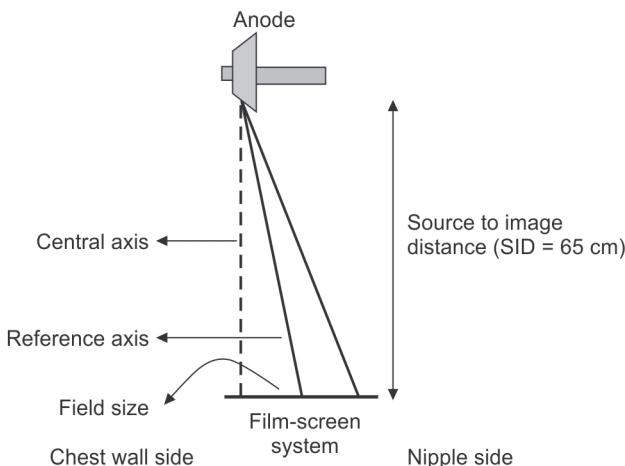
**FIG.10.4:** Bremsstrahlung spectrum of Mo target with Mo and Rh filters, applied voltage, 30 keV

and high energy bremsstrahlung X-rays. These filters remove the bremsstrahlung photons above the K-edge energy.

### HEEL EFFECT

Heel effect causes the X-ray intensity is always higher at the cathode side (Fig. 10.5). The shape of the breast requires higher intensity of radiation near the chest wall, to create uniform exposure to the screen-film. To achieve this, the cathode is positioned towards the chest wall and the anode is towards the nipple. This will permit easy positioning of the patient, since the anode side is bulky and is away from the patient. This will also increase the intensity of radiation near the chest wall, where greater penetration is needed.

However, the effective focal spot size is higher at the chest wall side, which reduces the spatial resolution. Hence, mammography systems use tilted X-ray tubes with long source to image distance (SID) of 60 or 80 cm, to overcome the above problem. Thus, the focal spot is made smaller to image the tissue near the chest wall. The anode is often grounded with zero potential and the cathode is given higher negative potential.



**FIG. 10.5:** Collimation of X-ray beam at 65 cm of SID

### COMPRESSION PEDDLE

Breast compression is required in all mammography examinations, to get good quality images. Compression also ensures that tissues near the chest wall are not underexposed and tissues near the nipple are

not over exposed. Breast compression have the following advantages: (i) it reduces the overlapping of anatomy, spread out the tissue, reduces the thickness of breast and require only low kVp, thereby improving subject contrast, (ii) since compressed breast is thinner, scatter radiation is reduced and contrast resolution is improved, (iii) compression brings the breast closer to the receptor, minimizes magnification and reduces focal spot blurring with lesser radiation dose to the breast, (iv) it also helps to have uniform thickness of breast, to reduce the dynamic range of exposures, and (v) it also immobilizes the breast, minimizes motion related blur and reduces exposure times.

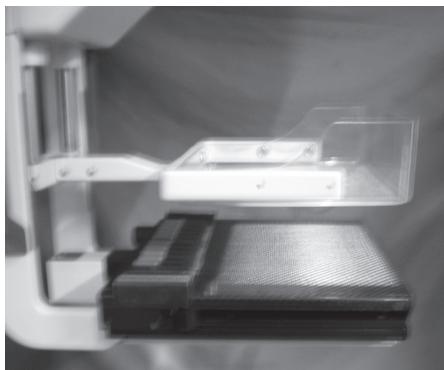
All mammography units are provided with compression device that is parallel to the receptor surface (Fig.10.6). The compression peddle (Lexan plate) is a radio-translucent plate attached to a mechanical assembly. It is flat and parallel to the breast support table. It should match the cassette size of  $18 \times 24$  cm or  $24 \times 30$  cm. It can be operated by foot from either side of the patient. Compression gives a force of 10–20 newton that spreads the breast tissue out.

Spot compression peddle with smaller area (about 5 cm dia.) is also used to provide compression in a particular region of interest in breast. It is very useful to reduce thickness locally, which needs further examination. It will eliminate super imposed anatomy by spreading the tissue further. This makes it easier to detect any pathologic conditions. The optimum degree of compression is unknown; sometimes over compression may leads to patient discomfort.

Overall, breast compression provides improved image quality that helps to detect small, low contrast lesions and high contrast micro calcifications. It also improves spatial resolution and contrast resolution and reduces patient radiation dose.

## GRID

Scattered radiation is an important factor that reduces contrast in mammography. Scatter increases with breast thickness, breast area and independent of kVp. Scatter can be reduced by grid, or air gap



**FIG.10.6:** Breast compression device

technique with breast compression. Hence, special grids are made for mammography, and are placed between the breast and cassette. Generally, moving grids with grid ratio 4:1, focused to the SID is used to improve image contrast. Parallel linear grids with grid ratios 4:1 or 5:1 are also commonly used. Aluminum and carbon fiber are used as interface material for grid. Moving grids have line densities about 30–60 lines per cm, whereas, stationary grid will have about 80 lines per cm. Currently, high transmission cellular grid (HTC) is employed. It is basically crossed grid with grid ratio of 3.8:1, which reduces radiation in two directions. Copper and air are used as grid strip and inter space material.

Use of grid decreases scatter but increases patient dose up to 2–3 times. The use of grid improves the contrast to about 40%. However, compared to the dose, the improvement of contrast is significant. In air gap technique, the breast is positioned away from the cassette and closer to the X-ray tube. This will also reduce the scatter effectively and reduce the patient radiation dose. However, the patient is close to the focal spot and facing higher X-ray output, which contributes radiation dose to the breast.

## COLLIMATION AND HVL

Metal apertures or variable shutters are used to collimate the X-ray beam. It matches the cassette sizes of  $18 \times 24$  cm or  $24 \times 30$  cm. Nowdays automatic collimation systems are used to sense the cassette size. Collimator light and mirror assembly uses low attenuation mirror to reflect the light. The light field should match the radiation field within 2% of SID. The useful X-ray beam must extend to the chest wall edge of the cassette without field cutoff (Fig.10.4). This can be achieved by placing the central axis over the chest wall at the cassette edge. The reference axis that bisects the field, specify the field size. Since the focal spot varies along the cathode-anode direction, it is smaller at the reference axis than at the central axis. The nominal focal spot is specified at the reference axis.

The half value layer (HVL) of the mammographic beam is about 0.3–0.4 mm Al. This depends upon the kVp range and type of target/filter used in the tube. The HVL increases with increase of kVp. The thickness of the compression peddle also influences the HVL value.

## APPLIED VOLTAGE

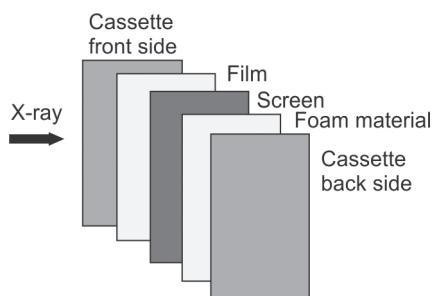
The operating voltage ranges from 25–30 kVp with tube current of 80–100 mAs. The exposure times are about 1–4 s. The tube is operated less than 35 kV, within the space charge effect. Hence, there is no linear relation between filament current and tube current. Feedback circuit is used to adjust the filament current as a function of kV, to deliver a tube current. Three-phase or high frequency generators are used to minimize voltage fluctuations and reduce exposure times. Photo timers and automatic exposure control systems are used to operate correctly at low energies.

## AUTOMATIC EXPOSURE CONTROL

Automatic exposure control (AEC) system employs phototimers to measure the X-ray intensity and quality. Usually, they are kept closer to the image receptor, to minimize the object to image distance (OID), thereby improving spatial resolution. There are two types of AEC available, namely, (i) ionization chamber type and (ii) solid state diode type. Each type will have single or multiple detectors along the chest wall-nipple axis. The detectors are filtered differentially to assess the beam quality. It will also assess the level of compression and type target/filter combination that is employed. In general, thick, dense breast is best imaged with Rh/Rh combination, whereas thin, fatty breast is best imaged with Mo/Mo combination. AEC must be accurate and reproducible with lesser radiation dose. It should hold the optical density within 0.1 OD, when the voltage is varied from 23–32 kVp, for a breast thickness of 2–8 cm, regardless of compression.

## SCREEN-FILM SYSTEM

Mammography is used with screen-film and digital detectors today. The cassette, screens and films are specially made for mammography. Single emulsion film with single back screen is used to avoid light cross over (Fig.10.7). Cubic grain emulsion is used instead of tabular grain. It increases the contrast in the toe region of the characteristic curve, which is useful for mammography.



**FIG. 10.7:** Mammography cassette and screen-film placement

The cassette is made up carbon fiber, to have low attenuation (low Z) and are available in 18 × 24 and 24 × 30 cm sizes. Its back side is made up of low absorbing material when used with AEC. Latching mechanisms are so designed to ensure good screen-film contact.

The X-ray film should have high resolution and small grain size and used along with single intensifying screen. The intensifying screen increases the speed, enhances the contrast and reduces radiation dose to the patient. The screen is placed on the back side of cassette so that X-ray travel through the cassette front side, film and then reach the screen. The emulsion surface of the film must face the screen. This facilitates the X-ray to have interaction at the surface layer of the phosphor, so that the light produced, will travel only a short distance to reach the film. This prevents loss of resolution due to light diffusion in the screen. If the screen is placed between the patient and film, then excess screen blur occurs that will reduce spatial resolution.

Gadolinium oxysulfide activated with terbium is used as screen phosphor, since it emits green light to match the green sensitive film. Mammography films have low film latitude, higher resolution and higher contrast compared to conventional films. The speed of the screen-film systems in mammography ranges from 100–150. It can give satisfactory images with a resolution about 15 lp/mm for a given radiation dose of 15 mR.

## MAGNIFICATION MAMMOGRAPHY

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Mammography can be done on magnification mode to improve visualization of mass margins and fine calcifications. It produces twice the size of the normal image. The purpose of magnification mammography is to examine in depth the small, suspicious lesions or micro calcifications seen in the normal mammogram. In this mode, the breast support platform is raised and kept midway between the focus and cassette. Examination is done with smaller focal spot (0.1 mm) with compression peddle and without the use of grid. A magnification of about 1.5–2 can be achieved.

Magnification has lots of advantages that include (i) increased resolution of the film, (ii) reduction of noise, and (iii) reduction of scattered radiation. Scatter radiation is mainly reduced because of the air gap between the cassette and the magnification stand. It also has disadvantages, which include (i) geometric blurring and (ii) poor spatial resolution on the cathode side due to bigger focal spot size.

Therefore, small focal spot should be used to reduce geometric blurring. Use of small focal spot may limit the current to 25 mA and warrant higher exposure times. Longer exposure time causes not only motion unsharpness and additional radiation dose to the patient. Hence, screen-film systems with higher speed are recommended in magnification mode. This procedure is used only in special circumstances.

## **VIEWING CONDITIONS**

Mammography films are exposed to high optical densities; hence, their viewing conditions should be optimal. View boxes should have a minimum luminance of 3000 cd per m<sup>2</sup>. This is two times higher than the view boxes used in general radiography. Use of magnifying glass may help visualize fine details including microcalcifications. Masking the clear portions of the film and area of the view boxes, may also improve image contrast.

## **DIGITAL MAMMOGRAPHY**

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Instead of screen-film systems, digital detector can be used for mammography. Digital system uses full field of X-rays and hence called full field digital mammography (FFDM). In this, image can be captured indirectly or directly. In an indirect capture, X-rays fall in an scintillator (CsI:TI), which emits light. This light may be detected by a photodiode (a-Si) or a charged coupled device (CCD). In direct capture, the X-rays falls directly on the photo conductor (a-Se), which converts the X-rays into digital signal.

In the indirect system, there is some degree of light spread which degrades the resolution. This is eliminated in the direct capture system. In addition, spatial resolution is limited to pixel size, not to the thickness of photoconductor. In general, FFDM systems are employed in variety of forms, namely, (i) slot scanning with scintilllator with CCD arrays, (ii) flat panel scintillator with a-Si diode array, (iii) flat panel a-Se array, (iv) tiled scintillator with fiber-optic tapers and a CCD arrays, and (v) photostimulable phosphor plates (CR).

### **Slot Scanning with Scintilllator with CCD Arrays**

This system consists of a thallium activated CsI with fiberoptic coupling to a CCD. The X-rays are collimated into a fan beam, matching the detector array. The detector scans laterally across the breast in synchrony

with X-ray beam and forms the image. The detector is 1m wide in the scanning direction and consists of 4 CCDs. But, it is longer enough in length to cover the entire breast in the anterior posterior direction. The total scan time is less and hence motion related unsharpness is less. The tube employs tungsten-rhenium anode with Mo or Rh or Al filter materials. The applied kVp is also higher (31 kVp). The scatter rejection is good and hence no need of grid. It also requires longer breast compression.

#### **Flat Panel Scintillator with a-Si Diode Array**

This system uses a-Si thin film transistor (TFT). The a-Si diode arrays are constructed from a matrix of a-Si TFT deposited on a glass substrate. CsI crystals are deposited as linear columns on the a-Si detector array. When light falls on the diode, it liberates charge, which is read out and digitized.

It consists of  $1920 \times 2304$  detector elements on a  $19.2 \times 23$  cm area with each pixel size of  $100 \mu\text{m}$ . There is close bonding between CsI and photodiode and hence light loss is lesser. Since silicon diode gives strong signal, the detective quantum efficiency is also higher. The digital detector is linear over wide range exposures.

The limitation includes (i) smaller image receptor size and (ii) large pixel size. Up course decreasing the pixel size may increase spatial resolution with increased noise. Hence, there is trade off between resolution and SNR.

#### **Flat Panel a-Se Array**

The a-Se digital detector directly converts X-rays into electronic signal. It is a good photoconductor with high X-ray absorption capability (95%). Its quantum efficiency is higher than that of film-screens and CsI. Even with increasing thickness, it maintains its sharpness. The detector field of view is  $25 \times 29$  cm with pixel size of  $70 \mu\text{m}$ . Its limitation is that it requires large storage space per examination.

#### **Tiled Scintillator with Fiberoptic Tapers and CCD Array**

This system consists of a phosphor (CsI:Tl), a CCD camera and a fiberoptic taper to couple the phosphor and camera. The CCD camera captures the light produced by the phosphor and converts into an electric signal. The CCD array system is  $19 \times 25$  cm size with pixel size of  $40 \mu\text{m}$ . Since the pixel size is small, the spatial resolution is higher (12 lp/mm). However, it has large field size. So, that it require largest image storage matrix. Small field mammography ( $5 \times 5$  or  $8 \times 8$ ) is useful to take

stereotactic localization and biopsy. However, the spatial resolution of small field digital mammography is lesser than that of screen-film systems. However, this method is losing its importance as on date.

## **STEREOTACTIC BIOPSY SYSTEM**

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Mammography can also be used to perform stereotactic biopsy, in order to differentiate benign and malignant tissues. In this, 3D X-ray technique is used to guide a core needle to the biopsy site. Biopsy can be taken either in sitting position or lying position. In the later, the patient is asked to lie down prone on a movable couch. The table has an aperture, through which the breast is positioned in suspended manner. Then, the table is raised and the procedure is done under the couch. The breast is compressed between two plates.

The breast is X-rayed at two different angles, which give 2 digital images in a same screen. The suspicious area is marked on the image. The biopsy system is moved towards the suspicious area with computer generated x, y and z coordinates. A small nick is made at the skin to insert the needle with local anesthesia. The radiologist inserts the needle and advances it to the suspicious lesion. Again X-ray can be performed to verify the needle tip position. Thus, tissue sample is collected and sent for pathologist opinion. Biopsy sample can be taken either with core needle biopsy or vacuum assisted device.

## **BREAST TOMOSYNTHESIS**

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Breast tomosynthesis gives 3D picture of the breast using X-rays. It takes multiple X-ray images of breast from many angles. The breast is positioned similar to normal mammogram, but with little compression. The X-ray tube moves isocentrically in an arc around the breast and takes about 15 images. The computer collects the data and use CT based reconstruction algorithm to construct 3D images, throughout the breast.

Tomosynthesis can be used for screening, diagnostic and stereotactic biopsy. It provides clear, accurate view of the breast. Hence, radiologists can effectively diagnose the size, shape and location of abnormalities. This will benefit young women with dense breast. The advantages includes; (i) early detection of breast cancer, (ii) avoids unnecessary biopsies, (iii) reduce tissue superimposition, (iv) contrast enhancement of lesions, and (v) high depth and contrast resolution.

## RADIATION DOSE

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Mammography involves low kVp X-rays and hence photoelectric absorption mostly accounts for the radiation dose. Low energy scatter also contributes to the radiation dose to the tissue. The radiation dose to the tissue depends on (i) tissue thickness, (ii) X-ray beam quality, and (iii) detector sensitivity.

Tissue compression lowers the radiation dose, by minimizing the photon scatter especially low energy scatter. If efficient screen-film systems are used, then lesser photon density is enough to give optimal image. This will reduce radiation dose to the patient. The nature of anode material and filter also influence the patient dose. Choice of carbon fiber grid and cassette will also reduce radiation dose. To improve the film sensitivity, longer developing time and higher developer temperature are recommended.

Radiation dose, which can induce cancer in the breast, is the main risk in mammography. Glandular tissue is the site for carcinogenesis and hence mean glandular dose (MGD) is considered as dose index, to evaluate risk. The glandular dose varies with depth, beam quality, breast thickness, and optical density of the image. Since the glandular tissue is lying at different depths, the dose also varies and hence measurement is difficult. Therefore, MGD is estimated as follows:

$$\text{MGD} = K \times p \times g$$

where, K is the entrance surface dose (ESD) in air kerma, p and g are the conversion values for a given HVL of the beam. For example, if the HVL of the mammography beam is 0.3 mm, then the corresponding p and g values are 1.10 and 183 mGy/Gy, respectively. The entrance surface dose decreases with increase of kVp but reduces the subject contrast. The acceptable level of entrance surface dose is about 5–6 mGy and the MGD limit is 3 mGy. The entrance surface dose can be measured by keeping a standard breast phantom.

## QUALITY ASSURANCE

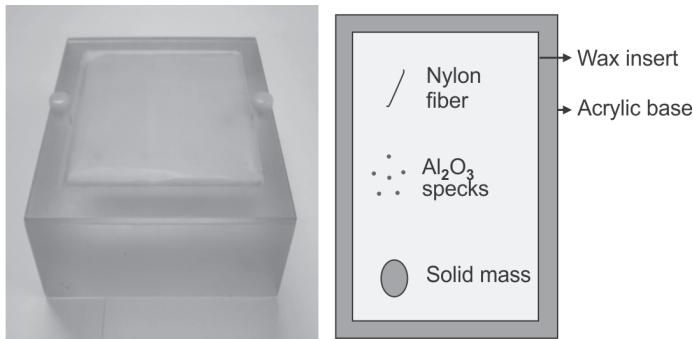
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The objective of quality assurance (QA) in mammography is to obtain high quality images with lesser patient dose. The American College of Radiology (ACR) has recommended QA programs for mammography that includes daily, weekly, monthly, quarterly and annually. Daily QA includes cleaning of darkroom and checking the processor quality. Screens, view boxes and phantom images are to be checked on weekly

basis. Repeat analysis, fixer retention in film is to be carried out quarterly. Darkroom fog, screen-film contact and compression must be checked half yearly.

Mammography phantom simulates the breast for the radiography examination. It is made up of an acrylic block, a wax insert and an acrylic disk attached to the top of the phantom. It represents a standard breast of 4.5 cm thickness composed of 50% adipose and 50% glandular tissue (Fig. 10.7). The wax insert contain (i) 6 cylindrical nylon fibers of decreasing diameter, (ii) 5 calcification groups of decreasing order ( $\text{Al}_2\text{O}_3$  specks) and (iii) 5 solid discs of decreasing diameter, and thickness (Fig. 10.8).

The phantom is placed on the receptor with compression peddle in action. Phantom images are obtained by exposing the phantom with usual clinical exposure factors. The exposure time or mAs reproducibility should be within  $\pm 15\%$ . Identification of smallest size for each category reveals the performance of the mammography unit. For an ideal unit, at least 4 fibers, 3 calcification groups and 3 masses must be seen in the phantom study, which involves an exposure  $< 3$  mGy of MGD.



**FIG.10.8:** (A) Mammography phantom,  
(B) Inner composition of wax insert

The images should be counted from larger to smaller size, with score of 1.0, 0.5 or 0, for correct location and correct orientation. If the entire length of the fiber is seen, then the score is 1, for half fiber the score is 0.5. If less than half of the fiber is seen, then the score is 0. In the case of speck group, if 4 out of 6 specks are seen, then the score is 1, the score is 0.5 for 2 out of 6 specks is visible. If less than 2 out of 6 specks are seen, then the score is 0.

A score of 1 may be given to mass, if the density difference is visible at the correct location, with circular border. A mass with density difference at correct location, with invisible circular border, may have 0.5 score. If there is only a trend is seen in density difference, then the score is 0. Then, using a magnifying glass the image is checked for artifacts. The score should be subtracted for the presence of artifacts that resembles that object. Subtraction can be done only up to next integer, for example 2.5 may be reduced to 2 only for an artifact.

# 11

# Computed Tomography Scanner

## INTRODUCTION

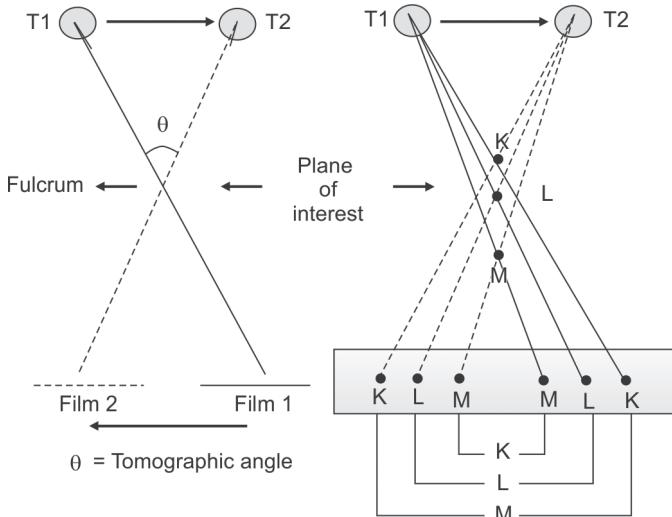
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In conventional radiography, all structures of the patient are exposed to X-rays. Therefore, the image of a particular structure within a patient is obscured by overlying and underlying objects. To overcome this, the image of the overlying and underlying objects may be blurred by moving the X-ray tube and film during exposure, about an axis through the structure of interest. The blurring of undesired images by movement of the X-ray tube and film is referred to as tomography. Tomography refers to slice view or sectional imaging and is usually referred to as body section radiography or linear tomography. It is an imaging technique that produces sectional view of the patient in a plane, parallel to the table top.

The essential parts of a linear tomography system are X-ray tube, X-ray film and a rigid connecting rod that rotate about a fixed fulcrum. If the tube moves in one orientation, the film moves in the opposite direction. The film is placed in a tray under the X-ray table, so that it is free to move without disturbing the patient. The fulcrum is the only point in the system that remains stationary. The amplitude of the tube travel is measured in degrees and is called the tomography angle. The plane of interest within the patient is positioned at the level of the fulcrum, and it is the only plane that remains in sharp focus. All the points above and below this plane are blurred.

In Figure 11.1, the point K is above, and point M is below the focal plane. As the X-ray tube moves, only the image of point L, which is in the focal plane, remains in sharp focus. This is because that L is the only image that moves exactly the same distance as the film. The image of point K moves more than the film, and the image of point M moves less than the film, hence both images are blurred.

The thickness of the section that is in focus depends on the amplitude of the tube travel. The longer the amplitude, thinner the section. The amount of blurring depends on the amplitude of tube travel, distance of the object from the focal plane and film.



**FIG. 11.1:** Tomography principle

## COMPUTED TOMOGRAPHY

Computed tomography (CT) is a special form of tomography in which a computer is used to make a mathematical reconstruction of a tomographic plane or slice. It generates images in transaxial section, i.e., perpendicular to the axis of rotation of the X-ray tube. The computed tomography scanner was invented by Sir Godfrey N Hounsfield in 1970 and was initially named as computerized axial tomography (CAT). The first commercial machine was designed to study the head (1973), and later it was modified to scan any part of the body (1975). Nobel prize was given to the discovery in 1979, for both GN Hounsfield (UK) and Alan M Cormack (USA). In 1963, Alan Cormack built laboratory model for image reconstruction.

The special features of CT image includes (i) images are cross sectional, (ii) eliminates the superimposition of structures, (iii) not influenced by the properties of the neighboring region, (iv) subtle differences in X-ray attenuation is 10 times higher than radiographic image, due to scatter elimination. The minimum contrast in radiography is 2%, whereas it is 0.1–0.3% in CT scan.

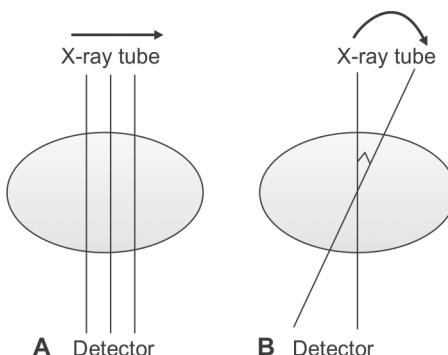
The spatial resolution is about 5–15 lp/cm and the acquisition time is few 100 ms, which is suitable to freeze any physiologic motion. In medicine, CT is used for cancer diagnosis, trauma and osteoporosis. In industry, it is used for nondestructive testing and soil core analysis.

## TERMINOLOGY

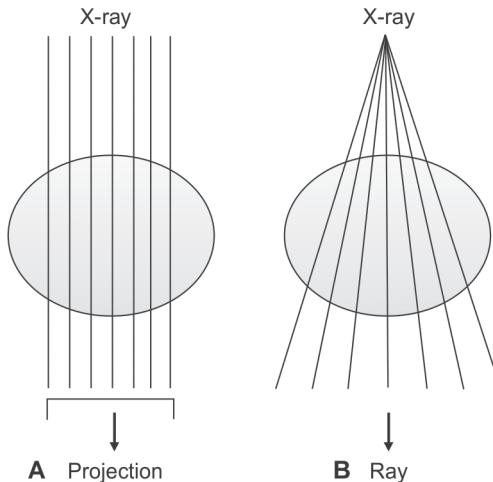
Translation is the linear movement of X-ray tube and detector. Rotation is the rotary movement of X-ray tube and detector (Fig. 11.2). Ray refers single transmission measurement and projection refers series of rays that passes through the patient at the same orientation. There are two projections, namely, (i) parallel beam geometry and (ii) fan beam geometry (Fig. 11.3).

Human body is imagined as a matrix and is divided into number of columns and rows. In general, 512 or 1024 columns and rows are used. Each matrix element is named as picture element (pixel) in a 2-dimensional (2D) concept. Volume element (voxel) represents a volume of tissue in the patient and it is a three dimensional (3D) concept. Each pixel on the monitor display represents a voxel in the patient (Fig.11.4). The field of view (FOV) is the diameter of the area being seen by the X-ray at the isocenter. The relation between the matrix size, pixel size and FOV is given below. If a CT scans a patient with a FOV of 250 mm and has a matrix element of 512, then the pixel size is about 0.5 mm.

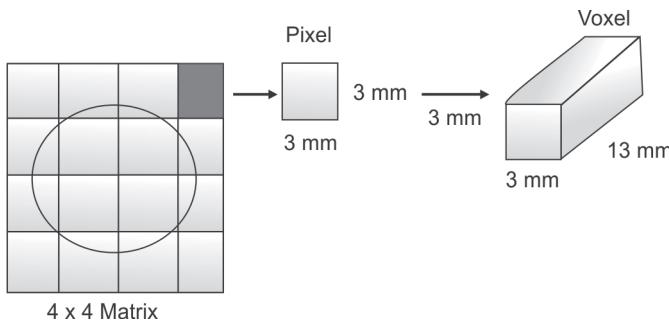
$$\text{Pixel size} = \frac{\text{FOV (250 mm)}}{\text{Matrix size (512)}} \approx 0.5 \text{ mm}$$



**FIG.11.2:** (A) Translation and (B) Rotation



**FIG.11.3:** (A) Parallel beam geometry, and (B) Fan beam geometry

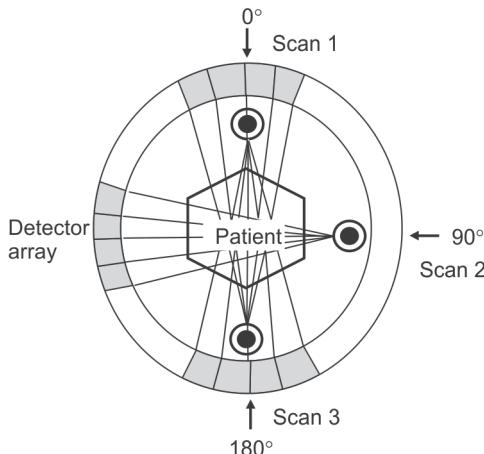


**FIG.11.4:** A pixel representing a voxel in a patient made up of  $4 \times 4$  matrix

## SCANNING PRINCIPLE

The basic principle behind CT is that the internal structure of an object can be reconstructed from multiple projections of the object. To carry out the reconstruction, the linear attenuation coefficient ( $\mu$ ) of the object is considered as base.

A X-ray tube emitting a fan beam from a small focus is coupled to a radiation detector. These two are moved together on a carriage, so that a plane of interest is scanned (Fig.11.5.) The tube potential is about 120–140 kVp and the X-ray beam is pulsed at the rate of 100 pulses per second. The beam is heavily filtered and the detectors are individually collimated and are made either with solid state crystal or with xenon gas ionization chambers. Each detector is calibrated and measures the intensity of the transmitted X-ray beam. Such measured



**FIG. 11.5:** Computed tomography scanning principle

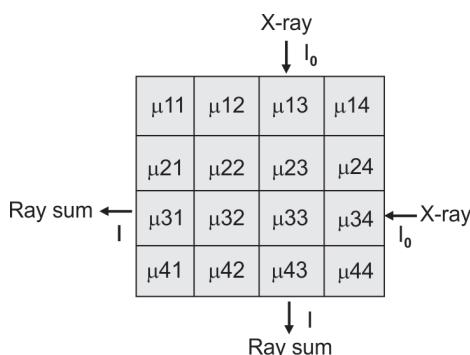
transmissions are called projections. During the scan, about 100,000 measurements are made and recorded in the computer. The above measurements are reconstructed with the help of an algorithm and the final image is displayed on a TV monitor.

An X-ray beam which is transmitted through a voxel, is given by the relation,

$$I = I_0 e^{-\mu x}$$

where,  $I_0$  is the number of initial X-ray photons,  $I$  is the number of transmitted photons,  $e$  is the base of natural logarithm (2.718) and  $\mu$  is the linear attenuation coefficient of the voxel. The value of  $I_0$ ,  $I$  and  $x$  can be measured and the only unknown is  $\mu$ . If the X-ray beam passes through four voxel as shown in the Figure 11.6, then,

$$I = I_0 e^{-(\mu_{13} + \mu_{23} + \mu_{33} + \mu_{43}) x}$$



**FIG. 11.6:** Scanning voxel in different directions

To solve the above equation, additional equations are required. Hence, the voxel are scanned in different directions, which give rise to four more equations. Thus, more than 100,000 equations are generated, by increasing the scan orientations. By solving these equations, the values of  $\mu_{13}$ ,  $\mu_{23}$ ,  $\mu_{33}$ , and  $\mu_{43} \dots \mu_{nn}$  of individual voxel are found.

## CT NUMBER

The attenuation coefficient,  $\mu$  is useful for computation, but not for display of images. Hence, after reconstruction, the attenuation values of each pixel is normalized to that of water, to get integer values as follows:

$$\text{CT Number} = \frac{k \times (\mu_{\text{measured}} - \mu_{\text{water}})}{\mu_{\text{water}}}$$

The attenuation coefficient of water (0.195) is obtained during calibration. When  $k = 1000$ , CT numbers are called Hounsfield Units (HU). CT numbers drive their contrast from physical properties of the tissue that influence Compton scatter; e.g. density, electron density, relative abundance of hydrogen. CT numbers are quantitative, useful for clinical diagnosis such as pulmonary nodule, levels of calcification, bone density, fracture risk, and tumor volume or lesion diameter.

Each pixel is displayed on the monitor as a level of brightness, which correspond to a range of CT numbers from -1000 to +3000. The CT number of -1000 represents air, +3000 represents dense bone and CT number of 0 indicates water. The attenuation coefficient of body organs vary with kV and filtration. Conversion of attenuation coefficient into CT number makes the image, independent of machine parameters. This makes the image more dependent on patient anatomy. However, CT numbers of the human body organs vary due to heterogeneity. Table 11.1 gives the CT numbers of various body organs.

## IMAGE DISPLAY

The reconstructed image is displayed on a cathode ray tube monitor, by allotting shades of gray to each CT number. There are 256 shades of gray in the system and each CT number is allotted one shade of gray. The monitor has a matrix size of  $512 \times 512$  and each pixel represents 12 bits or 4096 gray levels, which is greater than the display range of monitor.

Initially, the average CT number of particular tissue is selected, for example, the abdomen CT number is 20. Then, the computer is instructed

**TABLE 11.1** CT numbers of the human body organs

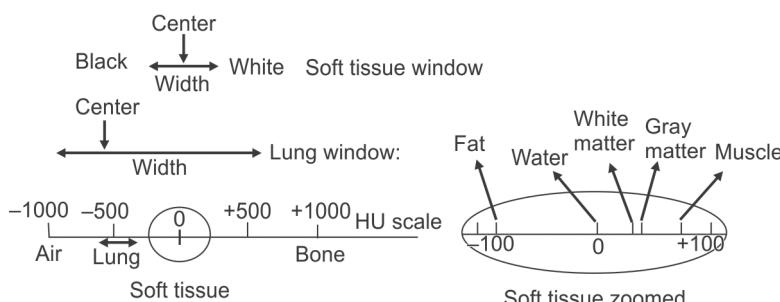
Tissue	CT Number	$\mu @100 \text{ kVp}$
Muscle	50	0.237
White matter	45	0.213
Gray matter	40	0.212
Blood	20	0.208
CSF	15	0.207
Fat	-100	0.185
Lungs	-200	0.093

to assign one shade of gray to each CT number from -108 to +148, so that the center CT number is called window level that determines the brightness level. Each pixel brightness is related to average attenuation coefficient of each tissue voxel.

The range of CT numbers above and below the window level (center number) is called window width, which determines the contrast (Fig. 11.7). A narrow window width provides higher contrast than wide window width. The window level and window width can be set to any desired value of CT number. The window width and window level settings only affect the displayed image, not the reconstructed image data.

The image can be stored permanently in the system or copied in a CD or film. Multiformat cameras (wet/dry) are used to record the images on a film. The typical window level of head, chest (lung) and abdomen (liver) are 40, -500 and 60, respectively. The corresponding window widths are 80, 1500 and 150, respectively.

CT scanning starts with a sinogram (scout image), which is an image of the raw data acquired by CT before reconstruction. It is obtained

**FIG. 11.7:** Window level and window width in a CT scan display

by advancing the patient couch through the gantry with the tube in a fixed position ( $0^\circ$ ). It is not used for clinical purpose and useful to understand the tomographic principle. In regular practice, it is used to identify the upper and lower anatomical borders of the scanning volume.

## EQUIPMENT FOR COMPUTED TOMOGRAPHY

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CT scanners are available as single slice scanner, helical scanner and multislice scanner in the market. In general, all the scanners possess a (i) control console, (ii) computer, (iii) gantry and (iv) couch. Recent developments has brought slip ring technology and multidetector array in day-to-day use. The Z-axis is the gantry rotation axis, longitudinal, and run along foot to head of the patient. The Y axis is perpendicular to the patient in the direction ground to ceiling. The X-axis runs side-side of the patient.

### CONTROL CONSOLE

There are 3 consoles in CT, one for the technologist to operate the imaging system, one for the technologist to post process images and the other for the physician to view images. The operating console is provided with meters, controls for selection of technique factors, movement of gantry and patient couch, computer commands for image reconstruction and transfer, selection of kVp, mA and slice thickness. Usually, 2 monitors are provided, one to annotate patient data (hospital, patient name, age, gender), and identification of image (number, technique, couch position), the other for the operator to view the image.

The physician control console is used to call up and manipulate the image, optimize the diagnostic information, contrast and brightness adjustments, magnification techniques, region of interest (ROI) viewing, use of on line software packages and picture archiving and communication systems (PACS) network.

### COMPUTER

The computer is used to solve more than 2,50,000 equations with the help of microprocessor/array processor and has primary memory. The software includes plot of CT numbers, mean and standard deviation of CT values of ROI, subtraction techniques, planner and volumetric quantitative analysis and reconstruction of images in coronal, sagittal and oblique planes.

## **GANTRY**

CT gantry has the following gadgets: (i) X-ray tube, (ii) collimation and filtration, (iii) detector, and (iv) high voltage generator.

### **X-ray Tube**

The X-ray tube uses intense pulse of X-ray and its performance must be stable. X-ray intensity must not vary over image acquisition cycle and X-ray spectrum should be narrow (does not alter  $\mu$ ). Tubes are operated for prolonged exposure time at high mA (e.g. 90 s, 120 kV, 200 mA). The Heat capacity of the tube is about 4 MJ and heat exchangers are provided to cool oil, air, and to maintain gantry at low temperatures. Usually, scanners are operated at 120 kV, and fixed settings are available between 80–140 kV. The X-ray tube is mounted with its anode-cathode axis, parallel to the axis of rotation, to reduce heel effect.

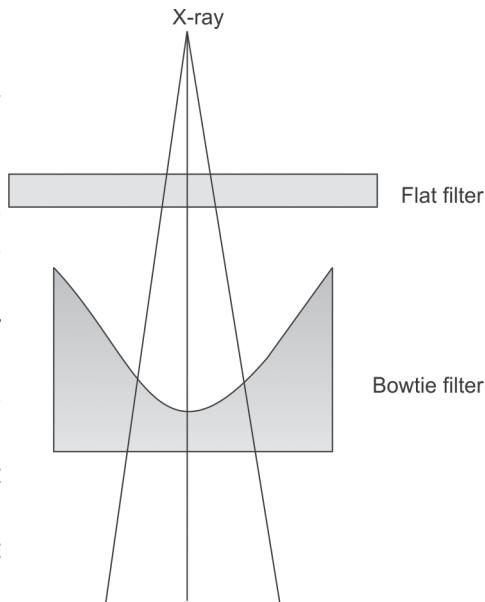
The focal spot determines the amount of information distributed over the detector array. As focal spot increases the information spread over large number of detectors, and limit the resolution. There are two focal spots (0.6–1.6 mm) and high resolution CT uses small focal spot size. The anode is flat for easy heat dissipation and the angle is smaller than normal. The cathode is angled and the focal spot position can be switched magnetically.

The multislice CT tube is large in size, anode disk is larger in diameter and thickness. The anode heat capacity is  $> 8$  MHU and the anode cooling is about 1 MHU. It can be energized up to 60 s continuously, and need high instantaneous power capacity. It requires high speed rotors for best heat dissipation.

### **Collimation and Filtration**

CT scanner uses one or two collimators, which reduces patient dose and improves image contrast, by limiting scatter radiation. Single slice scanner uses 2 collimators as pre and post-patient (predetector) collimation. Multislice scanner uses only single collimator as prepatient collimation. Prepatient collimator limits the area of the patient, determines patient dose (dose profile). Predetector collimator (i) restricts the X-ray beam seen by the detector array, (ii) reduces scatter and improves contrast, and (iii) defines the slice thickness (sensitivity profile). The collimator width is about 50 cm at the isocenter, to cover the full patient and the thickness in the Z-axis is about 1–10 mm.

The X-ray beam is not mono-energetic and hence filters are used to remove low energy photons. Aluminium (2.5 mm) + copper (0.4 mm) are used as filters and modern CT units use 6 mm Al with algorithm help. Since patient cross section is elliptical, X-ray passes through lesser tissue at the periphery than at the centre of the body. Hence, the noise levels may vary, and it is highest at the center and lowest at the periphery. As a result, dose is higher at the periphery than at the center. To solve this issue, bowtie filters of different sizes for head and body scanning are used (Fig. 11.8).



**FIG. 11.8:** Bowtie filter

### **Detectors**

The requirements of CT scan detector are (i) small with good resolution (600–900 for Single slice, width < 1.5 mm), (ii) high detection efficiency, (iii) fast response, negligible after glow, (iv) wide dynamic range and (v) stable noise free response. Currently, two types of detectors are in use, namely, (i) ionization chamber: Xenon gas filled detectors (single slice) and (ii) solid state detector: Scintillation detectors with photo multipliers or photodiodes (Multislice).

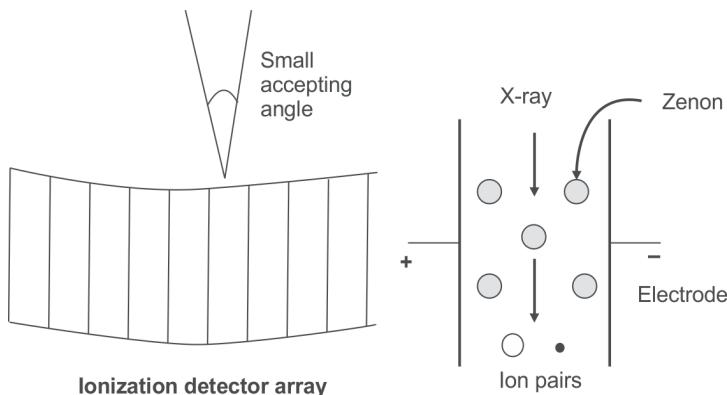
#### *Xenon Gas Detector*

Xenon gas filled ionization chamber detector has high atomic number (54) and its K-shell binding energy is 35 keV. High atomic number facilitates photoelectric absorption as main interaction in the detector. The gas is kept under high pressure of about 2 MPa and its detection efficiency is about 60%. When X-ray falls on the detector, ionization takes place and electric charges are produced (Fig. 11.9). These charges constitute an electric signal that is amplified and digitized. The digitized electronic signal is proportional to the incident X-ray intensity.

Xenon detector is small in size and is elongated in the beam direction. It is aligned with the focal spot, uniform, less dependent on stable

high voltage, and has in-built collimation. The number of chambers are up to 1000 and the detector aperture is 1–2 mm for, 100 mm length. The sensitivity is about 50% of the scintillation detector.

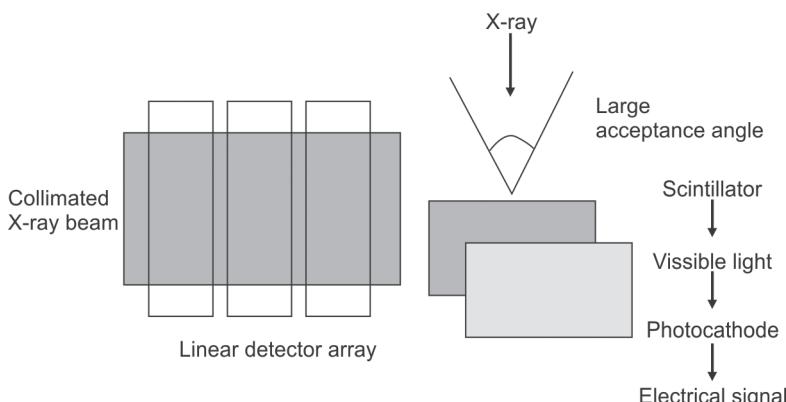
The high pressure xenon gas is kept in long cells and has low detection efficiency due to thin septa. It is highly directional, and it should be positioned in a fixed orientation.



**FIG. 11.9:** Xenon gas ionization chamber detector

#### *Solid State Detector*

Solid state detector consists of a scintillation phosphor coupled to a PMT or photodiode. The size of the detector is about  $1.0 \times 15$  mm and  $1.0 \times 1.5$  mm in the case of multi detector array. When X-ray falls on the detector light is produced, which is detected by the photodiode (Fig.11.10). The photodiode gives the electric signal that is digitized. The digitized electronic signal is proportional to the incident X-ray.



**FIG. 11.10:** CT scan solid state detector working principle

intensity. The detectors are wider (12 mm) than collimated X-ray beam thickness. The phosphors that are used in CT are sodium iodide ( $\text{NaI}:\text{TI}$ ), bismuth germanate ( $\text{Bi}_4\text{Ge}_3\text{O}_{12}$ ), cesium iodide + sodium iodide (UV light), cadmium tungstate ( $\text{CdWO}_4$ ) and yttrium, and gadolinium ceramics.

Solid state detectors has better X-ray absorption efficiency (90%) and large acceptance angle. Since they are closely packed, the detection efficiency is high (90%). The geometric efficiency is less due to gap between detectors to avoid cross talk. It reduces patient dose, provide faster imaging rate, and improves image quality by increasing SNR. It has negligible afterglow, and the stability of output signal depends on high voltage supply.

### **High Voltage Generator**

The high voltage generator is mounted on the gantry, which takes 0.3 s for  $360^\circ$  rotation. The gantry can tilt up to  $30^\circ$  and weighs about 500 kg. The generator is a high frequency generator with capacity of 60 kW. It provides stable tube current and the voltage is controlled by the microprocessor. The generator can give a tube current of about 800 mA @125 kV with pulse duration of 2–4 ms.

### **COUCH**

The couch supports the patient comfortably and it is made up of low z material, e.g. carbon fiber. It is motor driven, for smooth patient position and unaffected by patient weight. It moves longitudinally through gantry aperture, indexed automatically and the tabletop can be removable. In helical CT, the couch motion is defined by the pitch factor.

## **GENERATION OF CT SCANNERS**

A variety of CT geometries have been developed to acquire the X-ray transmission data for image reconstruction. These geometries are commonly called generations. The main objective of different generation is (i) scanning time reduction and (ii) simplification of mechanical motion.

### **FIRST GENERATION**

The first generation CT scanner is a rotate/translate, pencil beam system. It had two X-ray detectors and used parallel ray geometry with  $\text{NaI}$

detector (Fig 11.11A). It is translated linearly to acquire 160 rays across a 24 cm FOV and rotated between translations to acquire 180 projections at 1° interval. It took about 4.5 minutes per scan with 1.5 minutes to reconstruct a slice with a linear measurements of 28,800 rays (160 × 180).

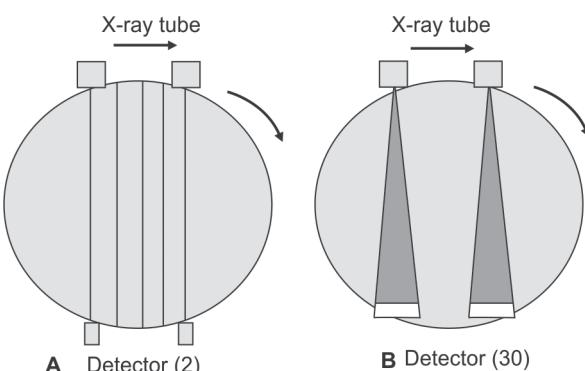
There is a large change in signal due to increased X-ray flux outside of head and hence patient's head is pressed into a flexible membrane surrounded by a water bath. The NaI detector signal decayed slowly, affecting measurements. The advantage of the system is the efficient scatter reduction, best of all scanner generations. The disadvantage includes (i) water bath acted as a bolus and (ii) afterglow of NaI.

## SECOND GENERATION

The second generation CT scanner is also rotate/translate system (11.11B), with narrow beam geometry (10°). Linear array of 30 detectors were used to acquire more data, to improve image quality (600 rays × 540 views = 3,24000). These scanners provided larger rotational increments and faster scans. The shortest scan time was 18 seconds per slice. Narrow fan beam allows more scattered radiation to be detected.

## THIRD GENERATION

The third generation scanner is a rotate/rotate system with wide beam geometry (Fig.11.12A). The number of detectors has increased substantially (> 800 detectors) and the angle of fan beam is increased to cover entire patient. It eliminated the need for translational motion. The X-ray tube and detector array are mechanically joined and



**FIG. 11.11:** (A) First generation CT scan with pencil beam geometry,  
(B) Second generation CT scan with narrow beam geometry

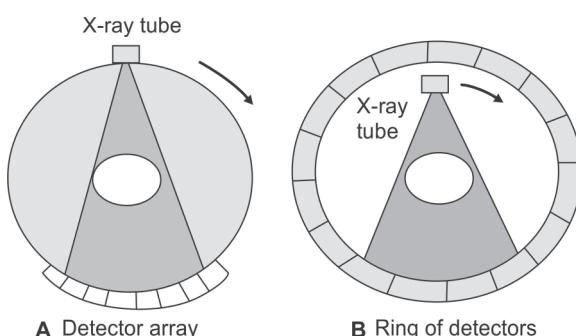
rotate together. Newer systems have scan times of the order of < 0.5 second.

The 3rd generation scanners lead to a situation in which each detector is responsible for the data corresponding to a ring in the image. Any drift in the signal levels of the detectors over time affects the  $\mu$  values that are back projected to produce the CT image, causing ring artifacts.

#### FOURTH GENERATION

The fourth generation scanners 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, very fast scan time is possible. Geometrical misalignment between detector ring radius and X-ray beam origin may be possible. It has inter scan delay times, since the X-ray tube had to return to its starting position (home).

Third generation fan beam geometry has the X-ray tube as the apex of the fan. In the 4th generation, the individual detector is the apex (Fig.11.12B). Though the X-ray tube forms the fan beam, data are processed for fan beam reconstruction with each detector as the vertex of a fan. The rays acquired by each detector are fanned out to different positions of the X-ray source. In the 3rd generation,  $I_t$  and  $I_o$  are measured at the center and at the edge of the detector array. Hence, the gain of the reference detector and the individual detector may not be equal. In the 4th generation, each detector has its own reference detector, and hence the gain of the reference and individual detector is equal.



**FIG.11.12:** (A) Third generation is a source fan and X-ray tube is the apex,  
(B) Fourth generation is a detector fan and detector is the apex

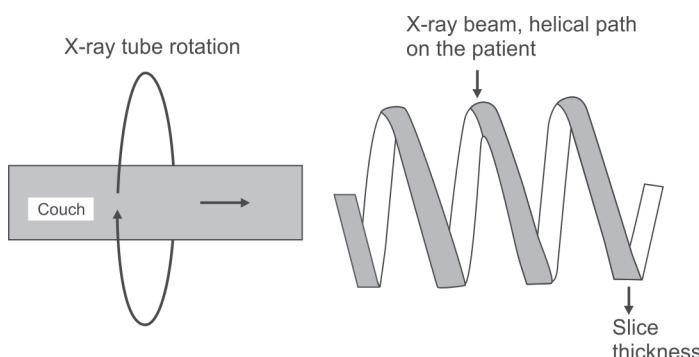
## FIFTH GENERATION

The fifth generation scanner is a stationary/stationary system, developed specifically for cardiac tomography imaging. No conventional X-ray tube is used, instead large arc of tungsten ( $210^\circ$ ) encircles patient and lies directly opposite to the detector ring. It uses an electron gun that deflects and focuses a fast moving electron beam along tungsten target ring in the gantry. Since the detector is also in the form of 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. These scanners are useful in cardiac imaging, pediatric and trauma patients. It can also be used as conventional CT by averaging multiple images with repeat scans.

## SIXTH GENERATION

Third/fourth generation + slip ring technology + helical motion = Sixth generation (1990). Slip ring is a circular contact with sliding brushes and allows the gantry to rotate continuously. It eliminates inertial limitations at the end of each slice and has greater rotational velocities, with shorter scan time. Helical CT scanners acquire data while the table is moving (Fig. 11.13). The total scan time required to image the patient can be much shorter, excluding time required to translate the patient table. It allows the use of less contrast agent and increases patient throughput. In some instances, the entire scan can be done within a single breath-hold of the patient. Raw data from helical scans can be interpolated



**FIG. 11.13:** Helical CT: couch motion and corresponding X-ray beam path

to approximate acquisition of planar reconstruction data. The speed of the couch motion is very important; hence the term pitch is defined.

## **SEVENTH GENERATION**

Seventh generation uses multidetector array (MDA), the collimator spacing is wider and more X-rays are used in producing image data. Opening up the collimator in a single array scanner increases the slice thickness, but reduces spatial resolution in the slice thickness dimension. Hence, slice thickness is controlled by detector size, not by the collimator.

A 4 contiguous 5 mm detector array gives 20 mm collimator spacing. The number of X-rays detected is 4 times higher than that of single array of 5 mm. Further, 10 mm, 15 mm, 20 mm slices may be obtained from the same acquisition. Though it offers flexibility of CT acquisition protocol, the number of parameters have increased. It has better efficiency for patient imaging and detector pitch needs to be defined.

## **IMAGE RECONSTRUCTION**

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When source-detector makes one sweep across the patient, the internal structures of the body attenuate the X-ray beam according their mass density and effective Z. The intensity of radiation detected varies according to this attenuation pattern and an intensity profile or projection is obtained. These projections are not displaced visually, stored in digital form, in the computer. The computer processes the projections that involve super position of each projection to reconstruct an image of the anatomic structures within that slice.

The individual value of the matrix elements ( $\mu$ ) are obtained by solving the simultaneous equations. The matrix of values that are obtained represents cross sectional anatomy. Dedicated array processor is used to do calculation and instantaneous image display.

The image reconstruction algorithms are (i) iterative technique, (ii) back projection and (iii) filtered back projection (FBP). They are basically a mathematical algorithm that takes the projection data and reconstruct cross-sectional CT image. The reconstruction involves millions of data points that may be performed in seconds. Thus, the reconstruction creates an image, which is a map of X-ray attenuation/CT number of the tissue in the plane of interest. Most modern scanners use filtered back projection image reconstruction.

## ITERATIVE METHOD

Iterative method is the original method used by G. Hounsfield. It uses an exact mathematical solution for reconstruction (Fig. 11.14). To obtain a solution, it starts with an assumption that all the pixels have the same value. These assumed values are compared with measured or collected data. Then, corrections are made in assumed values so that the assumed and collected data come closer. This is repeated until all the pixel values are equal to the collected data with reasonable accuracy. It is slow and takes longer computer time and gives imprecise CT values due to rounding errors (e.g.  $0.95 \approx 1.0$ ). The data must be collected before reconstruction.

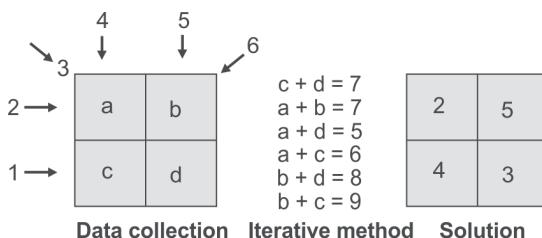


FIG. 11.14: Iterative method

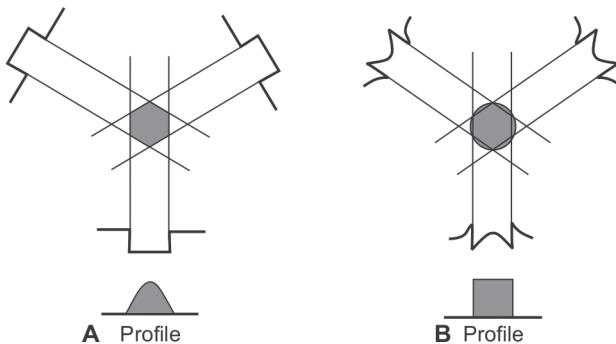
## BACK PROJECTION

Back projection is a mathematical process (algorithm), based on trigonometry, which is designed to emulate the acquisition process in reverse. In this method, required projections of an object are obtained by multiple scans. Then, the projections are back projected to produce the image of that object. All points in the back projected image receive density contribution from neighboring structures and creates noise. Hence, the image quality is very poor and large number of projection are required to improve the image quality.

The disadvantage of the method is the blurred image of the object that has characteristic  $1/r$  blurring and hence not in use today.

## FILTERED BACK PROJECTION

It is similar to back projection, but the raw data are mathematically filtered by a convolution kernel before being back projected (Fig. 11.15). The filtration compensates the sudden density changes that cause image blurring. In this process, the inside margins of dense areas are enhanced, while the central region is repressed. As a result, it reverses the image blurring and restores the true image of the object.



**FIG. 11.15:** (A) Back projection and (B) Filtered back projection of a cylindrical rod object

Convolution is an integral calculus operation and the kernel refers to the shape of the filter function in the spatial domain. Filtering is to be performed in the frequency domain, whereas the data are in spatial domain. Hence, Fourier transform (FT) is used to convert the spatial domain data into frequency domain.

$$P'(x) = 1/FT \{FT [p(x)] * FT [k(x)]\}$$

where,  $p(x)$  is the projection data in spatial domain,  $k(x)$  is the kernel in spatial domain and  $1/FT$  is used to convert the projection data back to the spatial domain.  $P'(x)$  is the back projected data after filtration.

The various type of convolution kernel includes (i) Lak filter, (ii) Shepp-Logan filter, and (iii) Hamming filter. Lak filter increases amplitude linearly as a function of frequency and works well when there is no noise in the data. However, X-ray images involve noise which is more at high frequencies. Shepp-Logan filter incorporates some roll-off at higher frequencies, reducing high-frequency noise in the final CT image. Hamming filter has even more pronounced high-frequency roll-off, with better high-frequency noise suppression.

Bone filter has less high frequency roll off and accentuates higher frequencies in the image at the expense of increased noise, by decreasing the SNR. Bone has very high contrast, so SNR is good, slight decrease in SNR is accepted to get sharper detail in the bone regions of the image.

Use of soft tissue filter gives images with reduced noise and increased SNR. It is used in situations where high spatial resolution is less important than high contrast resolution (e.g. metastatic liver). It gives images with reduced noise and lower spatial resolution. Thus, different kernels are

used for varying clinical applications such as soft tissue imaging or bone imaging.

### **MULTIPLANAR RECONSTRUCTION**

Multiplanar reconstruction (MPR) is a method, in which transverse images are stacked to form a 3D data set, which can be rendered as an image. It is mostly used in multislice spiral CT. The MPR algorithms include (i) maximum intensity projection (MIP), (ii) shaded surface display (SSD), and (iii) shaded volume display (SVD).

The MIP reconstructs an image by selecting the highest value of the pixels along any arbitrary line through the data set and exhibits only those pixels. It is the simplest form of 3D imaging and widely used in CT angiogram. It differentiates vasculature from surrounding tissue, but lacks vessel depth. Small vessels passing obliquely through the voxel may not be imaged, due to partial volume averaging.

The SSD is a computer-aided technique, borrowed from computer aided design and manufacturing applications, used for bone imaging, and virtual colonoscopy. The SVD makes the surface boundaries very distinct and provides an image that appears exact 3D. It is very sensitive to the operator selected pixel range and imaging of actual anatomic structures is very difficult.

### **CT FLUOROSCOPY**

CT fluoroscopy provides an image sequence over the same region of tissue. It is a pseudo-real-time tomography images, and there is no table movement. Images are reconstructed nearly real time during continuous rotation of the X-ray tube. CT images are constantly updated to include the latest projection data at the rate of 6 frames per second. Hence, 6 images are obtained in 1 s for a 360° rotation. The time taken for 1 mage = 1/6 s = 167 ms and the angle is 60°.

After 1 s, the CT scans 60° arc (167 ms), creates a subframe and the old is discarded. Thus, new subframe is added to the old 5 subframes accounting 17% new information and 83% old information. Most recent 6 subframes are summed to produce the CT display.

CT fluoroscopy images has excellent temporal resolution, motion at the image level can be followed in real time. The X-ray tube is operated with a current of 20–50 mA, whereas regular CT uses 150–400 mA. This procedure is commonly used for taking needle biopsies or to drain fluids.

## HELICAL AND MULTISLICE COMPUTED TOMOGRAPHY

### HELICAL COMPUTED TOMOGRAPHY

In helical scanning, the X-ray tube rotates continuously, and the couch moves the patient through the plane of rotating X-ray beam. Using a slip ring technology, the tube is energized continuously, and data are collected continuously. Image can be reconstructed at any desired z-axis position along the patient.

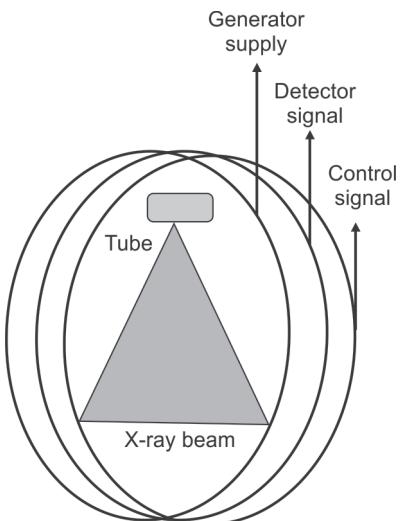
The slip ring was introduced in 1980, and it has three rings. Each ring makes connectivity to the X-ray generator, detector, and control signals. With slip rings arrangement, the X-ray tube can rotate faster (5 s/rot) and move more than 360 degrees.

Slip rings are electromechanical devices that conduct electricity and electrical signals through rings and brushes from a rotating surface onto a fixed surface (Fig. 11.16). The power is transmitted through stationary rings. There are brushes (silver graphite alloy) that transmit power to the gantry, glide in contact grooves on the stationary slip ring. Slip ring is used in spiral CT or multislice CT, allows the gantry to rotate continuously. It eliminates the need of electrical cables, which restrict continuous rotation.

Helical or spiral or volume scanning is a procedure in which both X-ray tube and couch move, and it came to clinical use in 1990. It has continuous acquisition of data, while couch moves and gives complete volume data in a single exposure. The X-ray source moves in a helical path around the patient, and hence it is called as helical CT.

#### X-ray Tube and Generator

Helical CT requires higher X-ray power, due to continuous large volume data acquisition. The X-ray tubes are used for longer exposure > 90 s and require good performance. Smaller focal spot is required for scanning thin sections with high resolution. It is provided with flying focal spot in which rapid deflection for each projection is possible, to



**FIG. 11.16:** Slip ring technology

increase the resolution. The anode cooling rate must be high and liquid metal bearings are used to withstand heat.

The tube current can be altered during the course of helical scan sequence, to suit individual body composition. It provides reduction of mean mAs per rotation of the order of 15–55%. The beam filtration normally used is 3 mm, Al + bow tie filter is usually made up of low Z material (teflon or copper). There is a pre- and post-patient collimator of 100 µm thick, made up of tantalum. The gantry mechanical design must be precise to make uniform motion. The specification for a typical helical CT is given in Table 11.2.

**TABLE 11.2** Specification for helical CT

X-ray tube	Heat storage	6.3 MHU
Generator	Focal spot (mm)	0.7,0.9,1.2
	Anode angle	7°
	Power	60 kW
	kV ranges	80,100,120,140
	mA (max)	440
	Acquisition time (max)	120 s

**TABLE 11.3** Comparison of Xenon gas and ceramic scintillation detectors

Property	Xenon gas	Ceramic scintillators
Efficiency	74% @ 120 kV	90% @ 120 kV
	71% @ 140 kV	85% @ 140 kV
Absorption	< 50% @ 120 kV	95% @ 120 kV
After glow	0.4% @ 100 ms	< 0.1% @ 100 ms

### Detector

In helical CT, efficient detector can reduce X-ray tube load. Usually, solid state scintillation or xenon gas ion chambers are used as detectors. Currently, ceramic scintillation phosphors are in use, which gives an absorption efficiency of about 99%. Table 11.3. gives the comparison of xenon gas and ceramic scintillation detectors with their inherent properties. Few commonly used ceramic scintillation phosphors are: (i) lutetium orthosilicate ( $\text{Lu}_2\text{O}_3:\text{Ce}$ ), (ii) gadolinium orthosilicate ( $\text{Gd}_2\text{O}_3:\text{Ce}$ ), and (iii) yttrium aluminum perovskite ( $\text{YALO}_3:\text{Ce}$ ).

The table movement in a helical scan is a variable one and it can be moved either fast or slow. Hence, to identify the nature of table motion, the term pitch is used and it is defined as follows:

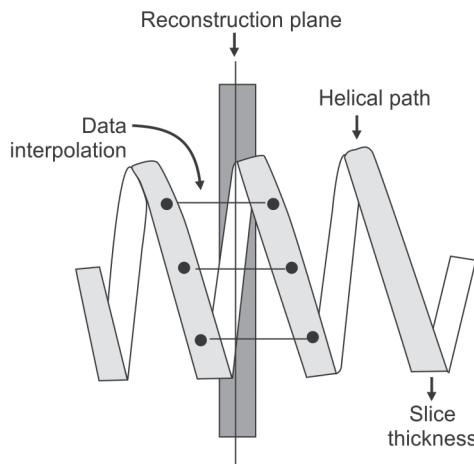
$$\text{Pitch} = \frac{\text{Tabletop movement per rotation}}{\text{Slice thickness}}$$

If a table moves 10 mm in one rotation of a gantry in 1 s, to make a 10 mm slice thickness of a patient, then

$$\text{Pitch} = \frac{10 \text{ mm/s}}{10 \text{ mm}} = 1$$

### Interpolation

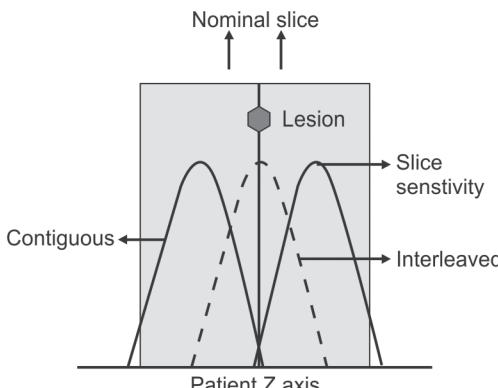
CT reconstruction algorithm assumes that the X-ray source path is circular, not helical around the patient. But, in a helical scan, the X-ray beam path is helical in nature and gives only helical data set. Hence, before the actual CT reconstruction, the helical data set has to be interpolated into a series of planar image sets. Then, with interpolated data set, CT images can be reconstructed at any position along the length of the scan (Fig. 11.17).



**FIG. 11.17:** Helical data set and interpolation for reconstruction

Interpolation is a weighted average of the data from either side of the reconstruction plane. Slightly different weighing factors are used for each projection angle. Interleaved reconstruction allows the placement of additional images along the patient, so that the clinical examination

is almost uniformly sensitive to subtle abnormalities (Fig. 11.18). This involves no additional dose to the patient, but additional time is required to reconstruct the images. However, slice thickness still decides the spatial resolution along the long axis of the patient.



**FIG. 11.18:** Interleaved reconstruction predicts lesion inbetween the scanned slices

The advantages of a helical scan is the scan speed and patient throughput: e.g. a chest scan with 10 mm slice can be done in single breath hold in 15–20 s (couch motion 10 mm/s, pitch 1.5) and avoids slice misregistration. Use of higher pitch reduces patient dose and exposure time.

### MULTISLICE COMPUTED TOMOGRAPHY

The multislice CT (MSCT) was started in 1992 with 2 parallel bank of detectors, that gives 2 slices. In 1998, solid state multi row detector was introduced to make 4 slices in each rotation. Multislice CT generally uses third generation CT with helical scanning and low voltage slip rings.

The special features of MSCT is faster rotation subsecond times (0.5–0.8 s), that reduces the examination time. The image quality is similar to that of single slice scanners. But, it is different in dose, pitch, image artifacts, and method of image reconstruction. To scan longer anatomic area, more than 4 slices are required and hence gantry speed has to be increased. Compared to 1 s single slice scanner, MSCT perform 0.5 s rotation with simultaneous acquisition of 4 slices. Thus, it gives 8 times higher performance than single section CT, for same scanning time. Recent advances has brought scanners with 4, 8, 16, 64 slices in practice.

### Multidetector Array

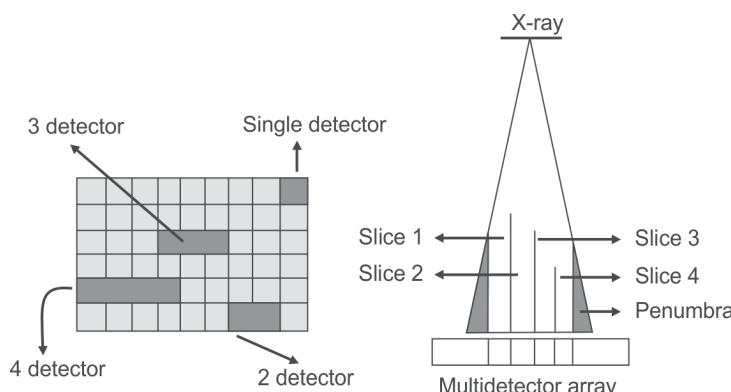
Multidetector array is a combination of several linear arrays. It is an assembly of multiple solid state detector array modules. In this, the X-ray tube is directed at multiple rows of detectors along the longitudinal (Z) axis. Each row has hundreds of separate detector elements (Fig. 11.19). Combined rows constitute the 2-D curved array that has detector elements greater than 35000. It has separate data acquisition channels for each detector element and can generate multiple channels (4,8,16, 64) of spatial data. In a single slice scanner, the detectors are wide (15 mm) and collimator determines (adjustment) slice thickness of 1–13 mm. In a MSCT, the individual detector elements along the z-axis are summed, to get several slice thickness. Thus, the slice width is determined by the detector not by the collimator. The detector dimensions are always referred at the isocenter.

It is available in 2 commercial designs, namely, (i) adaptive array detector and (ii) linear or matrix detector (Fig. 11.20). In the adaptive array design, the detector width is unequal, e.g. it may have detector width as 1.0, 1.5, 2.5 and 5 mm from the center to edge. In the linear array, the width is equal, e.g. it may have 1.25 mm throughout the dimension with 16 detector module. If electronics are available for 4 detector array channels, then combination of  $4 \times 1.25$ ,  $4 \times 2.5$ ,  $4 \times 3.75$ , and  $4 \times 5.00$  mm may give different slice thickness in MSCT. Generally, 8, 16, and 64 slices CT systems have the following provision with 32 mm detector in the z-axis.

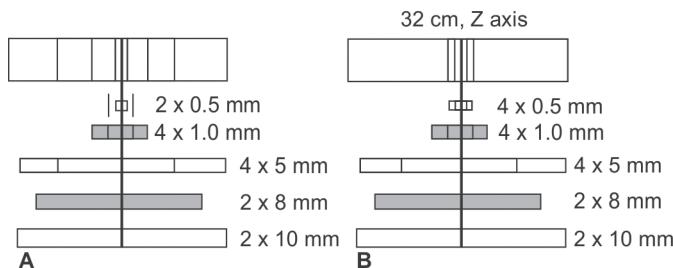
8 slice, 32 mm :  $8 \times 0.5$  mm,  $8 \times 1$  mm,  $8 \times 2$  mm,  $8 \times 4$  mm

16 slice, 32 mm :  $16 \times 0.5$  mm,  $16 \times 1.0$  mm,  $16 \times 2$  mm

64 slice, 32 mm :  $64 \times 0.5$  mm,  $32 \times 1$  mm



**FIG. 11.19:** Multidetector array and slice selection



**FIG. 11.20:** (A) Adaptive array: 4 slice, unequal width, and center slice 2 x 0.5 mm.  
 (B) Linear array: 4 slice, equal width, and center slice 4 x 0.5 mm

Usually, MSCT with multidetector array employs third generation CT with 16 detectors. If 750 detector elements are there in each array, then 12,000 ( $16 \times 750$ ) detector elements are required for a  $60^\circ$  fan beam geometry. Whereas 4th generation CT require much more detector elements to cover the entire detector ring of  $360^\circ$ .

MSCT can be used for conventional axial scanning and helical scanning. The width of the 2 center detector arrays dictates the slice thickness. To keep the sensitivity profile of the each detector similar, collimator is adjusted to keep the penumbra outside the detector. The radiation dose is higher with reduced artifacts.

### Pitch

In MSCT, the pitch influences radiation dose, image quality and scan time. The pitches are (i) collimator pitch and (ii) detector pitch. The collimator pitch is given by the relation:

$$\text{Collimator pitch} = \frac{\text{Table movement per } 360^\circ \text{ gantry rotation}}{\text{Collimator width at isocenter}}$$

Pitch = 1, refers normal scanning, pitch = 0.75 refers over scanning, and the table motion is slow. This gives increased image quality with increased radiation dose. If the pitch = 1.5, it is said to be under scanning and the table motion is faster with lesser patient motion. It consumes smaller volume of contrast agent and good for pediatric patients. Higher pitch is always advocated for pediatric CT protocol. The detector pitch is defined as follows:

$$\text{Detector pitch} = \frac{\text{Table movement per } 360^\circ \text{ rotation of gantry}}{\text{Detector width}}$$

The collimator pitch and the detector pitch of a given CT is related as follows:

$$\text{Collimator pitch} = \text{Detector pitch} \div N$$

where, N is the number of detector arrays in the multislice CT. The MSCT scanners with 4 detector arrays uses 3–6 as detector pitch values, e.g. a pitch of 6 corresponds to 1.5 for conventional scanner, since  $N = 4$ .

The clinical advantages of MSCT includes (i) increased speed, (ii) shorter acquisition times with improved temporal resolution (lesser motion artifacts), (iii) high axial resolution with thinner slices in the longitudinal (z) axis, (iv) retrospective creation of thinner or thicker sections from raw data, (v) accurate anatomical 3D reconstruction especially in angiography and virtual endoscopy with lesser helical artifacts, (vi) increased volume coverage per unit time, (vii) reduced partial volume artifacts and noise, (viii) detailed multi-planar reconstruction images, and (ix) use of lesser contrast materials and delivery of contrast material at faster rate, thus increasing contrast enhancement in images.

## **CT IMAGE QUALITY, ARTIFACTS AND RADIATION DOSE**

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### **IMAGE QUALITY**

The CT image quality is mainly controlled by four factors, namely, (i) spatial resolution, (ii) noise, and (iii) contrast.

#### **Spatial Resolution**

Spatial resolution is the ability to discriminate adjacent objects, and expressed in lp/cm. It depends on pixel size, focal spot size, detector size, choice of image reconstruction filter, and sampling frequency. The range of spatial resolution in CT is 5–15 lp/cm.

#### **Noise**

Noise is the fundamental limit to the quality of CT image. It reduces contrast resolution of small objects and worsens the spatial resolution of low contrast objects. There are three sources of noise, namely, (i) quantum noise, (ii) electronic noise and (iii) structural noise.

The quantum noise is the major contributor and it is caused by random variations in the number of photons detected. It depends on the selection

of scanning factors; mA, scan time, and slice width. Increase of mA and slice width reduces noise, but reduces spatial resolution and increases partial volume effect. Increase of kV, ensures penetration and higher number of photons, and reduces noise, but reduction in contrast.

Reduced FOV has fewer number of photons per pixel, which increases noise. In a single slice scanner, pitch does not affect noise, but reduces dose due to interpolation between rotations, and make use of detected photons. In a multislice scanner, increase of pitch increases noise, because of the cone beam effect, complex reconstruction, and data are used from all detector rows for each projection angle. Noise is apparent as window width is reduced, but reveals low contrast detail. The typical noise level in CT scan is 3 HU, and smaller the patient lesser the noise. Use of detail filter increases the noise and use of soft tissue filter decreases the noise.

### **Contrast**

CT contrast is the difference in the HU values between tissues. Contrast increases as tube voltage decreases, not affected by tube current or scan times. It increases with addition of contrast medium (iodine) and reduces as the photon energy increases. Higher photon energy reduces contrast much more in high Z, than soft tissue lesions. Lesion that differs by 5 HU ( $0.5\% \mu$ ) from its surroundings can be detected in CT scan. But it is 3–5% difference for screen/film radiography. Thus, CT gives better contrast than screen/film radiography. Narrow window level offers high contrast to the image.

## **ARTIFACTS**

Artifacts are the ones which not only degrade image quality, but also leads to wrong or miss diagnosis. The typical artifacts in CT images are (i) motion artifacts, (ii) streak artifacts, (iii) beam-hardening artifacts, (iv) ring artifacts, (v) partial volume artifacts, (vi) spiral artifacts and (vii) rod artifacts.

### **Motion Artifacts**

The patient motion is random or unpredictable during CT scanning (e.g. patient's sneeze). The image will display the object motion as a streak in the direction of motion. It depends on the density of the object in motion and densities much different from the surroundings produces more intense motion artifacts. Motion artifacts are more prevalent in scan times of 0.5–2 s, due to involuntary and voluntary patient motion.

Structures move from one voxel to another, and introduce errors in reconstruction. Motion artifacts appear as double images or image ghosting that may leads to rescanning.

### **Streak Artifacts**

Streak artifacts are due to absence of transmitted X-rays to the detector and it appears as dark and light lines. The sources of streak artifacts are high density material such as metal implants, dental amalgam, and shotgun pellet, etc. Streak artifacts increase with motion and metal correction algorithms are provided in some CT scanners.

### **Beam Hardening Artifacts**

Beam hardening artifacts or cupping artifacts are caused by polychromatic nature of X-ray beam (25–120 keV). As the beam passes through the patient, low energy is absorbed, and the mean energy increases. As a result, the beam become hardened, that causes underestimation of  $\mu$  and HU. It is possible to minimize the beam hardening effect by a suitable correction algorithm. Beam hardening artifacts are marked at high contrast interfaces (bone), e.g. petrous bones in the head.

### **Ring Artifacts**

Ring artifact is the result of mis-calibration or failure of one detector in rotate-rotate system of 3rd generation CT scanner. Due to the failure of a particular detector, incorrect data in every projection will appear as a ring in the image. Radius of the ring is determined by the position of the detector in the array and virtually disappeared in contemporary CT units.

### **Partial Volume Artifacts**

Partial volume artifact is result of averaging the linear attenuation coefficient in a given voxel that is heterogeneous in composition (e.g. presence of bone and soft tissue). This artifact increases with increasing pixel size and slice thickness. It is pronounced for softly rounded structures that are parallel to the CT slice. For example, when cranium shares few number of voxel with brain tissue, there is loss of details of brain parenchyma. Use of thinner slices and helical scan with interleaved reconstructions will reduce partial volume artifacts (e.g. 5 mm slices at the interval of every 2.5 mm).

### **Spiral Artifacts**

A helical scan gives an image similar to partial volume averaging (PVA). In one direction, the PVA is determined by collimation, and in the other

direction, it is by collimation and table increment per rotation. As a result, instead of a perfect circle, an ellipse like reconstruction is done. It is more apparent for large beam top angle, and large pitch, e.g. spiral scans of top of the brain: it is seen as two crescent shaped bands of increased density along the skull-brain interface, which mimics a subdural hematoma. These bands rotate around the brain as the X-ray tube rotates around the patient.

### **Rod Artifacts**

Rod artifacts are apparent when high contrast objects vary in shape/position. For example, if a cylindrical object angulated with respect to the scan plane, then every projection locates the cylinder at a different position. As a result, the cylinder would appear as ellipse, without the table motion. With table motion, the ellipse gets distorted, extending to surrounding tissues and commonly seen in the liver/rib area.

## **RADIATION DOSE**

The radiation dose in CT is unique in 3 ways: (i) the volume of tissue irradiated by the primary X-ray is small, (ii) the volume of tissue is irradiated in almost all angles (even distribution of dose), and (iii) radiation dose to the slice volume is higher, due to techniques used (kV, mAs). For example, a thoracic CT is done with 120 kV, and 200 mAs, whereas a PA chest radiograph is done with 120 kV, and 5 mAs.

In CT, the scattered radiation (Compton interaction) increases the dose, than the primary beam and tissues beyond the section are exposed to radiation. The dose profile is not uniform along the patient axis and doses at the patient surface may be higher than dose at the center of the patient. The surface–center dose ratio may be 1:1 in a head scan and 2:1 in a body scan. CT section dose profile is not perfectly square but has tails that extend beyond the section edges, due to scatter. Dose increases as the number of slice increases.

### **Multiple Scan Average Dose**

The multislice average dose (MSAD) is defined as the average dose, at a particular depth from the surface, resulting from a large series of CT slices (FDA, USA). MSAD is estimated by measuring (i) CT Dose Index and (ii) dose-length product.

### **CT Dose Index**

The CT dose index (CTDI) is the radiation dose at the center or peripheral point on a head or body phantom from a single scan, from 7 CT slices

in both directions. It is the integral of the axial dose profile for a single CT slice divided by the slice thickness. Dose is integrated over the 14 slices by a 100 mm ion chamber. CTDI are always stated for 100 mAs value. In general, head phantom (16 cm diameter) or body phantom (32 cm diameter) made up of PMMA material is used with pencil type ionization chamber, to measure CTDI. Measurements are made at the peripheral ( $\text{CTDI}_{\text{peri}}$ ) and at the center ( $\text{CTDI}_{\text{cent}}$ ).

CTDI increases with tube voltage and CTDI of body scans are lower than head scans, due to greater attenuation in the body. A weighted CTDI is expressed as  $\text{CTDI}_w$ , which is given by the relation.

$$\text{CTDI}_w = \frac{2}{3} (\text{CTDI}_{\text{peri}}) + \frac{1}{3} (\text{CTDI}_{\text{cent}})$$

### Dose Length Product

The  $\text{CTDI}_w$  does not quantify the patient risk, since slice thickness, number of slices, and organ sensitivity are not accounted. Hence, a term dose length product (DLP) is defined, which accounts the radiation risk.

$$\text{DLP} = \Sigma \text{CTDI}_w \times T \times N \times C$$

where,  $N$  is the number of slices,  $T$  is the slice thickness (cm) and  $C$  is the exposure, in mAs. DLP is proportional to the total dose (energy) imparted to the patient. It can be used as an indicator of the relative risk in CT.

### Effective Dose

CT scan exposure will not produce deterministic effect, but induction of cancer is the risk. This depends on dose and radio sensitivity of the organ/tissue. The radiation dose increases with tube current and scan time. To account the radiation risk as well as the radio-sensitivity of the tissue, the parameter effective dose ( $H_e$ ) is defined:

$$H_e = E_{\text{DLP}} \times \text{DLP}$$

where,  $E_{\text{DLP}}$  is the tissue specific normalized effective dose in mSv/mGy.cm. The typical CT dose index, dose length product and effective dose values are given in Table 11.4.

Effective doses for head CT in infants/children are 4 times higher than that of adults. Body CT scan doses in infants/children are double the dose of adults. Children doses are higher due to smaller organ size (dose = energy/mass). CT accounts only 8% of all examinations, but contributes 48% of total patient dose from X-rays.

**TABLE 11.4** Typical values of CT dose index, dose length product, and effective dose per slice, for various body site

Tissue	CTDI <sub>w</sub> (mGy)	DLP (mGy-cm)	Effective dose, H <sub>e</sub> (mSv)
Head	60	1050	2.4
Chest	30	650	1.1
Abdomen	35	800	1.2
Pelvis	35	600	1.1

Patient effective dose is proportional to the total energy deposited and the dose is proportional to the tube current and scan time. Increasing kVp from 80 to 140 increases the dose by a factor of 5. Effective dose is also proportional to the product of the slice thickness and number of slices. In helical CT, the dose is inversely proportional to pitch. A pitch of 1.5, will reduce the dose to 67% compared to that of pitch 1. Similarly, a pitch of 2, reduce the dose by 50%. Pre-and postcontrast CT scans of the abdomen doubles the patient dose. Skin doses are very high in CT fluoroscopy.

# 12 Gamma Imaging

## RADIOACTIVITY

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Radioactivity, a nuclear phenomenon, was first discovered by Henri Becquerel in 1896. Becquerel left some uranium salt on a photographic plate, wrapped in a black paper and was lying in a darkroom. When he developed the photographic plate, he found that the plate was affected. Later, he repeated the experiment with other salts of uranium and he concluded that uranium and its salts emit invisible radiations, which can pass through paper, wood, glass, etc., and affect the photographic plate. These radiations were found to have alpha ( $\alpha$ ), beta ( $\beta$ ) particles, and gamma ( $\gamma$ ) rays.

Radioactivity is the process by which a nuclei undergo disintegration and emits either alpha or beta and gamma radiations. During the radioactive process, the atom changes its atomic number and chemical identity. An atom, with unstable nuclei and perform radioactivity is called radioisotope. The initial atom that undergoes disintegration is called parent and the end-product is called daughter. Radioactivity may be classified as natural and artificial. The phenomenon of spontaneous emission of rays, such as  $\alpha$ ,  $\beta$ , and  $\gamma$  by heavy elements having atomic number greater than 82 is called natural radioactivity, e.g. radium-226 and potassium-40.

Artificial or induced radioactivity was discovered by Curie and Joliet in 1934, when they were studying the disintegration of light elements by  $\alpha$  particles. They found that when light elements, such as boron and aluminum were bombarded with  $\alpha$  particles, an unstable nucleus was formed and this nucleus disintegrated spontaneously. The artificial radioactive substance emits electrons, neutrons, positrons or  $\gamma$  rays. They follow the same laws of decay as natural radioactivity, e.g. cobalt-60 and phosphorus-32.

## NUCLEAR FORCES AND STABILITY

There are two forces acting inside the nucleus, namely, a weak repulsive, force between protons and a strong exchange force between neutrons. The neutrons exchange pions between them. The strong exchange force keep the nucleus together and maintain the nucleus to a short distance  $10^{-14}$  m. If a nucleus wants to be stable, it should overcome the repulsive forces. This means that the neutrons (N) must be higher than protons (Z) or N/Z ratio must be higher. The above ratio is equal to 1 for low atomic and 1.5 for higher atomic elements. Nuclei with odd number of neutrons and odd number of protons are unstable. But, nuclei with even number neutrons and even number of protons are more stable. Nuclear instability occurs whenever there is excess neutron or excess protons. An unstable nucleus attain stability by performing radioactivity.

## RADIOACTIVE DISINTEGRATION

Rutherford and Soddy found that the rate at which a particular radioactive material disintegrates was independent of physical and chemical conditions. Their law states that the number of atoms that disintegrates in unit time is proportional to the number of radioactive atoms present at that instant. Let N is the number of atoms at a particular time t, and dN atoms disintegrate in a time dt, then

$$= \frac{dN}{dt} = \propto N$$

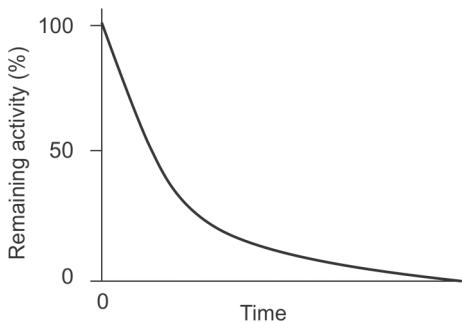
$$\text{or } = - \frac{dN}{dt} = \lambda N$$

where,  $\lambda$  is a constant known as decay constant or disintegration constant. It is characteristic of each radionuclide. Decay constant refers the fraction of remaining atoms that decays per unit time and its unit is  $s^{-1}$ . The minus sign indicates that N is decreasing with time. After integration, the equation can be represented by a mathematical relation

$$N = N_0 e^{-\lambda t}$$

where,  $N_0$  is the initial number of atoms, e is the base of natural logarithm ( $e = 2.719$ ). The equation shows that the number of atoms of a given radioactive element decreases exponentially with time (Fig.12.1). From Figure 12.1, it is found that the disintegration takes place at a very rapid rate initially, with gradual increase of time it

decreases and reaches zero. Theoretically, an infinite time is required to disintegrate all the atoms.



**FIG.12.1:** Decay curve of a radioisotope in a linear graph

## HALF-LIFE PERIOD

The half-life of a radioactive element is defined as the time taken for half the number of atoms to disintegrate. From the law of radioactive disintegration,

$$N = N_0 e^{-\lambda t}$$

If T is the half-life, then  $t = T$ ,  $N = N_0/2$ , substituting we get

$$N_0/2 = N_0 e^{-\lambda T}$$

$$1/2 = e^{-\lambda T}$$

$$2 = e^{\lambda T}$$

$$\log_e 2 = \lambda T, \text{ or } T = 0.6931/\lambda$$

The half-life of a radioactive element is inversely proportional to the decay constant of that element. The following are the half-life value for some important radioisotopes used in medicine.

Radium-226	:	1622 years
Cobalt-60	:	5.26 years
Cesium-137	:	30 years
Iridium-192	:	73.8 days
Iodine-131	:	8.04 days
Technetium-99m	:	6 hours

The mean life or average life ( $T_a$ ) is the average lifetime for the decay of radioactive atoms. It is defined as the lifetime of an imaginary source which decays at a constant rate equal to the initial activity. The mean life is related to the halflife as,  $T_a = 1.44T$ , so the mean life is directly proportional to the halflife.

## SPECIFIC ACTIVITY

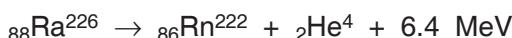
The specific activity is defined as the activity per unit mass and its unit is Bq/kg. Small miniature size sources can be made if the specific activity is higher. If the radioactive substance is liquid, then it is expressed in activity per unit volume (MBq/mL). This is very useful to calculate the activity to be injected to the patient for nuclear imaging study. It also gives an information about the presence of carrier (non-radioactive) material in the sample.

## NUCLEAR TRANSFORMATION

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### ALPHA DECAY

Radioactive nuclides with very high atomic numbers (> 150) decay mostly with the emission of alpha ( $\alpha$ ) particle, followed by gamma rays and characteristic X-rays. This is often followed by internal conversion and emission of Auger electron. As a result, the atomic number is reduced by 2 and the mass number is reduced by 4. A typical example of decay is the transformation of radium (Ra) to radon (Rn).



where, 6.4 MeV is the energy released in the process, called the transition energy. The  $\alpha$  particle is a doubly ionized helium atom with positive charge and about four times heavier than protons. The daughter radon is also radioactive and performs alpha emission. After nine transformations, it becomes a stable nucleus called lead.

### Properties

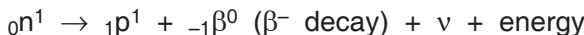
Alpha particles are positively charged and cause intense ionization in matter. The electric and magnetic field influence alpha particles. They have very short range and get attenuated within few  $\mu\text{m}$  in tissue and its range is 1 cm/MeV in air. A sheet of paper cutoff the entire alpha particles. Alpha is a high LET radiation and gives localized intense radiation in living organisms.

### BETA DECAY

The process of radioactive decay, in which an electron or positron is ejected, is called the beta ( $\beta$ ) decay. The electron emission is denoted by  $\beta^-$  and the positron emission is denoted by  $\beta^+$ . Neither the electron nor the positron exists inside the nucleus, but is created at the instant of radioactive decay.

### Beta Minus Decay

A nuclei having excess neutron performs beta minus decay to attain nuclear stability. In this transformation, one neutron is converted into proton, beta and antineutrino ( $\nu$ ). Neutrino is a subatomic particle, having neutral charge and mass lesser than electron. The antiparticle of neutrino is called antineutrino. Hence, the number of neutrons is reduced by 1 and the number of protons is increased by 1.



Beta particle is identical to electron having same mass and charge. When a  $\beta^-$  particle is emitted, the resulting new atom has the same mass number but the atomic number is increased by 1. The beta particle may carry the entire energy or part of the released energy. A typical example for  $\beta^-$  decay is the transformation of cobalt (Co) to nickel (Ni) as follows;

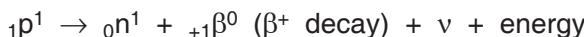


### Properties

Beta particles are negatively charged, and cause ionization in medium. They are deflected by electric and magnetic fields. The range of beta particle is higher than alpha and get absorbed by few mm thickness of aluminium. It causes localized biological effect on living tissues.

### Beta Plus Decay

A nuclei having lesser neutron performs beta plus decay to attain nuclear stability. In this transformation, one proton is converted into neutron, beta plus and neutrino ( $\nu$ ). Hence, the number of neutrons is increased by 1 and the number of protons is decreased by 1.



Beta plus particle is positively charged and is named as positron. It is an antiparticle of electron having same mass, but opposite charge. Positrons are polyenergetic and the mean energy is equal to 1/3 of  $E_{max}$ . The radioisotopes that are produced by accelerator are neutron deficient and decay with positron by increasing the N/Z ratio. For example, F-18 transforms to O-18 with emission of positron.



The positron does ionization and excitation in their path and comes to rest. They react violently with electron and their entire

mass (1.02 MeV) is converted into two gamma photons of energy 0.511 MeV. This process is called *annihilation* in which the photons are emitted at 180 degrees apart, to conserve momentum. This is typical example of Einstein's mass energy equivalence,  $E=mc^2$ . This is the principle used in positron emission tomography. Positron decay is likely in lighter proton richer nucleus, in which the parent-daughter transition energy is equal or greater than 1.02 MeV.

### **ELECTRON CAPTURE**

Nuclei deficient of neutrons may undergo decay by absorbing its own K or L shell electron. When an electron is absorbed, it combines with a proton and converted into a neutron with emission of neutrino. The electron capture is similar to that of positron decay. The atomic number decreases by 1 and mass number remains constant, resulting in increase of N/Z ratio.



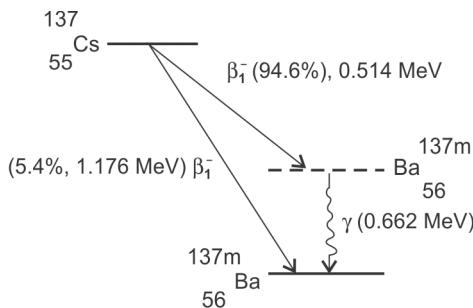
Electron capture creates a vacancy in the shell, which is filled by another electron from high energy shells, followed by characteristic X-rays. If the nucleus is left in the excited state, it will emit gamma rays. These characteristic X-rays and gamma rays are used in medical applications. Electron capture is common in nuclides with parent-daughter transition energy < 1.02 MeV. If energy is greater than 1.02 MeV, both electron capture and positron may take place. Heavier proton richer nuclides likely to undergo electron capture decay process.

### **ISOMERIC TRANSITION**

In most of the radioactive process (either alpha or beta), the daughter nucleus exists in the excited state for an appreciable time. This exited state is known as meta stable state or isomeric state and is denoted by 'm'. The nucleus in the meta stable state is called an isomer. The nucleus goes to the ground state, by emission of gamma photons and attain stability. This process is called isomeric transition. Thus, gamma ( $\gamma$ ) rays are emitted from an exited nucleus and have discrete energies. The isomeric transition of Cs-137 is shown in Figure 12.2.

### **GAMMA RAYS**

Gamma rays are produced, due to isomeric transition of radionuclei from excited state to ground state. In this transition, there is no change in the mass number, atomic number and neutron number.



**FIG. 12.2:** Decay scheme of Cs-137

There is only change in the energy state and no emission and capture of particles. For example,  $^{55}\text{Cs}^{137}$  nucleus decays to the metastable state of  $^{56}\text{Cs}^{137}$  nucleus with the emission of  $\beta$ -particle. Then, it reaches the stable ground state by emitting photon of energy 0.662 MeV by isomeric transition. Gamma rays have the following properties:

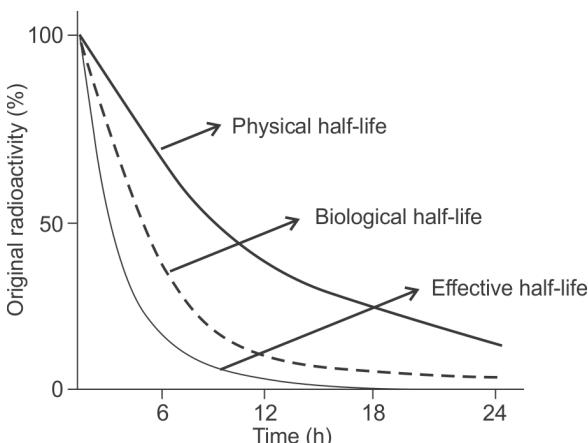
1. Gamma rays are electromagnetic radiations of shorter wavelength (nm).
2. They have no mass and charge.
3. They are not deflected by electric or magnetic fields.
4. They are highly penetrating, their penetrating power is about 100 times greater than that of beta particle. They can pass through many cm length of lead.
5. They affect photographic plate/film and cause fluorescence in materials.
6. They cause ionization in matter.
7. They travel with velocity of light.
8. They damage biological cells and molecules.
9. Gamma rays produce photoelectric effect when they incident on solid surfaces.

## BIOLOGICAL HALF-LIFE

The physical half-life is the time required for a nuclide to decay to half of its original activity. It is expressed by the relation  $T_{1/2} = 0.693 / \lambda$ , where  $\lambda$  is the decay constant. The biological half-life ( $T_b$ ) is determined by the clearance of the radionuclides from the organ, tissue or body. The effective half-life ( $T_e$ ) of a radionuclide in any organ consists of both radioactive decay and biological clearance. The relation between the effective, biological and physical half-life is given by

$$1/T_e = 1/T_b + 1/T_{1/2}$$

For example, if a radionuclide has a physical half-life of 6 hours and a biological half-life of 3 hours, then  $1/T_e = 1/6 + 1/3$ , and  $T_e = 2$  hours. The effective half-life is always less than either the physical or biological half-life (Fig.12.3).

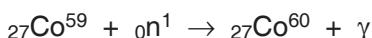


**FIG. 12.3:** Physical, biological and effective decay of a radioisotope

## PRODUCTION OF RADIOISOTOPES

The majority of radioactive isotopes used in medicine are produced artificially by bombarding a stable target nucleus with suitable high energy particles. Alpha, proton and neutron are used as particles for bombarding the nucleus. Neutrons are found to be more effective in producing radioisotopes, using nuclear reactors.

Radioisotope of an element is prepared by placing small quantities of the pure element, in small containers made of aluminum in an atomic reactor for a period of several weeks. The element is converted into radioactive isotope due to continuous bombardment by neutrons inside the reactor. For example, cobalt-60 is produced with  $(n,\gamma)$  reaction.

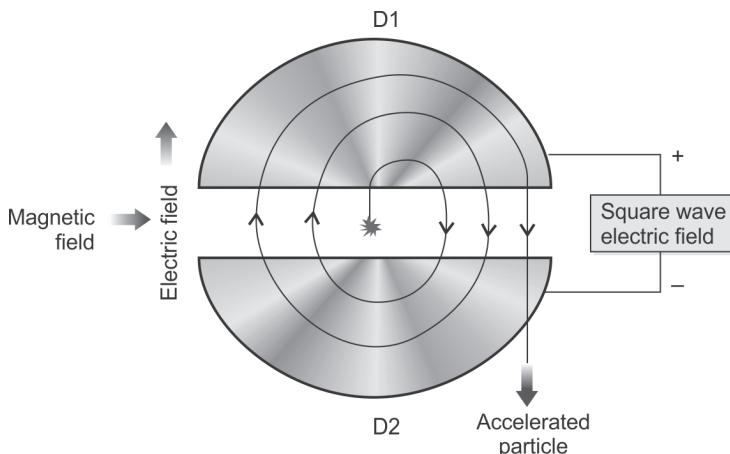


In another method, radioisotopes are produced by bombarding the element with accelerated particle from a cyclotron.

## CYCLOTRON

The spiral type of accelerator called cyclotron was first developed by Lawrence in 1930. It consists of a hollow cylinder divided into two

sections D<sub>1</sub> and D<sub>2</sub>. Each section is known as Dee because it resembles the letter D (Fig.12.4). They are kept separated and placed inside a vacuum chamber. The Dees are connected to a high frequency oscillator. The whole apparatus is placed between the pole pieces of a strong electromagnet. The magnetic field (B) is perpendicular to the plane of the Dee.



**FIG. 12.4:** Cyclotron principle

When a positive ion with charge  $q$  and mass  $m$  is emitted from the source, it is accelerated towards the Dee having the negative potential at that instant. Due to the magnetic field, the positive ion moves along a semicircular path. By the time the particle arrives at the gap, the polarity of the Dee gets reversed and the particle is once again accelerated and enters the other Dee with a greater velocity describing a semicircle of greater radius. The charged particle of mass  $m$  describes a circular path of radius  $r$  when its velocity is  $v$ , then the centripetal force and centrifugal forces are equal;

$$Bqv = mv^2/r \text{ or } \frac{v}{r} = \frac{Bq}{m} = \text{constant}$$

where,  $Bq$  is the magnetic force and  $mv^2/r$  is the centripetal force for circular motion. The time taken ( $t$ ) to describe a semi circle is given by

$$t = \frac{\pi r}{v} = \frac{\pi rm}{Bqr} = \frac{\pi m}{Bq}, \text{ by substituting } v$$

It is seen that 't' is independent of the radius (r) and velocity (v) of the particle. The period of motion of a charged particle is independent of velocity under a uniform magnetic field. This is the basic principle of cyclotron. If v is the frequency of rotation, then

$$2\pi v = \omega = v/r = Bq/m$$

$$v = Bq/2\pi m$$

Since  $Bq/2\pi m$  is a constant, then the frequency of rotation is a constant. The RF oscillator is adjusted to satisfy this condition in a given magnetic field B for the charge q. After spiralling several times within the Dees and acquiring large velocity (kinetic energy), the particle is finally extracted out through a window by means of a deflector plate.

The cyclotron can accelerate protons, deuterons, and alpha particles. At high velocities,  $Bq/2\pi m$  is not a constant due to relativistic mass variation of the particle. As the velocity increases, the mass also increases. That means the frequency of rotation of the particle decreases, and the particle takes longer time to complete its semicircular path. It results in phase instability and the particle will not arrive at the gap just when the polarity reverses. However, this effect can be overcome by decreasing the frequency of the alternating voltage over short intervals to keep in step with the accelerated particles. This is the principle of synchrocyclotron.

Alternatively, the magnetic field may be increased in increasing radius by having special design of magnet called hill and valley design. This will provide azimuthally varying magnetic field, so that the relativistic mass variation is taken care of. Of course, the increasing magnetic field may cause the particle to deviate the median plane in which it is revolving. Hence, optimal varying magnetic field is applied to keep the particle in the gap as well as in the median plane.

### **Medical Cyclotron Facility**

The cyclotron is employed in medicine for particle acceleration, which is used for (i) radiotherapy and (ii) medical imaging. The cyclotrons used in medical imaging are negative ion accelerators, which accelerates both proton and deuteron for the production of positron radionuclides. The advantage of negative ion accelerators are (i) extraction with thin foil ( $5 \mu m$ ) is possible, (ii) extraction efficiency is better, (iii) size is compact and simple, (iv) electron stripping does not induce radioactivity, etc. However, its vacuum requirement is stringent.

They are available in the form of self shielded design, which can reduce the radiation level up to 1–5  $\mu\text{Sv/h}$ . For self shielding, high density polystyrene (HDP), lead and boronated water are used in 8 tanks as shield. This will attenuate both neutrons and gamma rays. A typical medical cyclotron consists of ion source system, RF system, vacuum system, magnetic system, extraction system and target (Fig.12.5.). In addition, power supply with UPS, cooling system, gas distribution, radiosynthesis hot laboratory and quality control laboratory are required.



**FIG. 12.5:** Medical cyclotron facility, GE  
(Courtesy: Dr Kamakshi Memorial Hospital, Chennai)

The ion source system allows hydrogen gas to flow across tantalum cathodes, where hydrogen gets ionized and accepts one electron and become negative ion. The ion source is mounted at the center of the cyclotron between the Dees. Two resonators with RF power system accelerate the particle under the influence of the azimuthally varying magnetic field. The Dees may be 2 or 4 and in the latter case, the particle is accelerated 8 times (one push and pull, for each gap). Vacuum ( $1.2 \times 10^{-5}$  mbar) is required to avoid collision of accelerated particles and gas molecules as well as to insulate the Dees.

The extraction carousels use 6 carbon foils, which remove 2 electrons and the accelerated particle becomes positive ion. The target body is made up of silver and has provision for liquid/gaseous target under helium gas cooling. The target material of water enriched with  $^{18}\text{O}$  is used for  $^{18}\text{F}$  [Fluorodeoxyglucose (FDG), half-life 110 min] production through (p,n) reaction. The other positron emitters that can be produced in the medical cyclotron are  $^{15}\text{O}$ ,  $^{13}\text{N}$ ,  $^{11}\text{C}$ . The cyclotron may produce 10 Ci activity in 2 hours under a beam current of 100  $\mu\text{A}$  with particle energy of 16.5 MeV.

### Cyclotron Produced Radionuclides

Cyclotron accelerators produce radionuclides by bombarding stable nuclei with high energy particles. Protons, deuterons and  $\alpha$  particles are commonly used to produce radionuclides. Gallium-67 is an example of a widely used cyclotron-produced radionuclide. The production reaction is written as



where, Zn-68 is the target and a proton (p), accelerated to about 20 MeV, is the bombarding particle. Two neutrons are emitted during this reaction. In some cases, the nuclear reaction produces radionuclides that decays and gives newer radionuclides, e.g. I-125, I-123, and Tl-201. The commonly used radionuclides and their characteristics are listed in Table 12.1.

**TABLE 12.1** Characteristics of radionuclides

Nuclide	Photons (keV)	Production mode	Decay mode	Half-life
$^{67}\text{Ga}$	93,185,296,388	Cyclotron	EC	78 hours
$^{99\text{m}}\text{Tc}$	140	Generator	IT	6 hours
$^{111}\text{In}$	173, 247	Cyclotron	EC	68 hours
$^{123}\text{I}$	159	Cyclotron	EC	13 hours
$^{125}\text{I}$	27, 36	Reactor	EC	60 days
$^{131}\text{I}$	364	Fission product	$\beta$	8 days
$^{133}\text{Xe}$	80	Fission product	$\beta$	5.3 days
$^{201}\text{Tl}$	70,167	Cyclotron	EC	73 hours

EC = electron capture, IT = isomeric transition

### Nuclear Reactor Produced Radionuclides

Nuclear reactors are also used to produce radionuclides. Neutrons, being uncharged, have an advantage of penetrating through the nucleus without being accelerated to high energies. The nuclear reactor uses two methods, namely, (i) nuclear fission and (ii) neutron activation, to produce radionuclides. The radionuclides, obtained from the fission process are molybdenum-99 (Mo-99), iodine-131 (I-131), and xenon-133 (Xe-133). Examples of radionuclides produced by neutron activation are P-32 and Cr-51.

## RADIOPHARMACEUTICALS

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Radiopharmaceuticals should have desirable characteristics for nuclear imaging as given below:

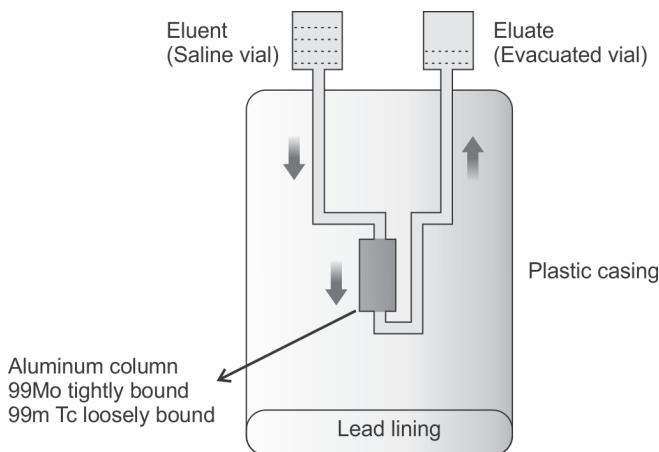
1. The physical half-life should be in few hours, equal to duration of preparation and injection. It should decay to a stable daughter.
2. It should emit gamma rays (50–300 keV), without alpha, beta particles and very low energy photons. The energy is high enough to exit the patient and low enough for collimation.
3. It should have monoenergetic gamma energy, for easy scatter elimination. Decay by isomeric transition and electron capture is preferable.
4. It is easily attached to a pharmaceutical at room temperature, but no effect on its metabolism. It should localize largely and quickly in the target of interest.
5. It should have high specific activity, with low toxicity and is readily available at the hospital site.
6. It is easily eliminated from the body with an effective half-life similar to duration of examination.

## TECHNETIUM GENERATOR

Tc-99m emits gamma energy of 140 keV with a half-life 6 hours and has 90% of clinical use. Its energy is suitable for easy absorption and collimation by a thin crystal with good spatial resolution. Its half-life and pure gamma emission helps to inject large activity to the patient, resulting in reduced noise in the image. It is obtained from Mo-99 on daily basis from the generator, which is a lead shielded container (Fig. 12.6). It contains an exchange column of alumina beads, in which the parent <sup>99</sup>Mo compound is absorbed.

Mo-99m is produced by nuclear fission of U-235, and is in the form of ammonium molybdate ( $\text{NH}_4^+ \text{MoO}_4^-$ ) and has a half-life of 67 h. When it is supplied to the hospitals, the Tc-99m activity has built up to a maximum, equal to the parent (Mo). The daughter and parent decay together with the half-life of the parent, 67 h. Hence, they are said to be in transient equilibrium.

The ammonium molybdate is loaded onto the alumina column (porus) and the Mo-99 decays to Tc-99m. Sterile isotonic saline (0.9%) is passed through the column to remove Tc-99m. This process is called elusion and takes only few minutes. The Mo-99 is not soluble in saline and hence remains in the column. When the saline is passed through the

**FIG.12.6:** Technetium generator

column, the chloride ions easily exchange with the  $\text{TcO}_4^-$  ions, producing sodium pertechnetate  $\text{Na}^+(\text{<sup>99m</sup>TcO}_4^-)$ . This flows under pressure and is collected in sterile rubber capped vial. After the elusion, the  $\text{Tc-99m}$  decays with half-life of 6 h.

$\text{Tc-99m}$  is used in clinical medicine as sodium pertechnetate-  $\text{99mTc}$  which is used for imaging tissues, e.g. thyroid, gastric mucosa and salivary glands, due its similarity to iodine and chloride ions. It is blocked from thyroid by administration of potassium per chlorate, and can be used for cerebral blood flow, and testicular imaging. It is mixed with bran porridge for gastric emptying studies. Technetium is easily labelled to wide variety of compounds as shown in the Table 12.2.

### IODINE-131

Iodine-131 is a reactor produced radionuclide, highly reactive and an excellent label. It is easily trapped and metabolized by the thyroid organ. It is the first radionuclide used for imaging and it is inexpensive, and has long half-life of 8.06 d. It decays by beta emission to stable Xenon-131, average beta energy (90%) is 192 keV, and the dominant photon energy is 364 keV (82% abundance). It gives a whole body dose of 0.5–3.5 rad per mCi and thyroid dose of 100–2000 rad per mCi.

It is clinically administered as iodide, much less satisfactory isotope for imaging to day, because of the high radiation dose to the patient. When iodine is administered as iodide ion, it is readily absorbed from the GI tract and distributed in the extracellular fluid. It is concentrated in the salivary glands, thyroid, and gastric mucosa. It is mainly excreted

**TABLE 12.2** Technetium labeled compounds and their clinical use

Labeled compounds	Clinical use
Methylene diphosphonate (MDP)	Bone imaging
Hexamethyl propylene amine oxime (HMPAO)	Cerebral imaging
Dimercaptosuccine acid (DMSA) and Mecapto-acetyltriglycine (MAG3)	Renal study
Iminodiacetic acid (IDA)	Biliary study
Human serum albumin (HSA) colloidal particles (0.5 µm size)	Liver, spleen and red marrow imaging
HAS macroaggregates (15–100 µm microspheres)	Lung perfusion study
Diethylene triamine pentacetic acid (DTPA) aerosol (5 µm particles)	Lung ventilation study
Analogous red cells	Cardiac function study
Tetrofosmin or sestamibi	Cardiac perfusion study

through urine (35–75% in 24 hours). Iodine is trapped and organified by the normal thyroid and has an effective half-life of 7 days. Nowadays iodine-123 is replacing iodine 131, which is a cyclotron produced and more expensive radionuclide. It can be labeled with hippuran for renal imaging study.

## OTHER RADIONUCLIDES

Apart from technetium and iodine, there are radionuclides that have very useful features and clinical applications. That includes Xenon-133 ( $^{113}\text{Xe}$ ), Krypton-81m ( $^{81\text{m}}\text{Kr}$ ), Gallium-67 ( $^{67}\text{Ga}$ ), Indium-111( $^{111}\text{In}$ ), Indium-111m ( $^{111\text{m}}\text{In}$ ), and Thallium-201 ( $^{201}\text{TI}$ ), etc.

$^{113}\text{Xe}$  is a reactor produced radionuclide, having half-life of 5.2 d, emits beta and low energy gamma rays of 81 keV. It is an inert gas, soluble in blood, used for lung ventilation imaging.  $^{81\text{m}}\text{Kr}$  is a generator produced inert gas, has half-life of 13 s and emits gamma rays of 190 keV. It is inhaled with air for pulmonary ventilation study.  $^{67}\text{Ga}$  is a cyclotron produced radionuclide, with half-life of 78 h, emits gamma rays of energies 93,185 and 300 keV, respectively. It is used as gallium citrate to detect tumors and abscesses.

$^{111}\text{In}$  is a cyclotron produced radionuclide, has a half-life of 67 h, emits gamma rays of 173 and 247 keV energies. It is used for labeling white blood cells and platelets, to detect abscess and

thromboses.  $^{111m}\text{In}$  is a generator produced radionuclide, having a half-life of 100 m with gamma energy of 390 keV. It is often used as replacement to indium-111.  $^{201}\text{TI}$  is a cyclotron produced radionuclide, has a half-life of 73 h and emits X-rays of energy 80 keV. It is used as thallous chloride in myocardial perfusion imaging.

Generally, a radiopharmaceutical is formed by mixing the radionuclide with a compound to be labeled at room temperature. Of course, it may require additional chemicals. It may require sterile work station, shielded syringes and a glove box, in a room filled with sterile air and positive pressure. As a part of quality control, these pharmaceuticals are tested for their radiochemical purity, chemical purity, sterility and pyrogen, before injected to the patient.

## **GAMMA CAMERA**

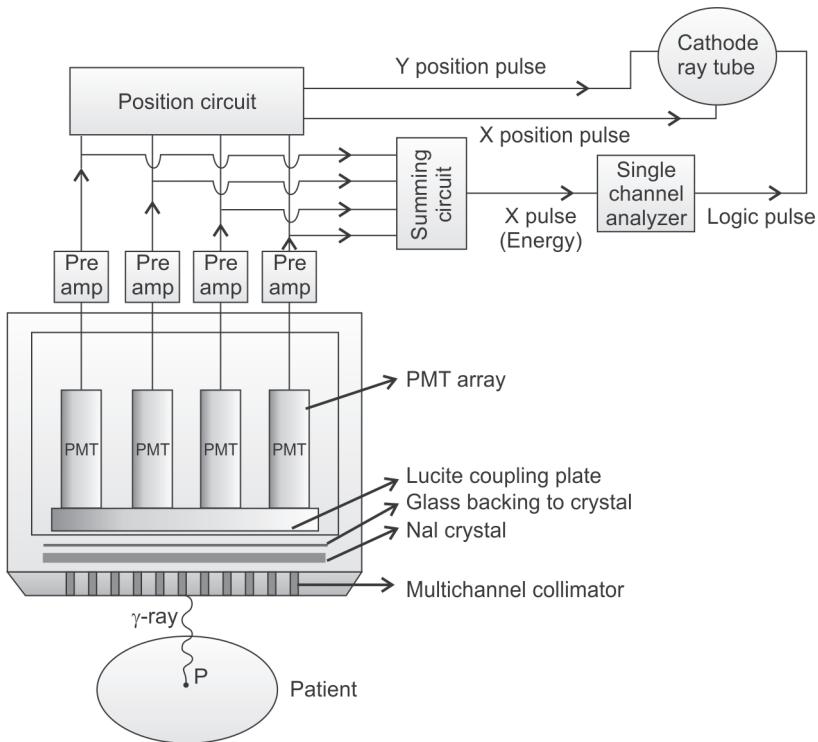
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The gamma camera was designed by Hal O Anger of Donner Laboratory, in Berkely, California in 1950. It uses Tc-99m gamma rays (140 keV) and characteristic X-rays to form images. Tc-99m is injected intravenously, concentrates in the organ of interest, and emits gamma rays, which are detected by the gamma camera. It is a planar imaging, that produces images of the radionuclide distribution in patients.

The equipment consists of (i) multihole collimator, (ii) scintillation crystal, (iii) photomultiplier, (iv) position and summing circuits, (v) pulse height analyzer (vi) computer and display (Fig. 12.7). It is available in the market as single head or double or triple head with different specifications. FOV of 250 mm as mobile camera with good resolution for TI-201, can be used for cardiac study. A FOV of 400 mm, general purpose equipment may suits for Tc-99m. A 500 mm FOV may cover whole patient width, and is suitable for bone and gallium study.

## **MULTIHOLE COLLIMATOR**

The multihole collimator is a lead disk of 25 mm thick and 400 mm diameter. It has 20,000 circular or hexagonal holes of diameter 2.5 mm with septa of 0.3 mm. This is suitable for Tc-99, whose HVL= 0.3 mm. Each hole accepts gamma rays, originating from perpendicular direction to the crystal. The photons originating from nonperpendicular direction is absorbed by the collimator walls (septa). However, scattered rays can pass through the collimator, but have lesser energy, may be rejected by the pulse height analyzer, at the later stage.



**FIG. 12.7:** Gamma camera with scintillation crystal and multihole collimator

## SCINTILLATION CRYSTAL

Scintillation crystal is a single, large circular crystal of 500 mm diameter 9–12 mm thick, high atomic number ( $Z = 53$ ) and high density, e.g. Nal (Tl). It absorbs 90% TC-99m gamma rays by photoelectric process, and 30% for I-131. It is fragile, hygroscopic, and influenced by temperature. Hence, it is encapsulated in an Al cylinder, with one transparent face, for the entry of gamma rays.

Gamma ray entering the crystal, gives flash of light and UV (1:5000). These light photons undergo multiple reflections in  $< \mu\text{s}$  and reaches the photomultiplier tube (PMT) via transparent face. About 4000 light photons may reach the PMT out of 5000. The other face of the crystal is coated with titanium compound, to ensure reflection. Distribution of light leaving the crystal depends on the collimator hole. To avoid loss of light, a flat Lucite light coupling is provided between crystal and the PMT.

## **PHOTOMULTIPLIER TUBE**

The crystal is coupled to a number of PMTs (91), and each PMT is an evacuated glass envelope, having a photocathode (Figs 6.4 and 6.5, chapter 6). It absorbs light and emits photo electrons (1 electron per 5–10 photons). The electrons pass through series of dynodes and produce additional electrons. All the electrons ( $10^6$ ) are accelerated towards the anode, to create a charge pulse.

Each PMT is provided with a preamplifier, by which the signals are further amplified. The amplitude of the pulse produced by each PMT is proportional to the amount of light it receives, following a gamma interaction in the crystal. The pattern of light emission in the crystal forms a 2D projection of 3D activity distribution in the patient. The PMT which is nearer to the photon interaction receive more light and produce larger voltage pulse, than those at far distance. Hence, the relative amplitude of the pulses from the PMTs contains sufficient information to determine the location of each interaction in the plane of the crystal.

## **POSITION AND SUMMING CIRCUITS**

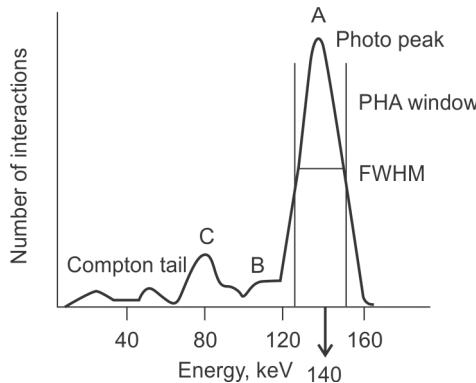
The pulses from the preamplifier are fed to the position and summing circuits. The position circuit receives the pulses from the individual preamplifier, after each photon interaction in the crystal and produces an X position pulse and Y position pulse respectively. The X, Y pulses together specify the position of the interaction in the plane of the crystal.

The summing circuit adds the pulses from the individual preamplifier to produce a Z pulse (energy). The amplitude (height) of Z pulse is proportional to the total energy (keV) deposited in the crystal, and the height is stated in keV. The Z pulse is fed to a pulse height analyzer (PHA), for energy analysis.

## **PULSE HEIGHT SPECTRUM**

In the spectrum, the pulse height is plotted against time. The only pulse that falls inside the PHA window (photo peak) is selected to form image. The spectrum consists of photo peak and tail. The photo peak (A) is formed by the pulses, that is produced by photoelectric absorption in the crystal (Fig. 12.8).

The spread of the spectrum is due to statistical fluctuations of gamma rays and electrons. The tail represents low energy gamma rays suffered



**FIG.12.8:** Pulse height spectrum of NaI (TI) detector with  $^{99m}\text{Tc}$

by Compton interactions in patient or crystal. In the spectrum, only the photo peak is used to locate the position of radioactivity in the patient, and the Compton tail is rejected.

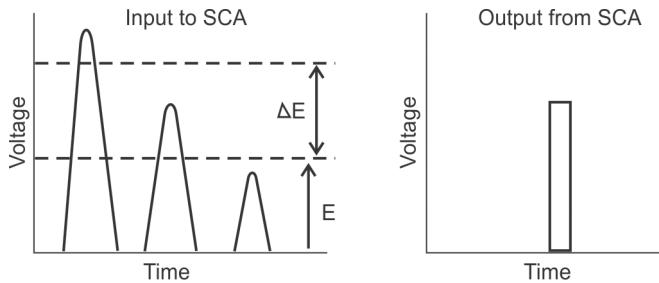
The photo peak A is caused, when incident gamma rays (140 keV) is completely absorbed in the crystal by photoelectric process. The peak width is measured by full width half maximum (FWHM). The FWHM is used to express energy resolution:

$$\text{Energy resolution} = \frac{\text{FWHM}}{\text{Peak energy}} \times 100\%$$

Escape peak (B) is due to Iodine K-shell characteristic X-rays (28 keV), and the measured gamma energy is only 112 (140–28) keV. Lead X-ray peak (C), is due to primary gamma rays that interact with the lead shield or collimator and emits characteristic X-rays (70–90 keV). Tail or Compton tail is due to Compton scatter recoil electron, which is absorbed in the crystal and a peak (0–50 keV) is produced.

## PULSE HEIGHT ANALYZER

A photo peak  $\pm 10\%$  is set as window in the pulse height analyzer (PHA). The window for Tc-99m is 126–154 keV, that is centered on 140 keV gamma energy (Fig. 12.9). PHA gives logic pulse, only if the pulse is within a preset range of energies, and the selected pulses are referred as counts. The pulses that are lower or higher than the preset values are rejected. More number of PHA is required to detect multiple energies. Modern cameras use 2 to 4 PHAs for Ga-67 and In-111, which emits more than one energy.



**FIG. 12.9:** A single channel pulse height analyzer with window width and output pulse

## COMPUTER AND DISPLAY

The pulses are digitized by an analog digital converter (ADC). The X, Y pulses and the logic pulse of Z are then fed to a computer, which records the Z pulse as a count in memory location, corresponds to x,y coordinates. As more pulses arise, the count builds up in each memory location, and stored as digital image in the  $128 \times 128$  pixel matrix. The image is displayed on a monitor screen.

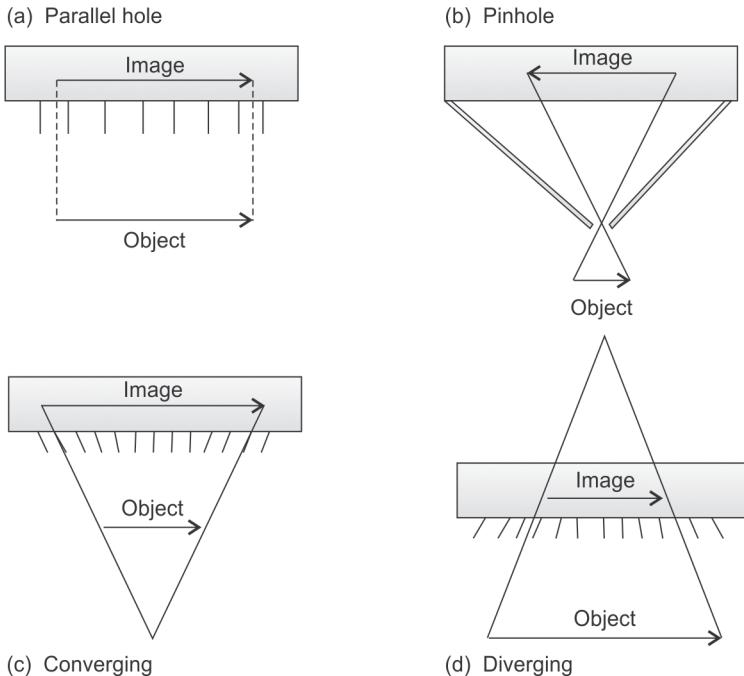
The brightness of each pixel is proportional to number of counts, in turn to gamma rays originated from the patient and the activity at that location. Thus, images are built up by using 50,000 to 1 million counts per image frame. If the counts are too long, or short, the monitor screen would appear as uniformly bright or grainy.

## TYPES OF COLLIMATORS

The collimators direct the photons arising from certain directions to the crystal. Collimators are made up of high atomic number, high density material, such as lead. There are four types of collimators, namely, parallel hole, pinhole, converging, and diverging collimators (Fig. 12.10). Parallel hole, and pinhole collimators are most commonly used in gamma camera.

### Parallel Hole Collimator

Parallel hole collimator has thousands of parallel holes, which may be round, square, triangular or hexagonal. Septa must be sufficiently thick to absorb the photons incident upon them. Hence, thick septa collimators are used with radiopharmaceuticals that emit high energy photons. No collimator will give the expected spatial resolution and efficiency and there is always a compromise between them.



**FIG. 12.10:** Type of collimators used in gamma camera

Variety of parallel hole collimators are available that includes, low energy high sensitivity (LEHS), low energy all purpose (LEAP), low energy high resolution (LEHR), medium energy and high energy. The collimator to object distance (COD) will not affect the image size, but degrades the spatial resolution with increasing COD. The FOV, in air sensitivity are the same at all distances from the collimator face.

#### Divergent and Convergent Collimators

Divergent hole collimator is used with small diameter camera, to get larger FOV, to image large organs, e.g. lung. This type of collimator minimizes the image. Convergent hole collimator, magnifies the image, but reduces the FOV, useful in children, or small organs. Spatial resolution deteriorates at the edges of the collimator and both suffer from geometrical distortion. The FOV and in air sensitivity vary with distance in this type of collimator.

#### Pin Hole Collimator

The pin hole collimator is a cone of lead with single hole of few mm diameter. It gives magnified but inverted image of a superficial organ, e.g. thyroid. Sensitivity is the measure of gamma ray detection and

it is < 1% for this collimator. Wider and shorter the hole, greater the sensitivity, and require only lesser radionuclide with lesser patient dose. This will reduce spatial resolution, which is the limitation of gamma camera.

### **DYNAMIC IMAGING**

Function of a organ can be studied by acquiring series of separate image frames rapidly, e.g. kidney, lung, heart, etc. It can be displayed on screen as cine loop or recorded in a multi format film. There is a cursor to define region of interest (ROI), and the total counts in the ROI is measured and can be displayed over time, e.g. renogram. In multigated cardiac study (MUGA), each image of 49 s duration is acquired at 20–30 points in a cardiac cycle. At each point, several hundred successive images are added, pixel by pixel, to improve statistics, and reduce noise.

### **SINGLE PHOTON EMISSION COMPUTED TOMOGRAPHY**

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In planar imaging, 2D projection of a 3D distribution of a radiopharmaceutical is obtained. It involves superimposition of organs, loss of depth information and reduction of contrast. This is overcome in emission tomography imaging. Gamma camera is used as single photon emission computed tomography (SPECT). The camera with parallel hole collimator rotates around the patient in 6° interval. Each interval it halts (20–30 ms) and acquires patient view, resulting 60 views. Thus, 3 million counts are acquired in 30 minutes.

The sensitivity can be improved by dual headed or triple headed camera. Counts from the center is less than edges, due to attenuation and require attenuation correction algorithm. Filtered back projection or iterative methods are used for image reconstruction, which converts the 60 views into transverse images. Thus, 20–30 transverse parallel sections can be obtained. In addition, sagittal, coronal, and oblique sections can also be obtained. 3D display and image rotation similar to that of multi slice CT is also possible.

#### **SPECT Image Noise**

SPECT images contain noise, since few photons form the image view (20–30 ms). Longer imaging is not possible due to patient motion. Thicker slices reduce noise, but increase partial volume artifacts. Hence, pixel matrix of  $64 \times 64$ , with 60 views are used. Noise reduction is done

by mathematically, but resolution is reduced. High sensitivity collimators reduce spatial resolution (18 mm), which is higher than that of planar gamma camera image and CT image.

Image reconstruction magnifies the effect of noise and field non uniformity. PMT's are affected due to camera rotation, needs shielding and automatic balance. Camera can be put in elliptical orbit around the patient to reduce gap and improve resolution. Alternatively, multiple cameras that are equally spaced will improve resolution and throughput.

### SPECT Application

Thallium study of myocardial infarctions and ischemia, with quantitative cerebral blood flow are the main uses of SPECT. Detection of tumors and bone irregularities are possible with combination of SPECT and CT. Gated acquisition is also possible with SPECT as a planar imaging device, e.g. MUGA study. Cardiac gated myocardial SPECT can be used to obtain quantitative information of myocardial perfusion, myocardium thickness, left ventricular ejection fraction, stroke volume and cardiac output.

## POSITRON EMISSION TOMOGRAPHY

Positron emission tomography (PET) generates transverse images that describe the distribution of positron-emitting nuclides in patients. It is equivalent to a CT scanner in diagnostic radiology, which detects radiation, transmitted from the patient. However, PET scanner gives, cross-sectional view of the activity, in a plane through the origin of interest in the patient.

### PRINCIPLE

Positron-emitting isotope (10 mCi) is injected to the patient intravenously. The radiotracer decays by emission of positron ( $\beta^+$ ), with short half-life, e.g. fluorodeoxyglucose, the FDG. The positron annihilation produces two photons of energy 511 keV each emitted in opposite directions ( $180^\circ$ ) in the body site. These photons are detected by the set of ring detectors surrounded the patient. If two or more detectors detect the photons at the same time, it is called annihilation coincidence detection (ACD).

Scintillation detectors are used to detect the ACD. This implies that the annihilation has occurred close to the line connecting the direction of two photons. If the photon is detected by a single detector ring,

then the coincidence circuitry in the scanner determine the line in space along which the positron annihilated. Thus, coincidence counting establishes the trajectories of photons, in order to locate the annihilation events.

The above information is stored in the computer. The computer then performs the filtered back projection of this data, to create the transverse images. Finally, visual display of the slice of interest is obtained. The ACD method of detection is much powerful and needs no collimator. Hence, the efficiency of PET scanner is about 10–20 times that of a SPECT camera.

### POSITRON EMITTERS

PET scanner makes use of short-lived positron emitters  $^{11}\text{C}$  ( $T_{1/2} = 20$  min),  $^{13}\text{N}$  (10 min),  $^{15}\text{O}$  (2 min),  $^{18}\text{F}$  (110 min), and  $^{68}\text{Ga}$  (68 min). Their physical characteristics are given in Table 12.3.  $^{18}\text{F}$  is labeled as fluorodeoxyglucose, and is used to study glucose metabolism. It is similar to glucose molecule in the human body and undergoes metabolism as glucose-6-phosphate. It gets trapped wherever lesion is present. It travels only 2 mm in the patient, before annihilation and require cyclotron for in house production (Fig. 12.11).

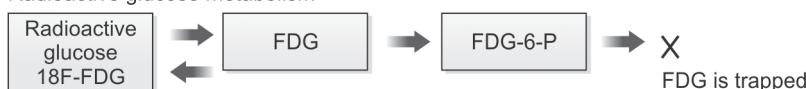
**TABLE 12.3** PET radionuclides and their physical characteristics

Nuclide	Half-life	Tracer	Application
O-15	2 minutes	Water	Cerebral blood flow
C-11	20 minutes	Methionine	Tumor protein synthesis
N-13	10 minutes	Ammonia	Myocardial blood flow
F-18	110 minutes	FDG	Glucose metabolism
Ga-68	68 minutes	DOTANOC	Neuroendocrine imaging
Rb-82	72 seconds	Rb-82	Myocardial perfusion

Glucose metabolism



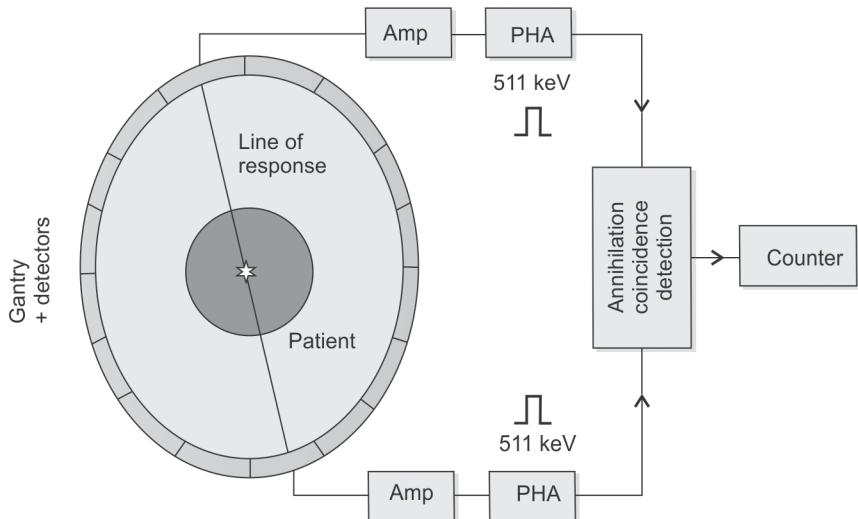
Radioactive glucose metabolism



**FIG. 12.11:** Mechanism of FDG in human body

## PET EQUIPMENT

PET equipment consists of a ring of detectors surrounding the patient (Fig. 12.12). Scintillation crystals (10,000–20,000) are used as detectors. The various scintillation detectors are: Bismuth germanate (BGO), lutetium oxyorthosilicate (LSO), and gadolinium oxyorthosilicate (GSO). Generally, detectors must have high detection efficiency, less after glow, and good energy resolution.



**FIG. 12.12:** Typical PET scanner geometry

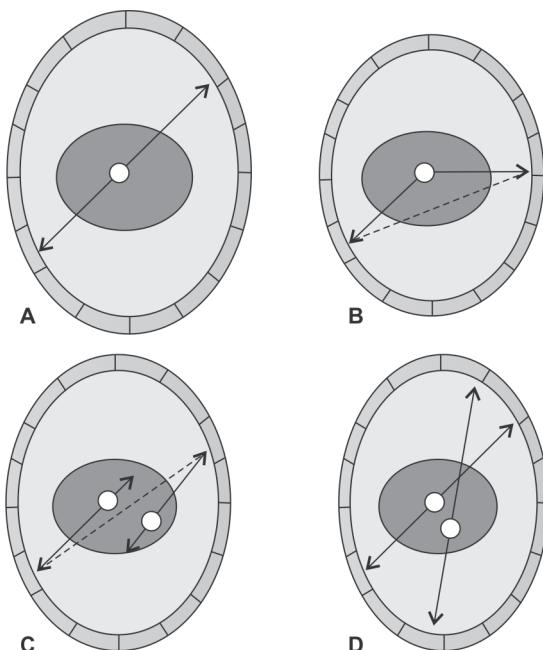
### Detectors

Detectors are made in block format coupled with photomultiplier tubes (PMT). A single larger crystal, is coupled to more than one PMT. Detectors are used in pulsed mode to permit pulse height discrimination and ACD. The signals from the PMT are sent to the pulse height analyzer (PHA) circuits, which reject the pulses other than 511 keV energy. PET scanners detect the emission photons and convert them to light signals. This scintillation event is converted into an electric signal by PMT, that can be displayed on a monitor.

### Coincidence Detection

Annihilation coincidence detection circuit, detects coincidence interaction within each pair of detectors and obtain line of response (LOR). The effective attenuation suffered by the two photons is determined. The time of flight (TOF) of the photons are also detected.

The coincidence may be true coincidence, random coincidence, scatter coincidence and multiple coincidence (Fig. 12.13). A true coincidence arises from a single event, in the same plane. The scattered coincidence is due to single event, but in different plane, due to scatter. The random coincidence is due to two independent events, not in the same plane. Multiple coincidences is the detection of two independent events that are detected at the same time.



**FIG. 12.13:** (A) True coincidence, (B) Scatter coincidence, (C) Random coincidence, and (D) Multiple coincidence

### Data Acquisition

The computer performs a filtered back projection of the data, to produce transverse images of the radionuclide distribution in the patient. Modern scanners are capable of performing simultaneous acquisition of  $> 45$  slices over 16 cm of axial distance.

It starts with a sonogram, which is a sum of the counts of each LOR that is plotted, angle vs distance from isocenter. Sinogram contains number of blurred sine waves with different amplitude and phases. The septa is used to avoid cross plane coincidence. For a given angle, horizontal rows of the sinogram can be converted into a slice. Thus, each transverse slice has its own sinogram. These sinograms are computer analyzed to give series of 2D projection views.

2D acquisition uses septa and sinogram for data acquisition, which is employed in old generation scanners. It accepts data only from parallel 2D planes. The count rate is reduced and involves random and scatter coincidences resulting loss of axial sensitivity. Hence, 3D acquisition is used in modern scanners, which has 6 times greater sensitivity. It avoids septa, and accepts data from all the coincidence planes (Micelogram). It has improved count rate and sensitivity and correction for random and scatter coincidence, to improve image quality. 3D reconstruction algorithm is also used.

### Data Correction

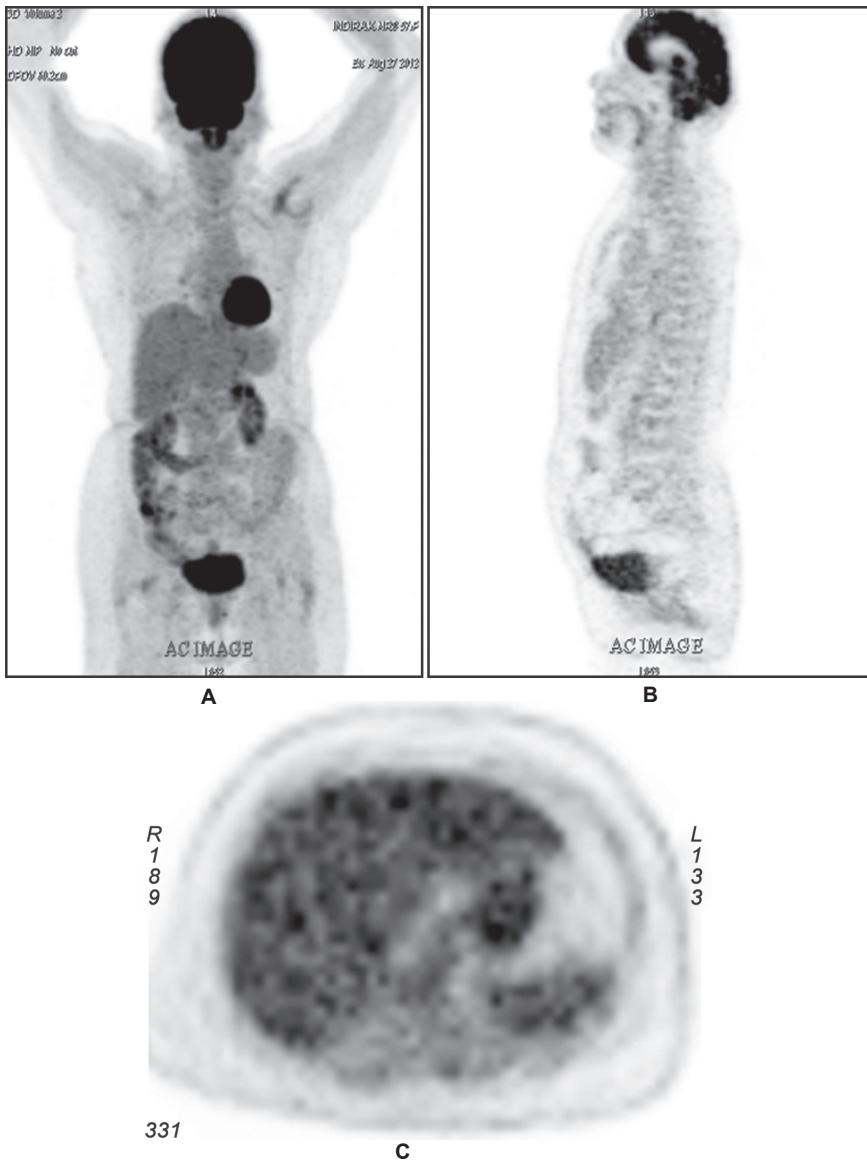
Data correction is required for tissue attenuation, which depends on patient dimension and tissue structure. This is done with long lived isotope (Ge-68, or Cs-137), located in the gantry of the PET. The attenuation correction is carried out by the CT or MRI data in PET-CT and PET-MRI, respectively. Since the detector is a circle, the LORs are bunched together at the gantry side, whereas it is uniformly spaced at the centre. This requires correction and is called arc correction. This is suitable for large organs, in which the LOR may be away from the centre. In addition, large number of corrections are required as given below:

1. Normalization for nonuniformities of detectors.
2. Scatter correction based on energy discrimination.
3. Random correction for events without spatial information.
4. Dead time correction by using paralyzable and nonparalyzable models.
5. Reconstruction and display.

The objective is to provide quantitative cross-sectional images of radiopharmaceutical distribution. The various methods of reconstructions are:

1. 2D Analytic reconstruction using Filtered Back Projection.
2. 3D Analytic reconstruction using Single Slice Rebinning, FORE.
3. 3DRP (3 Dimensional re-projection algorithm).
4. Iterative method.

The end-result of PET acquisition and reconstruction is a 3D image. Each voxel represents regional tissue, and radioactivity concentration. Images are displayed in an axial, coronal and sagittal sections (Figs 12.14A to C). One can visualize the images in gray scale, and each level represents particular activity concentration.



**FIGS 12.14A to C:** PET image display in (A) coronal, (B) sagittal, and (C) axial plane (Courtesy: Dr. Kamakshi Memorial Hospital, Chennai)

The features of PET scanner are: (i) PET images gives functional and physiological information, (ii) it is a noninvasive, and investigative procedure, and (iii) it provides absolute quantitative information on perfusion and metabolism with low noise. Usually, images are fused with CT (PET-CT) and MRI (PET-MRI). There are dedicated PET-CT

and PET-MRI equipments available with PET and CT or MRI in the same gantry.

## PET-CT SCANNER

PET-CT is a medical imaging device which combines in a single gantry system, both Positron emission tomography (PET) and X-ray computed tomography (CT). PET-CT is a hybrid imaging device which has a CT scanner for producing high quality images of anatomy and PET scanner to produce high quality images of function (Fig. 12.15). Images acquired from both devices can be taken sequentially, in the same session from the patient and combined into a single superposed (Co-registered) image.

In addition, CT provides attenuation correction that is required for PET images. There is a computer and software to fuse and display images. Functional imaging obtained by PET, which depicts the spatial distribution of metabolic or biochemical activity in the body can be more precisely aligned or correlated with anatomic imaging obtained by CT scanning. Two-and three-dimensional image reconstruction may be rendered as a function of a common software and control system. No patient motion between the imaging studies. It displays anatomy, function and the parameters fused in a series of images.

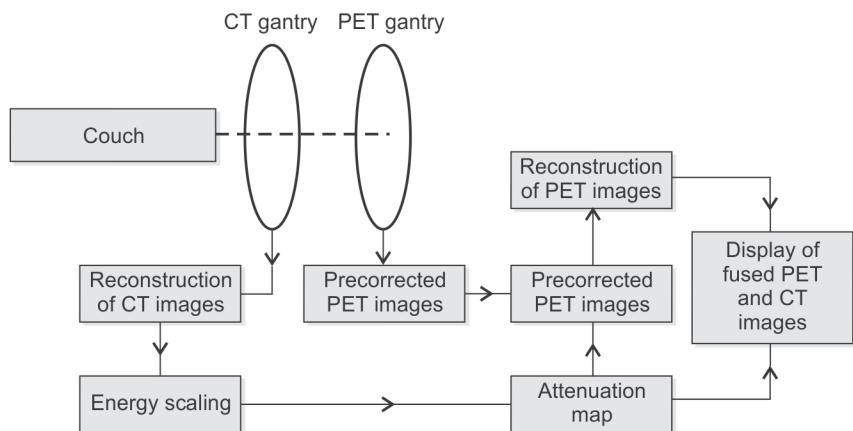


FIG.12.15: PET-CT functional diagram

## PATIENT PREPARATION

Patient should have 6–12 hours overnight fasting. Twenty four hours prior to study, they can have low calorie meal and avoid exercise.

Blood sugar cutoff level should be about 150–200 mg/ml. During post injection, they should relax and do not talk and wait for about 45–60 min. They should empty the bladder before the study begins. During the study, they should lie down and breathe regularly.

## DATA ACQUISITION

The data acquisition starts with a CT topogram, spiral CT scan followed by PET scanning. CT acquisition parameters includes (i) rotation speed (0.33–1 sec), which affects temporal resolution, scan time and dose, (ii) tube current (20–400 mA), which controls the number of X-ray photons and dose, (iii) tube voltage (80–140 kVp), which determines X-ray energy, (iv) table motion (10–100 mm/rotation), which is proportional to scan time, (v) collimation, which affects the axial resolution and scan time and (vi) scan length.

## ATTENUATION CORRECTION

There is a differential attenuation of photon beam from deeper and superficial structures. As a result, image will show high activity near the surface and low activity near the center. CT transmission scan can be used to generate an attenuation map, to correct this attenuation effect. Scaling of attenuation coefficients is a must, since the CT X-ray photon energies (100–140 kVp) are lower than 511 keV emission photons.

## PET-CT IMAGE DISPLAY AND ARTIFACTS

The fused images are shown in Figures 12.16A to C. The images are analyzed in terms of standardized uptake values (SUV) as follows: (Strauss and Conti. J Nucl Med 1991):

$$\text{SUV} = \frac{\text{Tissue FDG uptake}}{\text{Injected dose/patient Weight}}$$

PET-CT images are prone to several artifacts, similar to other imaging, however, the important artifacts are (i) motion artifact, (ii) truncation artifacts, (iii) metallic artifact, e.g. pacemaker and (iv) oral contrast artifact.

Motion artifact arises from change of patient position that leads to misregistration on the fusion images. It commonly occurs in the head and limbs, where movement is likely to occur. This may lead to false interpretation of uptake in normal structures. Similarly, misalignment of PET and CT images can occur at the diaphragm due to respiratory



**FIGS 12.16A to C:** (A) PET image, (B) PET-CT fused image, uncorrected for attenuation, (C) PET-CT fused image attenuation corrected (*Courtesy: Dr Kamakshi Memorial Hospital, Chennai*)

motion and movement of internal organs above and below the diaphragm. This misalignment can cause attenuation correction artifacts. The liver dome and spleen may be seen ‘floating’ above the diaphragm.

Truncation artifacts appear as dark lines extending craniocaudally along the patient. Because PET emission images are acquired over many minutes (25–40 minutes), patients are often unable to keep their arms above their head for the duration of the scan. The CT scanner has a relatively narrow axial field of view of only 50 cm and often the arms, shoulders, and hips lie outside the visualized area. Artifacts are produced at these sites because the CT attenuation correction reconstruction algorithm does not account for any attenuation of the CT X-rays by the tissues outside the field of view.

Pacemakers are one of the most common attenuation artifacts produced by metallic implants in the body. A focus of increased activity can be seen at the site of the pacemaker on attenuation corrected PET images. In this case, a focus can be seen along the metallic

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pacer lead wires. This occurs because of the significantly higher radio-density of the implant on the CT compared to normal body tissues and subsequent over correction of the PET images at this site. The non-attenuation image, however, does not show any 'hot-spots' at that site and confirms it to be an artifact.

Normally, a lower-density barium oral contrast is used in PET-CT. At normal concentrations, the diluted oral contrast is not radio-dense enough to produce attenuation over-correction artifacts. However, with time, the contrast can become compacted in the bowel with a significant increase in the radio-density and foci of increased uptake can be seen. A radiodensity less than 400–500 Hounsfield units should not produce any artifact. Inspection of the non-attenuation corrected images would show normal uptake at the region.

### **RADIATION DOSE**

PET-CT imaging is different from standard nuclear medicine imaging. In a gamma camera imaging,  $^{99m}\text{Tc}$  140 keV photons are used, for which the HVL and TVL in lead are about 0.3 mm and 0.99 mm, respectively. In the case of PET radionuclides, 511 keV photons are involved and their corresponding HVL and TVL values are 4 mm and 13.2 mm respectively for a narrow beam geometry.

Table 12.4 compares the radiation levels at 0.1 m and 1 m from the patient for TC-99m and F-18 FDG. The radiation dose rate is 5–10 times higher in PET-CT scanner compared to gamma camera, which is a matter of concern. The staff radiation dose is about 3–14 mSv per study, and 50% dose is due to dose administration. The patient radiation dose is 2–5 times higher in PET/CT. The PET contributes about 5–11 mSv, whereas CT contribution is about 1–3 mSv for head, 5–20 mSv for abdomen respectively. The radiation dose at 1 m from the patient is 3 mSv per hour per 37 MBq. The patient dose is higher because of multi slice CT, which scans larger volume and use 80–140 kVp and 100–380 mA settings under sub-second rotation time. However, this can be compromised, since the benefit outweighs the risk.

### **ADVANTAGES OF PET-CT**

PET-CT provides physicians with superior information for determining tissue characterizations and classifications, staging of cancers, restaging

**TABLE 12.4** Comparison of radiation level from the patient

Radiopharmaceutical	Dose rate at 0.1 m, $\mu\text{Sv}/\text{h}$	Dose rate at 1 m, $\mu\text{Sv}/\text{h}$
Tc-99m MDP (600 MBq)	114	5
F-18 FDG (350 MBq)	550	70

of cancers, patient prognosis and monitoring the effectiveness of cancer therapies. Advantages include:

1. Superior lesion localization from near-perfect anatomical/functional registration with fewer motion artifacts.
2. Better distinction between physiological uptake and pathological uptake.
3. Consolidation of patient's imaging studies.
4. Shorter scan time (average of 30 minutes, vs 60 minutes with standard PET) by using CT for attenuation correction. This aids in patient comfort and minimizes claustrophobia problems.

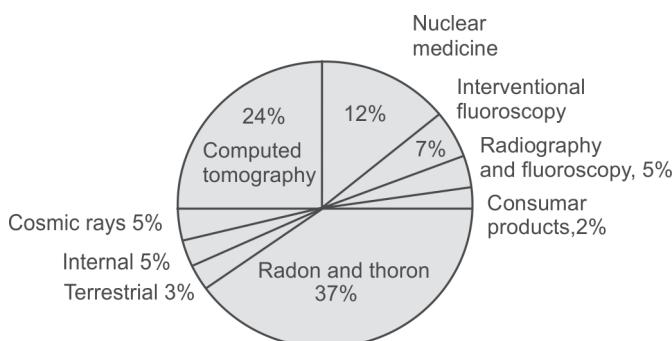
# 13

# Radiological Health and Safety

The hazards of radiation were realized in the beginning of this century, soon after the discovery of X-rays. Radiation exposure limits were introduced by the International Commission on Radiological Protection (ICRP) in 1928. The ICRP reports form the basis for many national and international radiation protection programs. In India, the Atomic Energy Regulatory Board (AERB), established in 1983 is the competent authority which control over the use of radiations in medicine and industry.

## SOURCES OF RADIATION

The sources of radiation are classified into (i) natural background exposure, (ii) medical exposures, (iii) consumer products, and (iv) occupational exposures (Fig. 13.1). The annual average per capita total effective dose is 6.2 mSv (NCRP-160). About 50% of the above exposure (3.1 mSv) arise from natural background exposure, 48% (3.0 mSv) arise from medical exposures, in which computed tomography is the major contributor (24%). Background radiation levels vary with region, Kerala has high background levels of radiation, especially due to monazite in sand.



**FIG.13.1:** Different sources of radiation and their contribution in percentage

## NATURAL BACKGROUND EXPOSURE

The natural background radiation exposure includes (i) radon and thoron, (ii) cosmic rays, (iii) terrestrial, and (iv) internal sources.

### **Radon and Thoron**

Naturally occurring radionuclide emits radon-222, a noble gas, which is produced by U-238. It is responsible for exposure due to inhalation. It decays to polonium with alpha particle and finally become stable lead. Radon inhalation deposits dose in lung and accounts an effective dose of 2.1 mSv per year, which is 68% of natural background. Radon concentration vary widely, it emanates from soil and is restricted by structures. It is also dissolved in water supply. Weather proofing, energy conservation techniques reduce ventilation, resulting in higher indoor radon concentration.

### **Cosmic Rays**

Cosmic rays are outer world radiations comprising both primary and secondary rays. The primary cosmic rays are 80% protons, that collide with atmosphere, producing showers of secondary particles (electrons, muons) and electromagnetic radiations. The average per capita effective dose is 0.33 mSv per year, which makes 11% of the natural background.

Cosmic exposures increase with altitudes and it is doubling in every 1500 m and hence air travel increases individual's cosmic ray exposures. It is greater at earth poles than equator, due to earth's magnetic field. Structures provide some protection against cosmic rays, and hence the indoor effective dose is 20% lesser than outdoor. Secondary cosmic rays mainly consists of carbon-14.

### **Terrestrial**

Terrestrial radionuclides have been present since the formation of earth. They mainly contribute radiation in the form of external exposure, inhalation, and ingestion. U-238, Th-232 and their decay products and K-40 are mainly responsible for external exposure and they account an effective dose of 0.21 mSv per year. It accounts to 7% of the natural background exposure.

### **Internal Exposure**

Ingestion of food and water containing radionuclides such as K-40, Th-232, and U-238 cause internal exposure. The K-40 is the most significant, which accounts an average dose rate of 0.15 mSv per year. Th-232 and U-238 are also found in food and water and contribute about 0.13 mSv per year.

## MEDICAL EXPOSURES

The major contributors of medical exposure are computed tomography (24%) and nuclear medicine (12%). They contribute an effective dose of 1.5 mSv and 0.75 mSv per year, respectively. Though their use is only 22% in medical imaging, but they provide 75% of radiation dose. Radiography and fluoroscopy find 74% application, resulting in only 11% effective dose at the rate of 0.33 mSv per year. Increased use of computed radiography has increased the dose compared to screen-film systems. Interventional radiology accounts an effective dose of 0.44 mSv per year.

## CONSUMER PRODUCTS

Consumer products include tobacco products, building materials and air travel. Tobacco consists of pb-210 (lead) and Po-210 (polonium) and accounts an effective dose of 0.36 mSv per year to a smoker, who consume one pocket per day. Building materials like brick, concrete and granite contain uranium, thorium, radium and potassium and accounts 0.035 mSv per year.

Enhanced natural sources like mining and agricultural activity, combustible fuels including coal, and natural gas and consumer products like smoke alarms (americium-241), gas lantern mantles (thorium), dental prostheses, certain ceramics, and optical lenses (uranium) also contribute annual effective dose to a lesser extent.

## OCCUPATIONAL EXPOSURE

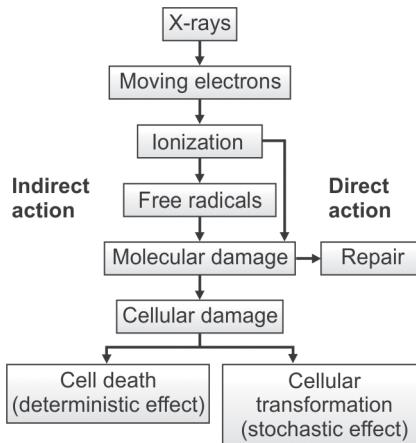
Occupational exposures arise from medical, aviation, industry, including nuclear and research activities. In the case of medical occupation, the mean exposure is 1.1 mSv per year. The interventional radiologists may account for higher radiation exposure. However, all the medical exposures are partial body in nature.

Nuclear medicine technologists receive 2–3 mSv per year. PET procedures may account for higher radiation exposures. Technologists who perform dose administration and patient positioning may account an effective dose 10–15 mSv per year. Air crew receives an annual effective dose of 3.1 mSv per year, which is some times higher than the medical occupation exposure level.

## BIOLOGICAL EFFECTS OF RADIATION

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Radiation can cause biological damages on cells either by indirect and direct action. If radiation falls on a human body, it produces moving



**FIG.13.2:** Radiation interaction with tissue

electrons. The electrons cause further ionization, excitation, resulting in chemical and molecular changes (Fig.13.2). Radiation can also produce free radicals, which are unpaired electrons that are chemically reactive. For example, radiation can interact with water molecule and produce hydroxyl (OH) and hydrogen (H) radicals. These free radicals can further interact with DNA, RNA or protein molecule and cause damage tissue. As a result, the constituents of the body cell will be affected, thereby impairing its normal functions. Since human body consists of 70% water, most of the radiation damage is caused by water, especially by OH radicals (indirect action). Presence of oxygen may enhance the effect of radical damage.

Chromosome breaks and aberrations are examples of biological damage caused by radiation. Chromosome aberrations include single break, ring formation, translocation, and dicentric, etc. It occurs spontaneously and scoring of human lymphocytes is used as biological dosimeter. The minimum whole body radiation that can be detected by this method is 250 mGy and 400 mGy for acute and chronic exposures.

Radiation may induce structural changes in a DNA molecule that includes (i) hydrogen bond break, (ii) molecular breakage, and (iii) inter and intramolecular cross linking. Hydrogen bond break disturbs the base pairs such as adenine-thymine, resulting in genetic changes. Molecular breakage may involve single strand break and double strand breaks. Single strand breaks are mostly repairable. There are enzymes within cells that will repair the radiation damage. Double strand break are irreparable and cause loss of base or change of base called *mutation*.

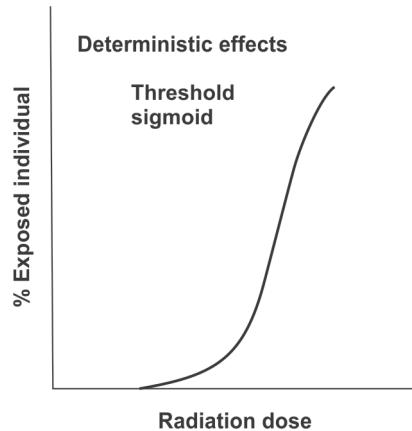
This results in chromosome aberrations, leading to carcinogenesis. High LET radiations are capable of causing double strand break. Since the DNA molecules carry the genetic code, its damage can result *hereditary effects*.

High dose of radiation can cause cell death (deterministic effect) or cellular transformation (stochastic effect). Lymphoid tissue and rapidly proliferating tissues, such as spermatids and bone marrow stem cells are relatively radiosensitive. Nerve cells are the least radiosensitive cells.

The effects of radiation on tissue depend on a number of factors such as type of radiation, dose, dose rate, dose fractionation, cell cycle and presence of radio-protectors and radiosensitizers, etc. An acute dose, delivered in a short time, is more harmful than a chronic dose, delivered over a period of time. The radiation effects which manifest themselves soon after the irradiation are called early effects. Those effects that manifest after a period of time are called late effects.

## DETERMINISTIC EFFECTS

The harmful effects of radiation may be classified as deterministic and stochastic effects. A deterministic effect is one in which “severity increases with increasing absorbed dose”. These effects appear at high doses  $> 0.5$  Gy and generally result from cell death. Deterministic effects are characterized by threshold dose, below which the effect does not occur (Fig. 13.3). Deterministic effects include skin erythema, epilation, organ atrophy, fibrosis, cataract induction, blood changes and reduction in sperm count, etc. Radiation accidents can cause, deterministic effects and unlikely to occur at medical X-ray procedures.

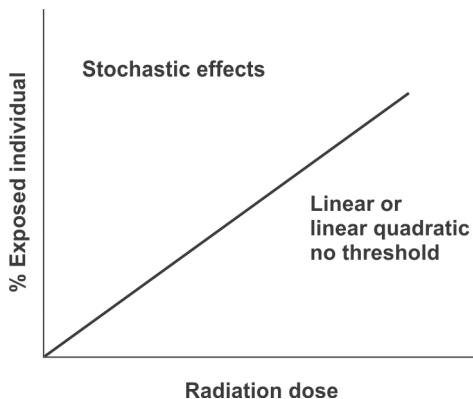


**FIG.13.3:** Deterministic effect and radiation dose, have threshold

## STOCHASTIC EFFECT

A stochastic effect is one in which “the probability of occurrence increases with increasing absorbed dose rather than its severity”. The stochastic effects are further classified into somatic and genetic effects. Stochastic

means random and the severity of stochastic effect is independent of the radiation dose. Radiation induced cancer and hereditary effects come under this category. It has no threshold dose, may occur even at low doses (Fig.13.4). Hence, the safety goal is to keep the exposures as low as reasonably achievable. Stochastic effect is the primary risk from low level radiations such as diagnostic X-rays. Stochastic risks are independent of sex and age at the time of exposure.



**FIG.13.4:** Stochastic effect and radiation dose

## ACUTE RADIATION SYNDROME

Whole body radiation exposures involving high level radiations delivered in shorter interval can cause acute radiation syndromes. It is a group of syndromes occurring in stages over a period of days to week. It includes hematopoietic syndrome, gastrointestinal syndrome and cerebrovascular syndrome. Soon after the radiation, early symptoms appear, known as prodromal radiation syndrome. Whole body doses  $> 100$  Gy may cause death in 24–48 hours, which is known as cerebrovascular syndrome. A dose level of 5–12 Gy may cause death in days, is called gastrointestinal syndrome. Dose of 2.5–5 Gy may cause death in weeks to months, called hematopoietic syndrome. The dose that causes 50% death over a specified time (60 days) is called lethal dose and is expressed in LD<sub>50/60</sub>, which is about 4 Gy for humans.

## RADIATION RISK

Risk is a probability that a given individual will incur a deleterious effect as result of a dose of radiation. When an individual is exposed to radiation, the expected risk includes somatic (cancer induction), genetic risk and fetal risk.

### Somatic Risk

The radiation effects, produced in an exposed individual during his life time are called somatic effects. Cancer induction is the largest risk of radiation exposure encountered in radiology. Bone marrow,

gastrointestinal tract mucosa, breast tissue, gonads and lymphatic tissue are most susceptible to radiation induced malignancy. Cancer risks are generally higher for children than for adults. Radiation may induce both benign and malignant tumors with latent period. The radiation induced cancer risk as per ICRP is 4% per Sv for low dose and dose rate, for working population.

### Genetic Risk

The radiation effects produced in the successive generation of the exposed individual are called genetic effects. The genetic effects are the result of radiation exposure to the gonads. There is no epidemiological evidence of genetic effect in humans. The current ICRP risk estimate for hereditary effects is 0.1% per Sv, which is based on the data for two generations.

### Fetal Risk/Pregnancy

The effect of radiation on embryo and fetus are, (i) lethal effects, (ii) malformations, and (iii) growth disturbances with malformation. The developmental period in utero has three stages namely, (i) pre-implantation, (ii) organogenesis, and (iii) fetal period. Pre-implantation is the most sensitive stage, which may cause lethal effects. The fetal risk depends on the gestation period of the pregnant women (Fig. 13.5). Mothers exposed to diagnostic X-rays in the third trimester, resulted in excess childhood leukemia. Diagnostic X-rays can increase the risk of childhood cancer by 40%. The excess absolute risk is about 6% per Gy. To avoid radiation induced congenital anomalies, an abortion may be advised only when dose exceed 100 mGy.

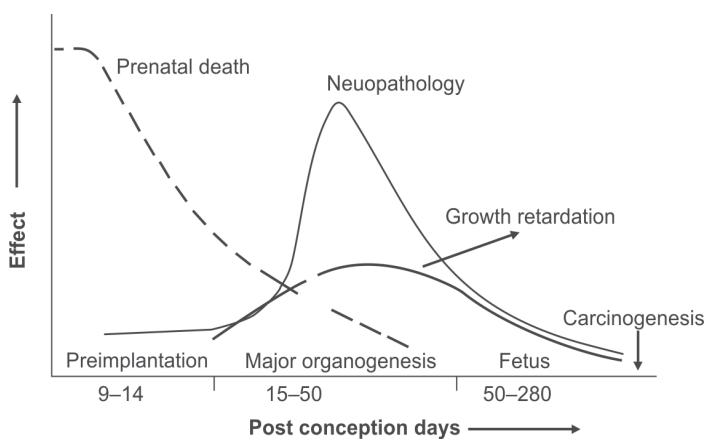


FIG.13.5. Radiation and fetal risk

## PRINCIPLES OF RADIATION PROTECTION

The aim of radiation protection is to prevent deterministic effects and to limit the probability of stochastic effects to levels deemed to be acceptable. This could be achieved, (i) by setting limits well below threshold dose of deterministic effects, and (ii) the probability of stochastic effects could be reduced by limiting exposures as low as reasonably achievable (ALARA). The whole radiation protection philosophy can be summarized as follows:

1. Justification of practice: No practice involving radiation exposures shall be adopted unless it produces a net positive benefit.
2. Optimization: Every effort shall be taken to reduce the dose as low as reasonable achievable, considering the clinical, social and economic factors.
3. Dose limits: The effective doses to the individuals shall not exceed the limits recommended by the commission.

The ICRP introduced the concept of permissible dose, defined as a dose of ionizing radiation that, in the light of present knowledge, is not expected to cause appreciable bodily injury to a person at any time during his lifetime. The recommended dose limits are given in Table 13.1.

**TABLE 13.1 Recommended dose limits (ICRP-60, 1990)**

Application	Occupational worker, mSv/year	Public, mSv/year
Effective dose	20*	1*
Eye lens	150	15
The skin	500	50
Hands and feet	500	50
Fetus, mean dose	2 after diagnosis	—

\*Averaged over any 5 consecutive years. 1 mSv = 100 mrem

The occupational dose limits exclude exposures from medical procedures and natural background. In the case of pregnant workers, the fetus is considered as a member of public. Pregnant radiation workers are monitored by a dosimeter worn on the abdomen under the lead apron. A measured dose of 2 mSv to the surface of the abdomen is normally considered equivalent to 1 mSv to the fetus. The dose limits for members of the public are generally 10 times lower than those for occupational exposure.

## **EFFECT OF TIME, DISTANCE, AND SHIELDING**

The three principal methods by which radiation exposures to persons can be minimized are: (i) Time, (ii) Distance, and (iii) Shielding as explained below:

### **Time**

The total dose received by a radiation worker is directly proportional to the total time spent in handling the radiation source. Lesser the time spent near the radiation source, lesser will be the radiation dose. As the time spent in the radiation field increases, the radiation dose received also increases. Hence, minimize the time spent in any radiation area. Techniques to minimize time in a radiation field should be recognized or practiced.

All radiation sources do not produce constant exposure rates. Diagnostic X-ray machines typically produce high exposure rates over brief time intervals. For example, chest X-ray produces a skin entrance exposure of 15 mR in millisecond (0.05), equivalent to 1080 R per hour. Hence, radiation exposure can be minimized by not energizing the X-ray tube, when personnel are nearer to the machine.

Nuclear medicine procedure produces lower exposure rate for extended periods of time. The exposure rate at 1 m from a patient injected with 10 mCi of Tc-99m, for bone scan is 0.5 mR per hour. It reduces to 0.25 mR per hour after 2 hours, due to decay and urinary excretion. Hence, both the knowledge of exposure rate and how it changes with time are the important elements in reducing personnel exposures.

The time spent near the radiation source can be minimized by understanding the task to be performed and the suitable equipment to complete them in short interval with safety. Hence, one has to plan the radiation procedure, practice the procedure without radiation and share the essential duties, to reduce radiation exposure. For example, fluoroscopy screening time should be kept short by the use of last frame hold facility, in addition to the use of foot switch.

### *Worked Example 13.1*

A radiographer is performing barium examination under fluoroscopy and the equipment is 'ON' for 5 minutes for each examination. The radiation level at the location of the radiographer is 60 mR/h. How many such procedures the radiographer can carry out per week?

The annual equivalent dose limit prescribed for the radiographer is (occupational worker) 20 mSv	= 2000 mrem » 2000 mR
The permitted weekly dose	= 2000 mR/50 weeks
	= 40 mR
Exposure rate at the location of radiographer	= 60 mR/h
	= (60/60) mR/min
The exposure in each procedure	= (60 mR/h) × 5 min
	= 5 mR
Hence, the number of procedures the radiographer can associate with in one week	= 40 mR/5 mR = 8

**Worked Example 13.2**

An operator is handling 5 mCi of I-131 source with 30 cm tongs. Within how much time the technician will receive the weekly permissible equivalent dose? (assume 1R = 1 rad,  $\Gamma_{20} = 2.18 \text{ R-cm}^2/\text{mCi-h}$  for I-131)

Exposure level at 30 cm from

$$\begin{aligned} 5 \text{ mCi of I-131 source} &= 2.18 \times 5 \text{ mCi}/(30^2) \\ &= 0.012 \text{ R/h} \\ &= 12 \text{ mR/h} \end{aligned}$$

Weekly permissible exposure

$$\begin{aligned} (2000 \text{ mR}/50 \text{ week}) &= 40 \text{ mR} \\ \text{Allowed time of work} &= 40 \text{ mR}/12 \text{ mR/h} \\ &= 200 \text{ minutes} \end{aligned}$$

**Distance**

Radiation intensity (exposure rate) from a point source decreases with distance, due to divergence of the beam. It is governed by the inverse square law, which states that the exposure rate from a point source of radiation is inversely proportional to the square of the distance. If the exposure rate is  $X_1$  at distance  $d_1$ , then the exposure rate  $X_2$  at another distance  $d_2$  is given by

$$X_2 = X_1(d_1/d_2)^2$$

Doubling the distance from the X-ray source decreases the X-ray beam intensity by a factor of 4. *Larger the distance, lesser will be*

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*the radiation dose.* This relationship is valid for point sources only, whose dimensions are very small compared to distance under consideration. Thus, the relationship is not valid near ( $<1\text{ m}$ ) a patient injected with radioisotopes, since the exposure rate decreases less rapidly than inverse square law.

In diagnostic radiology at 1 m from a patient, the scattered radiation is about 0.1–0.15% of the intensity of the primary beam. Hence, all personnel should stand as far away as possible during X-ray procedures. Personnel should stand at least 2 m from the X-ray tube and the patient and behind the shielded barrier or out of the room, whenever possible.

Imaging rooms should be designed to maximize the distance between the source and control console. Unshielded radiation sources never are manipulated by hand. Tongs or other handling devices are used to increase the distance between source and hand.

### Worked Example 13.3

The exposure rate from a fluoroscopic X-ray machine is 5 R/min at 50 cm. What would be the exposure rates at (i) 40 cm, and (ii) 60 cm?

$$X_1 = 5 \text{ R/min}, D_1 = 50 \text{ cm}, D_2 = 40 \text{ cm}, X_2 = ?$$

$$X_2 = [X_1 \times (D_1)^2]/(D_2)^2 = [5 \text{ R/min} \times (50 \text{ cm})^2]/(40 \text{ cm})^2 \\ = 7.81 \text{ R/min}$$

$$X_1 = 5 \text{ R/min}, D_1 = 50 \text{ cm}, D_2 = 60 \text{ cm}, X_2 = ?$$

$$X_2 = [5 \text{ R/min} \times (50 \text{ cm})^2]/(60 \text{ cm})^2 \\ = 3.47 \text{ R/min}$$

### Shielding

When maximum distance and minimum time do not ensure an acceptably low radiation dose, adequate shielding must be provided, so that radiation beam will be sufficiently attenuated. The material that attenuates the radiation exponentially is called shield and the shield will reduce exposure to patients, staff and the public. If  $I_0$  is the intensity of radiation at a point without shield and  $I$  is the intensity with a shield of thickness  $t$ , then  $I = I_0 e^{-\mu t}$ , where  $\mu$  is the linear attenuation coefficient of the shielding material. *Larger the shielding thickness, lesser the radiation exposure.*

The thickness of the shielding material that reduces the intensity to half is called half value thickness (HVT) and it is given by the relation,

$$\text{HVT} = \frac{0.693}{\mu}$$

The reduction factor offered by  $n$  HVT is  $2^n$ . For diagnostic X-rays, the HVT for soft tissue ranges from 2.5–3.0 cm. Tenth value thickness (TVT) is the thickness of material that attenuates the radiation beam by 90%. The corresponding reduction factor for  $n$  number of TVT is  $10^n$  and  $1 \text{ TVT} = 3.32 \text{ HVT}$ . Both HVT and TVT are used in room shielding calculations.

X and gamma rays undergo exponential attenuation in the shielding material. This means that even a large shielding material will not attenuate the radiation to zero intensity. However, optimal shielding thickness is required to bring down the radiation level below the permissible limit. Brick, concrete are used as shielding material for construction of X-ray room barriers. On the other hand, lead is used as protective material, in lead apron, thyroid shield and gonad shield.

## CALCULATION OF WORKLOAD

The calculation of barrier shielding require the understanding of five factors, namely, workload, use factor, occupancy factor, distance, and radiation exposure level.

### Workload (W)

Workload is a measure of expected exposure levels obtained from the patient load and machine ON time. It is expressed in mA-min per week in diagnostic radiology (< 500 kVp), which can be obtained by multiplying the maximum mA with beam on time in minutes per week.

#### *Workload for Diagnostic X-rays*

The workload of a hospital with 20 patients per day, 3 films per patient, and 50 mAs per film will be calculated as follows:

$$\begin{aligned} W &= (20 \text{ patients/day}) \times (5 \text{ days/week}) \times (3 \text{ films/patient}) \\ &\quad (5 \text{ mAs/film}) \times (1 \text{ min/60s}) \\ &= 25 \text{ mA-min/week} \end{aligned}$$

In general, higher kVp settings decrease the workload. This is due to (i) increase of output (mR/mAs) as the kVp increases and (ii) less attenuation of the incident beam by the patient which reduces the mAs, at higher kVp. The workload values for various types of radiographic rooms are given in Table 13.2 (NCRP Report No. 49). These are overestimated values, as it is based on slower (speed 100) film-screen receptors.

**TABLE 13.2** Workload for diagnostic X-ray

Procedure	Patient load/day	W (mA min/week), 100 kVp	W (mA min/week), 125 kVp
Chest, 3 films/patient	60	250	150
Fluoroscopy	24	750	300
General radiography	24	1000	400
Special procedures	8	700	280

### Workload for CT Scan

In the case of CT scan, all the walls in the room are secondary barriers, and the detector plays the role of primary barrier. The workload of a CT scan is calculated from average number of patients per week, fraction of head verses body scans, and the average mAs per patient.

The workload of a head CT scan having 20 abdominal scan per day, 30 slices per scan with 250 mAs per slice can be calculated as follows. 50% of the scans are pre- and postcontrast and 50% are without any contrast (assume 1.5 studies/patient).

$$\begin{aligned} W &= (20 \text{ patients/day}) \times (5 \text{ day/week}) \times (30 \text{ slices/study}) \times \\ &\quad (1.5 \text{ studies/patient}) \\ &= 4500 \text{ slices/week.} \end{aligned}$$

### Use Factor (U)

It is the fraction of the operating time during which the radiation under consideration is directed towards a particular barrier (wall) per week. The use factor for different barriers is as follows:

- Primary barrier:  $U = 0$  to  $1$
- Secondary barrier:  $U = 1$
- Floor:  $U = 1$
- Walls:  $U = 1/4$
- Ceiling:  $U = 1/4$  to  $1/2$

### Occupancy Factor (T)

The occupancy factor (T) relates to the amount of time rooms adjacent to the treatment room are occupied. An area below ground would have no occupancy at all and therefore, T would equal zero. Areas that are intermittently occupied, such as corridors, would have a slightly greater occupancy and an area, such as an office even greater. It is the fraction of the operating time during which the area of interest is occupied by the individual (Table 13.3).

**TABLE 13.3** Occupancy factors (NCRP 49)

Type of occupancy	Occupancy factor (T)
Office, reception, shops, children play areas, staff rooms, control room	1 (Full occupancy)
Corridor	1/4 (partial occupancy)
Toilet, bathrooms, outside areas with seating, store rooms	1/16 (occasional occupancy)

### **Distance (d)**

It is the distance in meters from the radiation source to the area to be protected. Inverse square law is assumed for both primary and stray radiation.

### **Radiation Exposure Level (X)**

The radiation exposure (mR per week) at a given location can be calculated by knowing the workload, X-ray tube output at 1 m in mR/mA-min and distance of that location from X-ray tube. It should be accounted for primary, scattered and leakage radiation, if any, at that location. Over all, the radiation exposure (X) per week at a location of distance d is given by

$$X = WUT \times \text{tube output @ 1 m, in mR/mA-min} \times (1/d^2)$$

The required wall thickness can be calculated by using the relation

$$\text{Thickness} = \frac{0.434 \times \ln X}{P} \times \text{TVT}$$

where, P is the permissible dose equivalent limit and it is 2 mR/week for general public and 40 mR/week for radiation worker. TVT corresponds to the material (brick or concrete) to be used in the construction of wall.

## **GOOD WORK PRACTICES IN DIAGNOSTIC RADIOLOGY**

### **X-ray Examination**

Any X-ray examination should be prescribed only after a critical evaluation of the patient's condition in order to avoid unnecessary exposures. No fluoroscopic examination should be conducted, if the required information can be obtained from radiography.

### **Quality Assurance**

Every new X-ray unit, shall be subjected to quality assurance tests before patient use. It should be used unless adequate protection and operational safety is confirmed by the Radiological Safety Officer. Only qualified X-ray technologists will be allowed to handle X-ray equipment.

### **Equipment Operation**

Personnel monitoring devices shall be used by all radiation workers, while on duty. Before making an exposure, the doors of the X-ray room must be closed. The X-ray beam should not be directed towards the windows, control panel or dark room wall. Medical students/trainees must not be allowed to operate an X-ray unit. When performing portable examinations, the operator should stand at least 2 m away from the patient.

### **Protective Shield**

All radiation workers must wear lead apron of 0.5 mm thickness, which reduces radiation exposure by a factor of 10. Use of leaded glass, lead gloves must be encouraged in fluoroscopy type of work. Repeat X-rays must be avoided in order to reduce patient dose. Gonad shield, eye shield and thyroid shields should be used, to protect the patient organ, during radiography.

### **Field Area**

Minimal field size to cover the patient volume should be used. Field size reduction reduces the scatter thereby reducing the dose to adjacent organs. The scatter incident on the detector also decreases, resulting in improved image contrast. Hence, the rule of thumb is “use smallest, possible field size and good collimation”.

### **Source to Object Distance**

Higher the source to object distance (SOD) and source to image distance (SID), lesser the patient dose. Increase of SOD/SID, reduces beam divergence, in turn reduce the volume of patient irradiation. This will enable us to decrease the integral dose. Increased SOD also facilitates reduction of patient exposure due to tube leakage, since the tube is away from the patient.

In radiography and fluoroscopy with stationary X-ray equipment, the SOD should be not less than 45 cm. When the SID is less than 100 cm, the quality of the diagnostic information becomes poorer; therefore longer SID has clinical advantages. Chest radiography should be performed with a SID of at least 120 cm. In the case of C-arm units,

fixed SID is used, therefore increase of SOD is the only way of reducing patient dose. In fluoroscopy, the minimum distance between source and the patient must be not less than 30 cm.

### **Occupancy in the Room**

Only persons whose presence is necessary should be in the imaging room during the exposure and overcrowding should be avoided. All such persons must be protected with lead aprons/shields. The X-ray room shall be kept closed during the radiation exposure.

### **Assistance to Patients**

Holding of children or infirm patients for X-ray examination shall be done only by an adult relative or escort of the patient. Hospital personnel should not hold the patients during imaging procedure. No person should routinely hold patients during diagnostic examinations, certainly not those who are pregnant or under the age of 18 years. Such persons shall be provided with protective aprons and gloves. In no instance shall the holder's body be in the useful beam, and should be as far away from the primary beam as possible. In no case shall the film of X-ray tube be held by hand.

### **Patient Motion**

Patient motion may cause motion artifacts, which may increase repeat X-rays and patient dose. To reduce patient motion (i) short exposure times, (ii) use of immobilization or sedation, (iii) entertainment, or distracting devices should be applied and adopted. Immobilization devices prevent movement of children during exposure.

### **Pregnant Women**

Radiological examination of the lower abdomen and pelvis of a pregnant woman must be conducted only when considered absolutely essential. One or two X-rays may be performed, if the last mensurel period (LMP) is within 28 days. In all other X-ray examinations, the abdomen and the pelvis must be covered with a protective shield.

### **Log Book**

Each X-ray equipment must have a separate log book, which provides information about the equipment manufacturer, model, serial number, date of purchase, cost, repair, down time, etc.

### **Records**

Records of all radiological examinations should be maintained. Reports and radiographs should be given to the patient for future reference.

## **PLANNING OF DIAGNOSTIC X-RAY INSTALLATIONS**

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### **GENERAL GUIDELINES**

#### **Room Size**

The room housing an X-ray unit shall be not less than 18 m<sup>2</sup> for general purpose radiography and conventional fluoroscopy equipment. The size of the room housing the gantry of the CT unit shall not be less than 25 m<sup>2</sup>. Also, not more than one unit of any type shall be installed in the same room, and no single dimension of these X-ray rooms shall be less than 4 m. In the case of mammography, the room size shall be not less than 10 m<sup>2</sup>, and no single dimension of the room shall be less than 3 m.

#### **Wall Thickness**

If the X-ray installation is located in a residential complex, it shall be ensured that:

1. Walls of the X-ray rooms on which primary X-ray beam falls are not less than 35 cm or 14 inch thick brick or equivalent.
2. Walls of the X-ray room on which scattered X-ray fall are not less than 23 cm or 9 inch thick brick or equivalent.
3. There is a shielding equivalent to at least 23 cm or 9 inch thick brick or 2 mm lead in front of the door(s) and windows of the X-ray room to protect the adjacent areas, either by general public or not under possession of the owner of the X-ray room. The density of the normal masonry brick is considered as 1.6 g/cc.
4. The ceiling must have a thickness of concrete (density 2.35 g/cc), not less than 6 inch or 13.5 cm.

#### **Control Room**

For equipment operating at 125 kV or above, should have a separate control room, and provided with appropriate shielding, direct viewing (1.5 mm lead equivalence) and oral communication facilities between the operator and the patient. The X-ray units operating below 125 kV in diagnostic radiology are exempted from the above class. In such a case, the control should be behind a mobile protective barrier of adequate thickness.

#### **Doors**

Doors are lined with 2 mm thick lead sheet with proper overlapping at the joint and junction and wall of 9 inch thickness of brick and ceiling of 6 inch of concrete.

### **Viewing Window**

Lead glass of suitable dimensions are provided as viewing windows with 1.5 mm thick lead equivalent.

### **Mobile Protective Barrier**

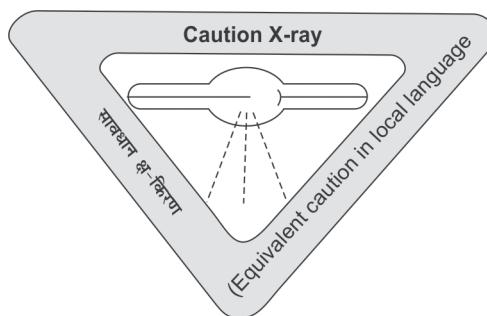
Control panel should be kept behind the mobile protective barrier (MBP) of thickness 2 mm lead equivalence.

### **Darkroom**

The darkroom should be located in such a way that the primary beam is not directed on it. Appropriate shielding must be provided for the darkroom to ensure that undeveloped X-ray films stored in it, will not be exposed to more than an air kerma rate of  $10 \mu\text{Gy}$  per week.

### **Placard**

A warning placard as shown in the Figure 13.6, must be posted outside the room entrance or door.

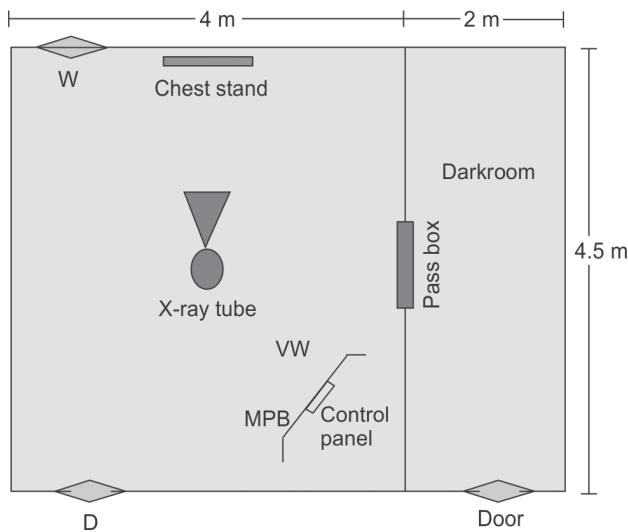


**FIG.13.6:** X-radiation warning placard

### **GENERAL RADIOGRAPHY INSTALLATION**

These X-ray units are operated in the range of 50–150 kVp. Walls that are irradiated directly by the X-ray beam are primary barriers. Hence, additional shielding must be provided for the wall behind the chest stand. Provisions are made to observe and communicate with the patient on the table. The mobile protective barrier with lead shield must be a permanent/mobile one with 2.1 m height. The viewing window at the mobile protective barrier must be 45 × 45 cm size and centered.

A typical model plan is shown in the Figure 13.7.



**FIG. 13.7:** Model plan for a general radiography room

### FLUOROSCOPY INSTALLATION

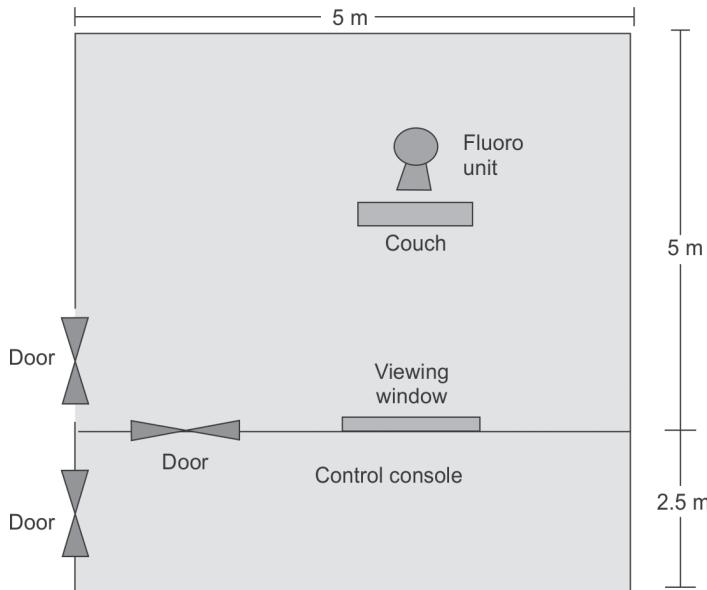
Fluoroscopic imaging systems are usually operated at potentials ranging from 60 to 120 kVp. A primary barrier is incorporated into the fluoroscopic image receptor. Therefore, fluoroscopic unit room design considers only secondary protective barriers against leakage and scattered radiations. Most fluoroscopic X-ray imaging systems also employ radiography. The shielding requirements for such a room are based on the combined workload of both fluoroscopy and radiography. A typical model plan is shown in the Figure 13.8.

### MAMMOGRAPHY INSTALLATION

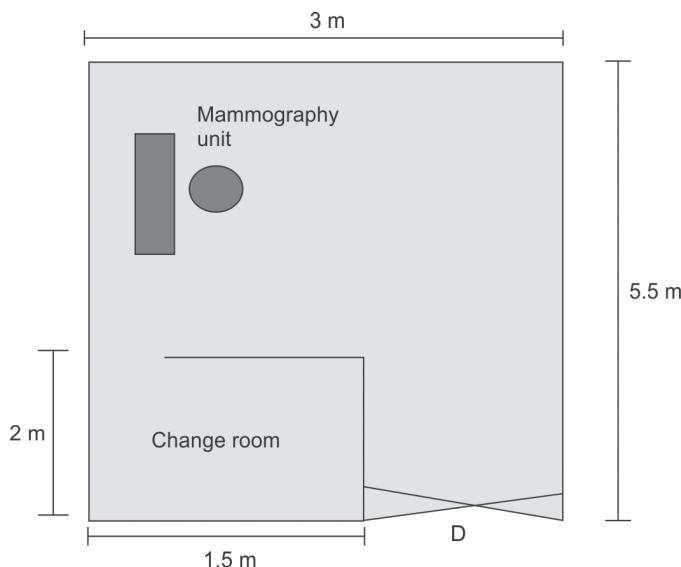
Mammography units are typically operated between 25–30 kVp. The walls are constructed with bricks or gypsum wall board. Adequate protective barrier of lead acrylic or lead glass are incorporated into dedicated mammography units. Gypsum wall board may contain voids and non uniform areas. Hence, higher thickness of gypsum wall board is recommended than that calculated. A typical model plan is shown in Figure 13.9.

### COMPUTED TOMOGRAPHY INSTALLATION

Computed tomography (CT) employs a collimated X-ray fan-beam that is intercepted by the patient and by the detector array. Consequently,

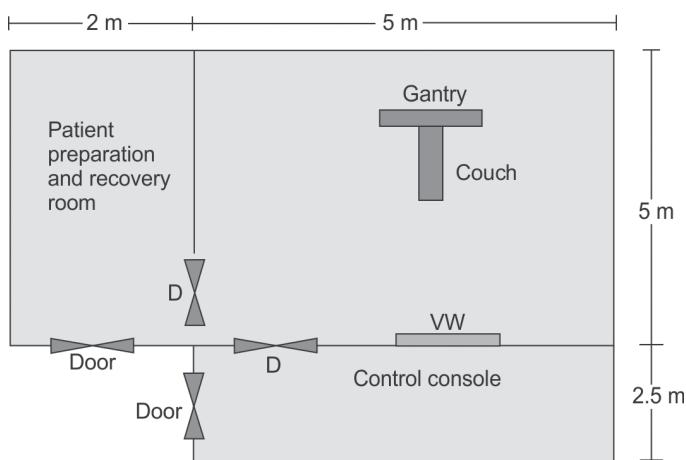


**FIG. 13.8:** A typical layout of a fluoroscopy installation



**FIG. 13.9:** Model plan for a mammography room

only secondary radiation is incident on protective barriers. The operating potential, typically in the range of 80–140 kVp, as well as the workload are much higher than for general radiography or fluoroscopy. Due to large amount of secondary radiation, floors, walls and ceilings need special consideration. Additionally, scattered and leakage radiations from CT systems are not isotropic. The radiation levels in the direction of the gantry are much less than the radiation levels along the axis of the patient table. A typical model plan is shown in Figure 13.10.



**FIG.13.10:** Model plan for a computed tomography room

## QUALITY ASSURANCE IN DIAGNOSTIC RADIOLOGY

Quality assurance (QA) describes a set of policy that is designed to control and maintain the standard of quality of patient care. The goal of QA in diagnostic radiology is to obtain optimal image with minimum radiation dose and at minimum cost. The QA program begins with acceptance/performance evaluation tests, during installation of the X-ray unit and are repeated at regular intervals or after major repair. In general, the mechanical characteristics, the control panel display/indicators and the tube housing details are checked initially and it is followed by the set of tests. The test vary with type of X-ray machines as given below.

## Acceptance/performance test for general purpose and interventional radiology diagnostic X-ray machines

Sl. No.	Parameters tested	Tolerance
1.	Correspondence between X-ray field and image reception area	$  c_1   +   c_2   +   d_1   +   d_2   \geq 4\% \text{ of FFD}$
2.	Central Beam alignment	< 1.5°
3.	Accuracy of operating potential	± 5 kV
4.	mA/mAs linearity	Coefficient of linearity < 0.1
5.	Timer linearity	Coefficient of linearity < 0.1
6.	Accuracy of irradiation time	≤ ± (10%)
7.	Reproducibility of radiation output	CoV ≤ 0.05
8.	Total filtration	<ul style="list-style-type: none"> <li>• For Less than 70 kVp Minimum total filtration 1.5 mm Al</li> <li>• For 70 &lt; kVp ≤ 100 Minimum total filtration 2.0 mm Al</li> <li>• For kVp &gt; 100 Minimum total filtration 2.5 mm Al</li> </ul>
9.	Exposure rate at tabletop	Without AEC mode: ≤ 5 cGy/min With AEC mode: ≤ 10 cGy/min
10.	Low contrast resolution	3.0 mm hole should be visible
11.	Spatial resolution	1.5 lp/mm should be visible
12.	On-position radiation leakage level from X-ray tube housing measured at _____ kV at _____ mA	< 1 mGy in one hour

## Acceptance/performance test for mammography X-ray machine

Sl. No.	Parameters tested	Tolerance
1.	Accuracy of operating potential	± 1
2.	Accuracy of timer	< 10%
3.	Linearity of tube current (CoL)	± 0.1
4.	Reproducibility of output (CoV)	CoV ≤ 0.05
5.	Radiation Leakage level from X-ray tube housing _____ kV at _____ mA	< 1 mGy in one hour
6.	Total filtration	0.3 mm Al ≤ HVL ≤ 0.37 mm Al at 28 kVp

### Acceptance/performance test for Computed tomography scanner

Sl. No.	Parameters tested	Tolerance
1.	Alignment of table to gantry	$\pm 5$ mm
2.	Gantry tilt	$\pm 3^\circ$
3.	Table indexing accuracy (mm)	$\pm 2$ mm
4.	Slice thickness (mm)	$\pm 1$ mm
5.	Accuracy of operating potential (kV)	$\pm 2$ kV
6.	Accuracy of timer	< 10%
7.	Linearity of tube current (CoL)	$\pm 0.1$
8.	Reproducibility of output (CoV)	CoV < 0.05
9.	Radiation dose test CTDI –(mGy/100 mAs) at 120 kV	$\pm 20\%$
10.	Low contrast resolution	5.0 mm at 1% contrast
11.	High contrast resolution	3.12 lp/cm
12.	Radiation leakage level from X-ray tube housing (140 kV at 100 mA)	< 1 mGy in one hour

## ATOMIC ENERGY REGULATORY BOARD

The Atomic Energy Regulatory Board (AERB), was constituted on November 15, 1983, by the President of India by exercising the powers conferred by section 27 of the Atomic Energy Act 1962 (33 of 1962). It is an apex body that regulates the use of ionizing radiation in the country. The mission of AERB is to ensure that the use of ionizing radiation and nuclear energy does not cause undue risk to health and environment. The major objectives are to develop and publicize specific codes and guides, which will deal with the radiation safety aspects of various applications of radiations. It will also issue authorization related to site, design, manufacture, construction, commissioning, operation, maintenance, and decommissioning and disposal of radioactive sources. The AERB implement the safety provisions by secondary legislation, viz, Atomic Energy (Radiation Protection) Rules-2004, which provides necessary regulatory infrastructure for effective implementation of radiation protection program in India.

## REGULATORY REQUIREMENTS

### Design Certification

Every medical diagnostic X-ray equipment shall meet the design safety specifications stipulated in the AERB safety code. The manufacturer/

vendor shall obtain design certification from the competent authority prior to manufacturing the X-ray equipment.

### **Type Approval/No Objection Certificate**

Prior to marketing the X-ray equipment the manufacturer shall obtain a type approval certificate from the competent authority for indigenously made equipment. For equipment of foreign make, the importing/vending agency shall obtain a No Objection Certificate (NOC) from the competent authority, prior to marketing the equipment. Only type approved and NOC validated equipment shall be marketed in the country.

### **Layout Approval and Registration**

Once the X-ray unit is installed, it should be registered with AERB along with lay out approval. In the case of CT and cathlab, one has to get licensee for operation along with lay out approval. For registration and license, quality assurance tests and nomination of RSO is mandatory.

### **Inspection of X-ray Installations**

The diagnostic X-ray installations shall be made available by the employer/owner for inspection, at all reasonable times, to the competent authority or its representative, to ensure compliance with the safety code.

### **Decommissioning of X-ray Installations**

Decommissioning of X-ray equipment shall be registered with the competent authority immediately by the employer/owner of the equipment.

### **Certification of RSO**

Any person accepting assignment to discharge the duties and functions of RSO in diagnostic X-ray installations shall do so only after obtaining certification from the competent authority for the purpose. Such certification shall be granted on the basis of adequacy of the person's qualification, experience and testing/survey/dosimetry equipments availability.

### **Certification of Service Engineers**

Only persons holding valid certificate from the competent authority shall undertake servicing of X-ray equipment. Certification shall be granted on the basis of qualifications, training, and experience and safety record of such person and availability of servicing facilities.

## **RESPONSIBILITIES OF THE EMPLOYER (RULE 20)**

1. Every employer shall:
  - i. Ensure that provisions of these rules are implemented by the licensee, RSO and other workers.
  - ii. Provide facilities and equipment to the licensee, RSO and other worker to carry out their functions effectively.
  - iii. Obtain dose records and health surveillance report of the workers from their former employer.
  - iv. Provide dose records and health surveillance reports of the worker to the new employer.
  - v. Furnish to each worker dose records and health surveillance reports of the worker annually.
  - vi. Inform the competent authority if the licensee or the RSO or any worker leaves the employment.
  - vii. Arrange for health surveillance of workers.
2. The employer shall be the custodian of radiation sources in his possession and shall ensure physical security of the sources at all times.
3. The employer shall inform the competent authority, within twenty four hours, of any accident involving a source or loss of source of which he is the custodian.

## **RESPONSIBILITIES OF THE LICENSEE (RULE 21)**

1. The licensee is responsible for the implementation of terms and conditions of the licensee.
2. The licensee shall comply with the surveillance procedures, safety codes and safety standards, specified by the competent authority.
3. Licensee shall establish written procedures and plans for controlling, monitoring and assessment of exposure for ensuring adequate protection of workers, members of the public and the environment and patients.
4. The licensee shall comply with the provision of rules for safe disposal of radioactive waste.
5. The licensee shall:
  - i. Maintain records of workers.
  - ii. Arrange for preventive and remedial maintenance of radiation protection equipment, and monitoring instruments.
  - iii. Investigate excessive radiation exposure and maintain records of such investigations.

- iv. Inform competent authority about the occurrence, investigation and follow-up actions, including steps to prevent future occurrences.
  - v. Carry out physical verification of radioactive material periodically and maintain inventory.
  - vi. Inform appropriate law enforcement agency in the locality of any loss of source.
  - vii. Inform the employer and the competent authority of any loss of source.
  - viii. Investigate and inform the competent authority of any accident involving source and maintain record of investigations.
  - ix. Verify the performance of radiation monitoring systems, safety interlocks, protective devices and any other safety systems in the radiation installation.
  - x. Prepare emergency plans in consultation with RSO.
  - xi. Conduct quality assurance tests of structures, systems, components and sources and related equipment.
  - xii. Advise the employer about the modifications in working condition of a pregnant worker.
  - xiii. Inform the competent authority if the RSO or a worker leaves the employment.
  - xiv. Inform the competent authority when he leaves the employment.
6. The licensee shall ensure that the workers are familiarized with contents of the relevant surveillance procedures, safety standards, safety codes, safety aides and safety manuals issued by the competent authority and emergency response plans.

## **RESPONSIBILITY OF RADIOLOGICAL SAFETY OFFICER (RULE 22)**

1. Radiological Safety Officer (RSO) shall provide advice and assistance to the employer and licensee on radiation safety.
2. RSO shall:
  - i. Carry out measurements and analysis on radiation and radioactivity levels in the controlled area, supervised area and maintain records of the same.
  - ii. Investigate any situation that could lead to potential exposures.
  - iii. Advise the employer to ensure regulatory constraints and the terms and conditions of the license, safe storage and movement of radioactive material within the radiation installation, to initiate

suitable remedial measures in any situation that leads to potential exposures, and regular measurements and analysis of radiation and radioactivity levels in and around the installation.

- iv. Report all hazardous situations with details and remedial actions taken, to the employer and licensee for reporting to the competent authority.
  - v. Conduct quality assurance tests on structures, systems, components and sources.
  - vi. Ensure periodic calibration of monitoring instruments.
3. RSO should assist the employer in (i) instructing the workers about hazards of radiation and safety and good work practices, (ii) safe disposal of radioactive wastes, (iii) developing emergency response plans to deal with accidents and maintaining emergency preparedness.
  4. RSO should advise the licensee on (i) modifications in working condition of a pregnant worker, (ii) the safety and security of radioactive sources. He should furnish to the licensee and the competent authority the periodic reports on safety status of the radiation installation. He should inform the competent authority, whenever he leaves the employment.

### **RESPONSIBILITIES OF WORKER (RULE 23)**

1. Every worker shall observe safety requirements and follow safety procedures and instructions. He should not do any work that is harmful to him, co-workers, installation and public.
2. Worker should inform the employer about his previous occupations. He should use protective equipment, radiation monitors and personnel monitoring devices. He should inform the licensee and the RSO, about accident or any potentially hazardous situation.
3. Female worker, once become pregnant, she should inform the same to the licensee and Radiological Safety Officer.

### **HEALTH SURVEILLANCE OF WORKERS (RULE 25)**

1. Every employer shall provide the services of a physician with appropriate qualifications to undertake occupational health surveillance of classified workers.
2. Every worker, initially on employment, and classified worker, thereafter at least once in three years as long as the individual is employed, shall be subjected to the following: (a) general medical

- examination, and (b) health surveillance to decide on the fitness of each worker for the intended task.
3. The health surveillance shall include: (a) special tests or medical examinations as specified by order by the competent authority, for workers who have received dose in excess of regulatory constraints, and (b) counseling of pregnant workers.

# 14

# Ultrasound Imaging

## INTRODUCTION

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Ultrasound describes sound waves of frequencies exceeding the range of human hearing ( $> 20$  kHz). Audible sound range is 15–20,000 Hz, the sound  $< 15$  Hz is called infrasound. In medicine, the ultrasound energy and the acoustic properties of the body, produce images from stationary and moving bodies. Diagnostic ultrasound uses 1–20 MHz frequency, and the velocity depends on the nature of medium through which it travels and independent of frequency. It is not an electromagnetic radiation, but undergoes reflection and refraction at interfaces. The reflection from tissue is called echo, which forms the image.

## BASICS OF ULTRASOUND

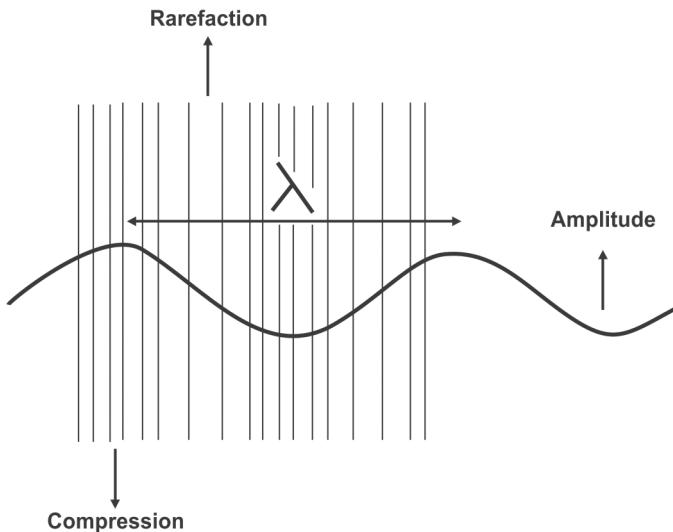
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### PROPAGATION OF SOUND

Ultrasound (US) is produced by a transducer by piezoelectric effect and US pulse is passed in straight line. Sound is a mechanical energy that propagates through an elastic medium in the form of waves with compression and rarefaction (Fig. 14.1). It is a longitudinal wave (sinusoidal) and wavelength ( $\lambda$ ) is the distance between successive wave crests. Frequency (f) is the number of cycle per second (hertz) and one hertz (Hz) = 1 cycle/second. Period (T) is the time taken for one complete cycle and it is equal to  $1/f$ . The velocity of sound (C), wavelength and frequency are related as follows:

$$C = \lambda f, \text{ m/second}$$

The velocity,  $C = \sqrt{B/\rho}$ , where B is the bulk modulus (measure of stiffness of the medium), and  $\rho$  is the density. Velocity is inversely  $\propto$  to compressibility and it depends on temperature of the medium.



**FIG. 14.1:** Propagation of ultrasound as longitudinal wave with compression and rarefaction

Sound wave travels faster in solids and slower in gases. Average velocity in soft tissue is  $1540 \text{ ms}^{-1}$  ( $7 \mu\text{s}$  to travel 1 cm). It is higher in bone and metal, lower in lung and air (compressible). Difference in velocity (change of  $\lambda$ ) at tissue boundaries is the basis for contrast in ultrasound image. Intensity of ultrasound is measured in watts per  $\text{mm}^2$  which is proportional to the square of the amplitude. Relative sound intensity is measured in a logarithmic scale, and the unit is decibel (dB).

### INTERACTION OF ULTRASOUND WITH MATTER

Ultrasound undergoes reflection, refraction, scattering and absorption in matter, depends upon the acoustic properties of matter. Reflection occurs at tissue boundary, where there is a difference in acoustic impedance. Reflection may be a boundary reflection or tissue reflection and tissue reflection mostly gives scattering. Refraction refers the change in direction of the transmitted ultrasound energy. Scattering occurs by both reflection and refraction (small particles). Absorption refers the conversion of acoustic energy into thermal energy in the medium. The acoustic impedance ( $Z$ ) is the product of density ( $\rho$ ) and speed ( $c$ ) of sound, i.e.

$$Z = \rho \times c, \text{ kg/sq m/s (Rayl)}$$

It depends on density and elasticity of the interface and independent of frequency. The acoustic impedance can also be related to modulus of elasticity ( $E$ ) as follows:

$$Z = \sqrt{E \times \rho}, \text{ Rayl}$$

The acoustic impedance of various body tissues are given in Table 14.1

**TABLE 14.1** Acoustic impedance of various body tissues

Material	Velocity (msec <sup>-1</sup> )	Acoustic impedance, Rayl
Air	330	$0.0004 \times 10^6$
Fat	1450	1.34 "
Blood	1560	1.65 "
Muscle	1600	1.71 "
Bone	3300	7.8 "
Metals	> 4000	> 30.0 "

### Reflection

As sound wave travels from medium of low  $Z$  to high  $Z$ , the reflected wave experience a phase shift of  $180^\circ$  (negative sign), independent of frequency, and has directional dependence (Fig. 14.2). Difference between  $Z$ , determines the amount of reflected energy at the interface. The angle of incidence ( $\theta_i$ ) = angle of reflection ( $\theta_r$ ), and it obeys Snell's law

$$\sin \theta_i / \sin \theta_r = C_1 / C_2$$

where,  $C_1$  and  $C_2$  are the velocity of sound in medium 1 and medium 2, respectively. The % of reflected intensity depends on the angle of incidence. The intensity of reflection coefficient ( $R$ ) is given by the relation:

$$R = [(Z_2 - Z_1) \div (Z_2 + Z_1)]^2$$

When the beam is perpendicular to the tissue boundary, sound is returned back as an echo. As angle of incidence increases, less likely the reflection and no reflection is detected when the angle of incidence is  $> 3^\circ$ . When  $Z_1 = Z_2$ , the transmission is 100% and the two medium are said to be acoustically matching. This is called an acoustic window; tissue acts as conduit and allows US transmission, e.g. lung.

In interface, such as air or gas, nearly 100% US is reflected, and there is no transmission. They cast shadow and underlying organs cannot be imaged and hence US imaging impossible in such cases,

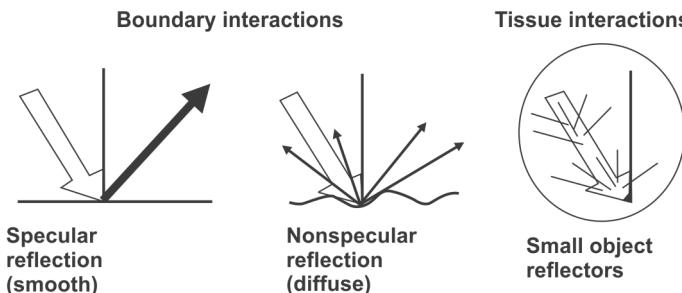


FIG.14.2: Boundary and tissue reflections

e.g. lung. Similarly, transducer-air interface gives 100% reflection and hence acoustic coupling gel is used between transducer and skin, to eliminate air pockets. However, only 20% of US is transmitted and to improve transmission, the matching layer is used. In bone-tissue interface, there is 30% reflection, and 70% transmission and hence bone imaging is difficult, appear as areas of void and shadowing.

#### Worked Example 14.1

Calculate the % of reflection and transmission in a muscle–air interface,  $Z_1 = 1.71 \times 10^6$  Rayl, and  $Z_2 = 0.0004 \times 10^6$  Rayl.

The reflection coefficient,  $R = [(1.71 - 0.0004)/(1.71 + 0.0004)]^2 = 0.998$  or 99.8%. The transmission coefficient,  $T = 1 - 0.998 = 0.002$  or 0.2%. Thus, in muscle-air interface, about 99% is reflected and < 1% is transmitted.

Actual intensity reflected at the boundary is calculated by multiplying the incident intensity (e.g. 40 mW/Sq cm) with reflection coefficient:

$$\text{i.e. } 40 \text{ mW/sq cm} \times 0.998 = 39.9 \text{ mW/sq cm.}$$

#### Refraction

Refraction describes the change in direction of the transmitted US energy at tissue boundary, when the beam is non-perpendicular. Frequency do not change, but change of velocity may occur due to change of  $\lambda$ . Refracted beam have longer wavelength, in denser medium and it obeys Snell's law;

$$\sin \theta_t / \sin \theta_i = C_2 / C_1$$

where,  $\theta_t$  is the angle of transmission. If the angles are very small, the above equation can be written as  $\theta_t / \theta_i = C_2 / C_1$ . When  $\theta_i > \theta_c$ , the critical angle, the refracted wave travels through the boundary ( $\theta_t = 90^\circ$ ), then  $1 / \sin \theta_i = C_2 / C_1$ . Ultrasound machines assume straight

line propagation, hence refraction effects gives rise to artifacts, not useful for imaging.

### Scattering

Scattering occurs when the size the reflector is  $< \lambda$  ( $10 \mu\text{m}$ ), e.g. blood corpuscle, tissue parenchyma. Scattered US appear in the form of cone, and only small fraction appears as echo. Echoes are small (1–10%), and it is frequency related, hence blood flow imaging require high frequency. Scatter depends on  $\lambda$  and magnitude of roughness. It is useful in imaging of curved surfaces and boundaries, which are not right angle to the incident beam.

Echoes reveal tissue characteristic, and organs signature. Interference of scattered echoes from different site is called speckle. Speckle pattern is not related to anatomical detail, but change of pattern reveal pathology, e.g. liver. Hyperechoic refers higher scatter amplitude and hypoechoic refers lower scatter amplitude. Hyperechoic is due to large, more number of scatters with large  $z$  difference.

In abdomen imaging, strong echoes arise from gas bubbles. Kidney, pancreas, spleen, and liver constitutes complex tissues containing scattering sites, gives rise to speckled texture. Bladder, blood vessels and cysts (fluids) have no internal structure, hence no echoes and appear as black. Vascular and perfusion imaging use contrast agents, which are encapsulated with microbubbles ( $3\text{--}6 \mu\text{m}$ ) containing air, nitrogen or insoluble gases (per fluorocarbons). They permits perfusion of tissues and echoes are generated by the large difference in  $z$ , between the gas and surrounding fluids or tissues.

### Attenuation

Attenuation is the loss of acoustic energy with distance, and it is exponential in nature. It refers both absorption and scattering of ultrasound. In absorption, energy is converted to heat due to frictional and viscous forces. Multiple interfaces offer scattering and partial reflection. Attenuation depends on viscosity, relaxation time, and US frequency. Higher the frequency, greater the attenuation (Fig.14.3). Attenuation coefficient is ( $\mu$ ) is given by;

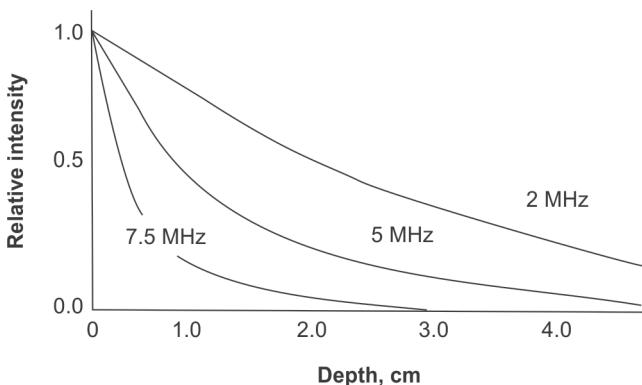
$$I_x = I_o \exp(-\mu x), \text{ dB cm}^{-1}$$

where,  $I_o$ , and  $I_x$  are the initial and final ultrasound intensity in tissue, and  $x$  is the depth in cm. The thickness of the tissue that reduces the sound intensity to half of its original value is called HVL. In soft tissue, attenuation linearly varies with frequency and HVL is 3 cm for

1 MHz frequency. The attenuation coefficient for any frequency is obtained by the relation;

$$\text{Attenuation} = f \times 0.5 \text{ dB/cm}$$

where,  $f$  is the frequency. Attenuation is little in water and greater in bone (Table 14.2). US is attenuated more rapidly than audible sound.



**FIG. 14.3:** Attenuation of US in tissue depth, higher the frequency greater the attenuation

**TABLE 14.2** Attenuation coefficient for various body tissues

Tissue composition	Attenuation coefficient, dB/cm @ 1MHz
Water	0.0002
Blood	0.18
Soft tissue	1.0
Brain	0.85
Liver	0.9
Fat	0.6
Bone	20

Depth of an echo producing structure is determined from the time between the pulse emission and the echo return. The amplitude of the echo is encoded as a gray scale, to form a 2D image. It is also used to carry out anatomic distance, volume measurements, motion studies, blood velocity measurements and 3D-imaging. Its contrast in soft tissue is equal to that of X-rays. High resolution, real-time imaging, harmonic imaging, 3D data acquisition and power Doppler are its

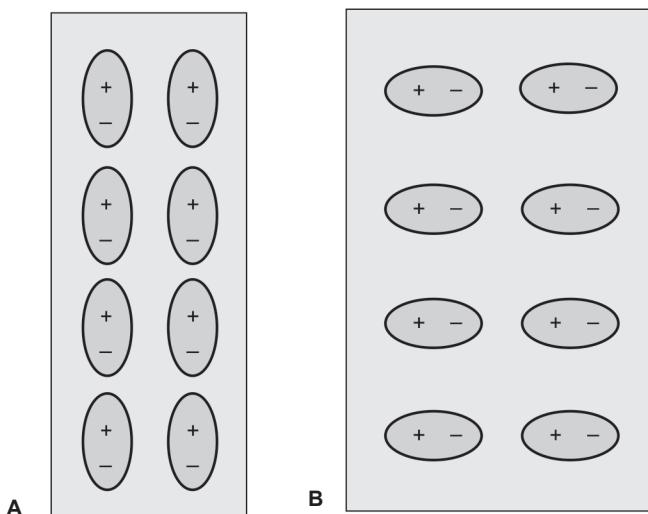
additional strength. Contrast agents can be used for better delineation of the anatomy. Measurement of tissue perfusion, precise drug delivery mechanisms, and determination of elastic properties of the tissues are also possible.

## ULTRASOUND TRANSDUCER

Transducer is a device which converts energy from one form to another. US transducer converts electrical energy into sound energy and vice versa. It works on the principle of piezoelectric effect. When a crystal (quartz) is subjected to mechanical pressure, electrical voltage is created and *vice versa*.

## PIEZOELECTRIC MATERIAL

Piezoelectric means pressure electricity. Piezoelectric materials have molecular dipoles, containing + and – electric charges, and net the charge is zero (Fig. 14.4). If electrical voltage is applied across the crystal, the dipole orientation changes, resulting in variation of crystal thickness. The crystal undergoes compression and expansion. Similarly, if mechanical pressure is applied to the crystal, molecular dipoles change their orientation, altering the electric field, produce voltage signal. The created potential difference is proportional to the pressure applied to the crystal.

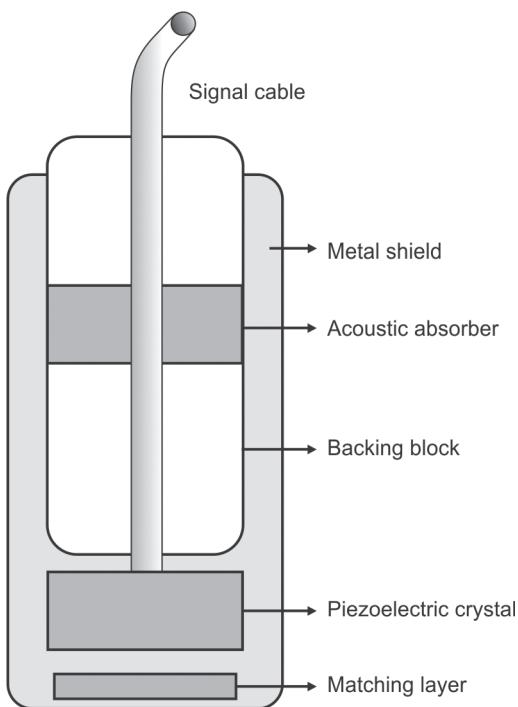


**FIG. 14.4:** Piezoelectric effect: (A) Compression and (B) Expansion

Natural piezoelectric material is quartz crystal and synthetic ceramic material is most commonly used as crystal, e.g. lead zirconate titanate (PZT,  $\text{PbZrTiO}_4$ ). PZT in its natural state has no piezoelectric properties. It is heated over its curie temperature ( $328\text{--}365^\circ\text{C}$ ) and external voltage is applied, that causes the dipoles to align in the ceramic. The voltage is maintained, until the ceramic is cooled below its curie temperature. Once it is cooled, the dipoles retain their alignment.

## TRANSDUCER DESIGN

Ultrasonic transducer consists of (i) matching layer, (ii) piezoelectric crystal, (iii) backing block, (iv) acoustic absorber, (v) metal shield, and (vi) signal cable. There are two type of transducers, namely, resonance and non-resonance transducers (Fig. 14.5).



**FIG. 14.5:** Ultrasound transducer design

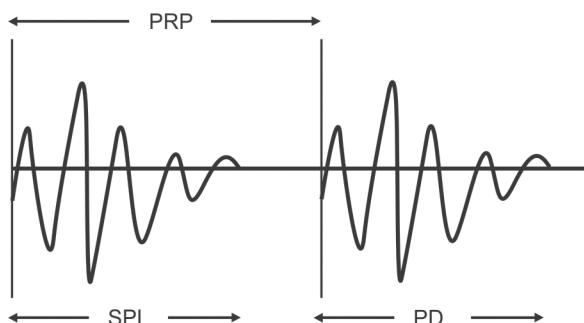
### Resonant Transducer

Resonant type transducers follow the relation  $\lambda = 2T$ , where  $T$  is the thickness of the crystal. A potential of 150 V is applied to the crystal in  $1\ \mu\text{s}$ , the ceramic initially contract, later vibrate at a natural frequency.

It is used in pulse echo ultrasound imaging. High frequency transducer require thinner crystals and low frequency require thicker crystals.

### Pulse Characteristics

Ultrasound pulse rises and falls rapidly (Fig. 14.6). Spatial pulse length (SPL) is the product of wavelength and number of cycles in the pulse ( $\lambda \times \eta$ ). The range of SPL of diagnostic ultrasound is 0.3–1.0 mm and it depends on the frequency. The pulse duration (PD) is the time taken for one spatial pulse length and its range is 0.4–1.5 ms. The SPL get altered during transmission through tissue. Higher frequencies are more attenuated than low frequencies, resulting in increased SPL. This will degrade image resolution. The pulse repetition frequency (PRF) is the number of pulses per second. The pulse repetition period (PRP) is the reciprocal of PRF and its range is 0.1–0.5 ms.



**FIG. 14.6:** Ultrasound pulse characteristics: increase of frequency decreases the pulse length

### Damping Block

Damping block is on the back side of the crystal, made up of tungsten particles suspended in epoxy resin ( $Z = 3 \times 10^7 \text{ kg m}^{-2} \text{ s}^{-1}$ ). It absorbs the backward US pulse and also attenuates stray US signals. Transducer and the damping block are separated from the casing with an insulator (rubber cork). Damping lessens the purity of the resonance frequency, and introduces a broadband frequency spectrum. The Q factor describes the bandwidth and  $Q = f_o/\text{Bandwidth}$ , where  $f_o$  is the center frequency. High Q transducer has narrow bandwidth, little damping, long spatial pulse length (SPL) and used for velocity measurements, and Doppler study. Low Q, has wide bandwidth, short SPL, provides high spatial resolution in axial direction.

### Matching Layer

The matching layer minimizes the acoustic impedance differences between transducer and the patient. Its acoustic impedance is intermediate to those of soft tissue and transducer. The thickness is equal to one-fourth the wavelength, which is known as *quarter wave matching*. The matching layer impedance is obtained as follows;

$$Z_M = \sqrt{(Z_T \times Z_L)}$$

where,  $Z_T$  impedance of the crystal and  $Z_L$  is the impedance of the tissue. The matching layer is made up of perspex or plexiglass ( $Z_M = 3.2 \times 10^6 \text{ kg/m}^2/\text{s}$ ), loaded with aluminum powder. In addition, acoustic coupling gel is used, between transducer and skin.

#### *Worked Example 14.2*

Calculate the thickness of the matching of a probe with following specifications;  $f = 7.5 \text{ MHz}$ ,  $C = 1540 \text{ msec}^{-1}$ , and  $\lambda = 0.2 \text{ mm}$ .

*Answer:* As per the quarter wave matching, the thickness is equal to  $\frac{1}{4}$  of wavelength

$$\text{Thickness} = (1/4) \times 0.2 = 0.05 \text{ mm}$$

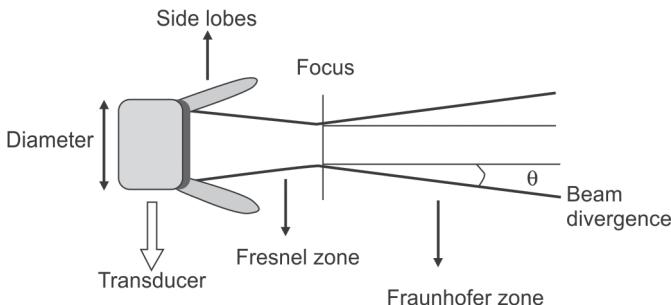
### Nonresonant Transducers

Nonresonant transducers produce multiple frequency, in which the center frequency can be adjusted in the transmit mode. Piezoelectric element is machined into a large number of small rods and then filled with an epoxy resin, to create a smooth surface. The acoustic properties are closer to tissue than a pure PZT material. This design facilitates reduction of matching layer, with increased transmission efficiency. The bandwidth of this transducer is about 80% of the centre frequency. A short square wave burst of 150 V is used to excite the transducer. They receive echoes of wide range of frequencies and useful in harmonic imaging, where low frequency is transmitted, and high frequency echo is received.

### Fresnel Zone

Fresnel zone or near field, is adjacent to the transducer face, has a converging beam profile, due to multiple constructive and destructive interference (Fig. 14.7). The near field length =  $d^2/4\lambda = (d^2 \times f)/4 \times C$  where  $d$  is the diameter of the transducer. Higher frequency and larger diameter always provide longer near field length. The lateral resolution depends on beam diameter, and it is best at the end of near field

(single element). The pressure amplitude characteristics are complex (minimum-maximum) in the near field. Peak US pressure occurs at the end of near field, where beam diameter is minimum.



**FIG. 14.7:** US beam: Fresnel zone, Fraunhofer zone, and side lobes

### Fraunhofer Zone

Fraunhofer zone or far field is one, in which the beam diverges and obey  $\sin\theta = 1.22 \lambda/d$ , where  $\theta$  is the beam divergence,  $d$  is transducer effective diameter. Less beam divergence occurs with high frequency and with large diameter. The US intensity decreases with distance.

### Side Lobes

Side lobes are unwanted emissions of US energy, directed away from the main pulse. It is caused by radial expansion and contraction of the element. In receive mode, side lobes can create artifacts in the image. Smaller individual element width ( $< \lambda$ ), and array transducer reduce side lobes. Apodization is a method in which the transmit pulse is changed from square to Gaussian function. This eliminates the side lobes and improves image quality.

In continuous mode, side lobe is significant in narrow bandwidth transducers (high Q). In pulsed mode, side lobe reduces in broad bandwidth transducers (low Q). In multi-element array, the side lobe is in forward direction. Grating lobes results, when US energy is emitted far off-axis by multi-element arrays. It is due to non-continuous transducer surface of the discrete elements. It creates image of highly reflective off axis objects in the main beam.

## TRANSDUCER ARRAY

Transducer array can be made with large number of elements. It is classified as (i) linear sequence arrays, (ii) phased array and (iii) annular transducer.

### Linear Array

In linear array, large number of elements is arranged in groups. Only one group (8–16) is excited during scan. A single element is added to the group while the last element is removed. Thus, aperture advances along the transducer length and gives images line by line defined by individual element (Fig. 14.8).

It provides wider beam with improved resolution and operated both in unfocused and focused mode. Focused mode is made by delaying or phasing the excitation pulses (ns), in order to obtain short or long focal length. If the linear array works in the above manner, it is said to be phased linear array. The field of view of the linear array is rectangular or parallelogram. It facilitates dynamic aperture to have focus at various depths. The frame rate = PRF/N (number of scan lines).

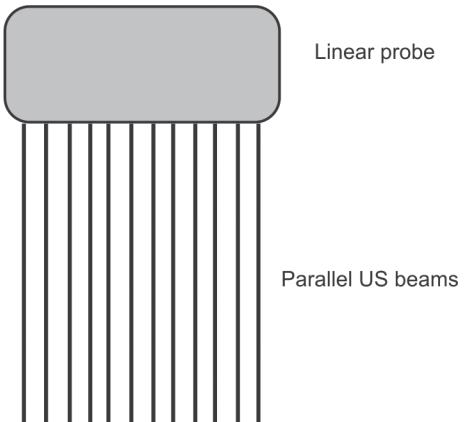
Linear array can be designed with convex transducer surface, used for sector scanning, where it is called as curved array. Curved array transducer are smaller in size, provides sequential element switching, and smaller foot print (surface contact area). The beam is wider at depth, covers larger anatomy.

The application of linear array includes abdomen (large body area), gynecology, thyroid; superficial vessel imaging and US guided biopsy. The advantages are good definition both in near and distant anatomy and good image quality across image depth and FOV. The disadvantage is large foot print and limited field of view. The linear array specifications are:

- Number of elements: 60–120
- Elements in a single group: 8–16
- 3.5–7.5 MHz frequency
- Element width 1–41
- Foot print size:  $2 \times 0.6$  cm to  $1.4 \times 10$  cm.

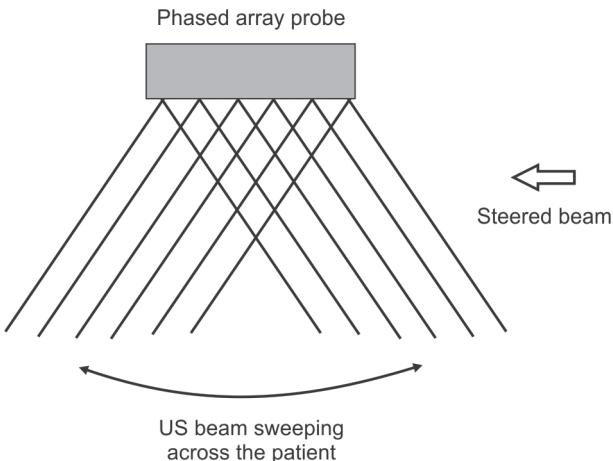
### Phased Array

In phased array, each element is excited via a delay which gives a swiveled (angled,  $\pm 45^\circ$ ) wave front (Fig. 14.9). This requires narrow



**FIG.14.8:** Linear array with multiple elements: Rectangular or parallelogram image display

elements of about  $\lambda/2$  dimension. Same delay is used in receive mode, and the transducer is sensitive to particular echo. Smaller number of elements (48–128) with smaller foot prints are used. Each element has separate transmit and receive circuits. It produces both directional and focused beams.



**FIG.14.9:** Phased array with multiple elements:  
Sector image display

A pie shaped/sector image is formed by polar coordinates. The pulse rate determines number of scan lines. In dynamic focusing, change of delay time during receive mode, vary the focal plane, but increase of focal plane decreases frame rate. It provides dynamic apertures, with varying element number. The advantages of phased array includes large FOV, and fast frame rate. The disadvantage is poor near field of view.

Its application includes small arrays, used in echocardiography, and large array used in abdomen, that gives larger FOV. Phased array probe is also used for internal examinations such as rectum, vagina, and esophagus. Tiny catheter sized probe is available for intraluminal study of blood vessel that allows imaging both in transverse and longitudinal planes. The specifications of phased array are:

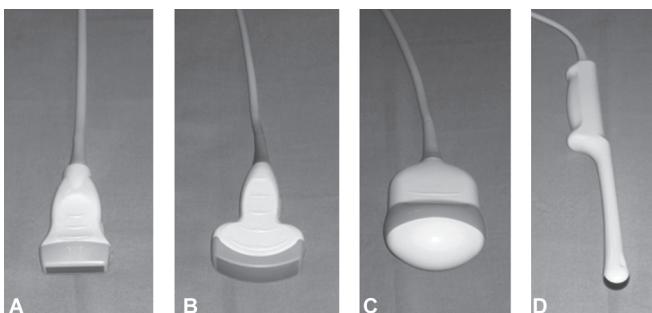
- Frequency: 2.5, 3.5, 5.0, and 7 MHz
- Frame rate up to 156 fps
- Imaging depth 25 cm
- Aperture 14–28 mm
- Swivel: 40–45°.

### Annular Transducer

The annular transducer contains series of concentric transducers that provides conical beam. The signals are delayed to each ring, to create a focused beam. It provides excellent image quality with superior slice thickness uniformity. It is used in dynamic focusing, but not suitable for Doppler study.

### Clinical Probes

The transducers are also made to suit the individual clinical applications. The clinical probes include (i) endo-vaginal-pelvic region, (ii) endorectal-prostate, (iii) transesophageal heart, and (iv) intravascular blood vessels, etc. (Fig. 14.10).



**FIG.14.10:** Clinical probes: (A) Rectilinear, (B) Curvilinear, (C) Real time 4D Doppler, and (D) Transvaginal

## IMAGE DATA ACQUISITION

US uses pulse echo method of imaging. Each pulse transmits directionally into the patient, and then experiences partial reflections from tissue interfaces that create echoes, which returns to the transducer.

### PULSE ECHO OPERATION

US is intermittently transmitted and major time is spent for listening the echoes. The pulse is created with short voltage waveform, with 2–3 cycles long. The time delay between the transmission pulse and echo is related to depth of the interface.

- Time ( $\mu\text{s}$ )  $= 2D/c$   
 $= 2D(\text{cm})/0.154 \text{ (cm}/\mu\text{s)}$   
 $= 13 \mu\text{s} \times D$

- Distance (cm) =  $(c \times \text{Time})/2$
- =  $(0.154 \text{ cm}/\mu\text{s} \times \text{Time})/2$
- =  $0.077 \times \text{Time} (\mu\text{s})$

The pulse repetition, period (PRP) is the inverse of the pulse repetition frequency (PRF) and the common PRF used is 2 or 4 kHz. An increase in PRF results in a decrease in echo listening time. Maximum PRF is determined by the time required for echoes from the most distant objects to reach the transducer. High PRF limits the penetration and low PRF limits line density and frame rate (ability to follow motion).

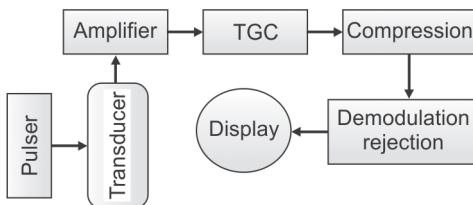
The duty cycle is the fraction of ON time = pulse duration/PRP. In real-time imaging, it is 0.2–0.4%, hence > 99.5% of the scan time is spent in listening the echoes. The PRF, PRP and duty cycle of various US modes are given in Table 14.3.

**TABLE 14.3** PRF, PRP, and duty cycle of different ultrasound modes

Mode	PRF (kHz)	PRP ( $\mu\text{s}$ )	Duty cycle (%)
M-Mode	0.5	2000	0.05
Real-time	2–4	500–250	0.2–0.4
Pulsed Doppler	4–12	250–83	0.4–1.2

## ULTRASOUND EQUIPMENT

The ultrasound hardware components include; (i) pulser, (ii) amplifier (iii) TGC, (iv) compression (v) demodulation and rejection, and (vii) display (Fig. 14.11).



**FIG. 14.11:** Ultrasound block diagram

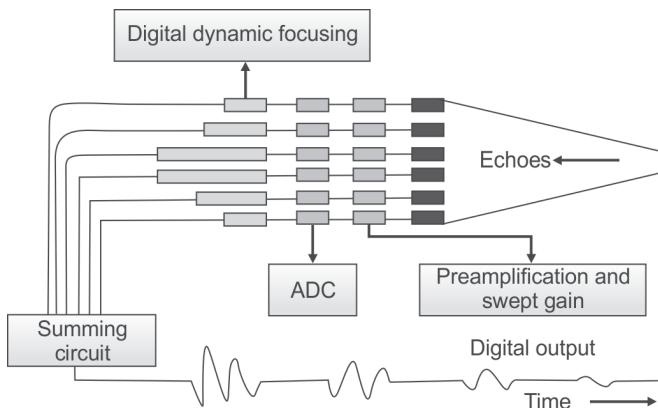
### Pulser

The pulser controls the output transmit power by adjustment of the applied voltage. It provides electrical voltage for exciting the transducer elements. An increase in transmit amplitude creates higher intensity, improves echo detection from weaker reflectors. It provides higher signal

to noise ratio, but the power deposition to the patient is higher. Pulser also has user control labels such as; output, power, dB, and transmit. It helps low power setting for obstetric imaging and also has power indicators; thermal index (TI) and mechanical index (MI).

### **Amplifier**

Each PZT has its own preamplifier and analog to digital converter (ADC). Each element produces a small voltage of the echoes, and has a preamplifier, combined with swept gain (Fig. 14.12). Initial pre-amplification increases the detected voltages to useful signal levels (100 dB) and swept gain compensate for the exponential attenuation of signal. The ADC has larger bit depth, to digitize the signals directly from the preamplification stage.



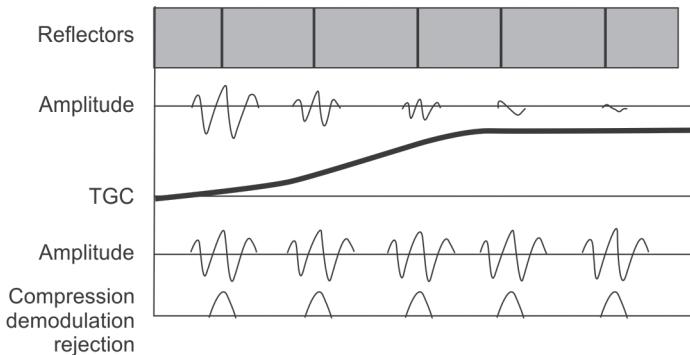
**FIG. 14.12:** Phased array echoprocessing

### **TGC**

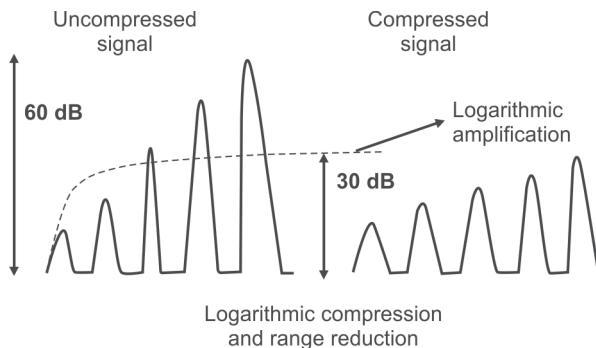
The time gain compensation (TGC) amplify the signal proportional to the time delay between transmission and detection of US pulses. The amplification may be linear or non linear and bring the signal in the range of 40–50 dB. This process compensates for tissue attenuation and makes all equally reflective boundaries equal in signal amplitude, irrespective of depth (Fig. 14.13).

### **Compression**

The logarithmic compression increases the smallest echo amplitudes and decreases the largest amplitudes (Fig. 14.14). It reduces signal range, to fit the dynamic range of video monitor and film (20–30 dB). The output signal is proportional to the logarithm of the input signal. Rectification inverts the negative amplitude echoes to positive.



**FIG.14.13:** Schematic description of TGC, equal reflective boundaries will have equal amplitude



**FIG.14.14:** Logarithmic compression

### Demodulation and Rejection

Demodulation converts the rectified amplitudes into a smoothed, single pulse. Rejection circuit removes a significant amount of undesirable low level noise and scattered sound (electronics).

### Display

Scan converter creates 2D images from echo information, from distinct beam directions, so that image can be displayed on video monitors. Analog scan converters drift easily and unstable over time, whereas digital scan converters are stable and allow image processing with variety of mathematical functions. Scan converter memory is configured as rectangular matrix elements ( $512 \times 512$ ). Each pixel has a memory address that defines its location within the matrix.

The digital signals are inserted into the matrix at memory addresses, corresponding to the relative reflector in the body. Transducer beam

orientation and echo display times determine the correct pixel address, in which the digital information is to be deposited. The final image is displayed in a  $512 \times 512$  matrix with a depth of 8 bits per pixel. In color display, the depth of 24 bits (3 bytes) is employed.

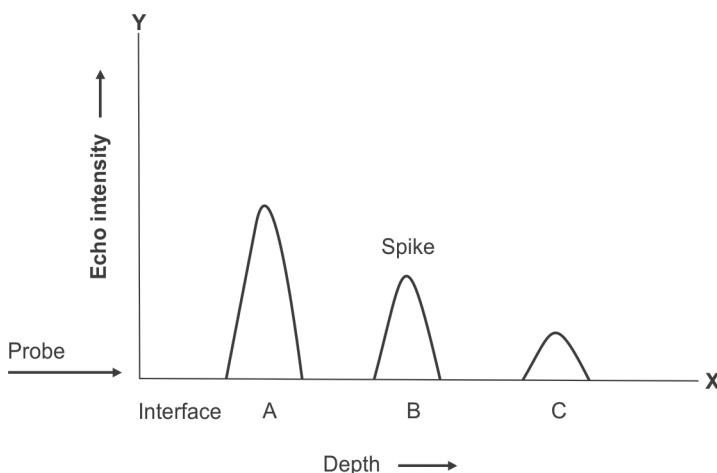
## ULTRASOUND IMAGE DISPLAY

The US image is an electronic representation of data generated from returning echoes and displayed on a TV monitor. The image is assembled, one bit at a time, like television image. Echo generates one bit of data, and many bits together forms the electronic image. This image may be displayed as (i) Amplitude (A)-Mode, (ii) Motion (M or TM)-Mode, and (iii) Brightness (B)-Mode.

### AMPLITUDE (A)-MODE

The probe is held stationary, and pulses of nanosecond duration is sent into the patient, and echo is generated. Each interface gives one echo pulse. Echoes are displayed as spikes projecting from baseline, which identifies the central axis of the beam. It displays the depth on X-axis and echo intensity on the Y-axis (Fig. 14.15). It is a simple US technique, shows only the position of interfaces.

The application of A-mode includes ophthalmology-distance measurements, echoencephalography, echocardiography, examinations of the eye, detecting a cysts in the breast, studying midline displacement in the brain, etc.



**FIG. 14.15:** A-mode display

## MOTION (M OR TM)-MODE

The A-mode spikes are converted into dots and brightness represents amplitude. When the interface moves, the dots also move back and forth. Sequential US pulses are displayed adjacent to each other, allowing the change in position of interfaces. It is recorded over a period of time. It provides excellent temporal resolution of motion patterns and displays the time on the X-axis and depth on the Y-axis (Fig. 14.16). Its application includes evaluation of cardiac valve motion and other cardiac anatomy. Real-time 2D echocardiography, Doppler and color flow imaging reduces the importance of M-mode today.

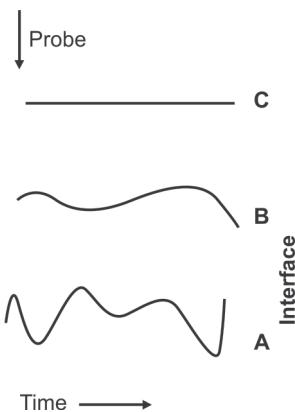
## BRIGHTNESS (B)-MODE

In B-mode, a slice of an anatomy of the patient is imaged. The transducer is moved back and forth, so that beam scans a 2D section of the patient. It may be a linear or sector scanning. Echoes are displayed as dots, and brightness is proportional to echo intensity (Fig. 14.17). Thus, thousands of echo signal strengths of varying brightness of points gives grey scale image. The image displays a section of an anatomy. The image depth depends on transducer frequency, focus, etc. The B-mode scanning is usually done with electronic scanning either with linear array or phased array.

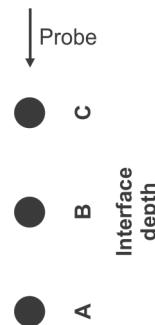
### Electronic Scanning

#### *Linear array*

This is performed with elongated array transducer with multiple piezoelectric elements. Individually, they give poor beam, with short near field, and widely diverging far field. Hence, they are operated in groups,



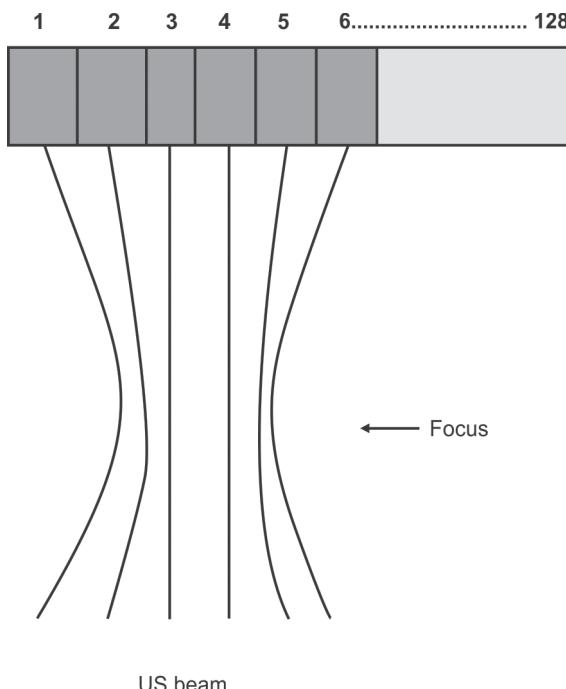
**FIG. 14.16:** Ultrasound M-mode display: Change of position of interface over time, amplitude is replaced by brightness



**FIG. 14.17:** B-mode display: The amplitude is replaced by brightness, interfaces at different depth gives varying brightness proportional to their echo intensity

say 1–6, 2–7, that gives well defined US beam (wave front). The elements are rectangular in shape and each one is individually pulsed (Fig. 14.18).

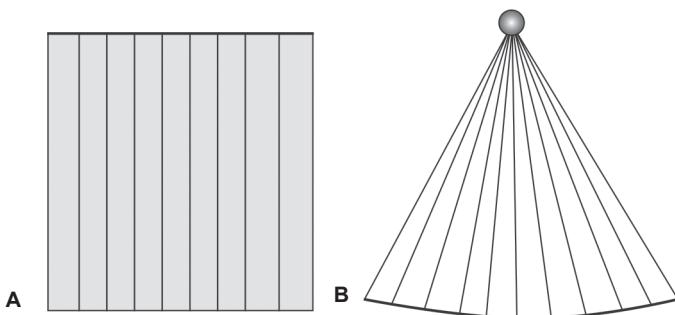
**Electronic focusing:** The elements group are energized from outer to inner, pulses reinforce at same point (focus), at a same time. The timing of the applied pulse can be varied to get a desired focal depth. Thus, it scans a rectangular area with large number of scan lines. Each element produces a scan line, whose length gives image depth. It electronically sweep the US beam across the VOI, enable dynamic imaging.



**FIG. 14.18:** Electronic scanning with linear array: In a group, outer most pair is energized, followed by inner most pair. US beam reinforces at the focus

### Phased Array

Phased array type uses shorter transducer with fewer elements. If energized simultaneously, acts as a single transducer, and the beam travel forward. If energized separately, in rapid sequence, the pulses reinforce only in one direction, called steering. For example, energizing 1,2,3, the reinforce is in right, whereas energizing 3,2,1, change the beam to left. Thus, scan line sweep across the patient, giving a sector field (Fig. 14.19).



**FIG. 14.19:** (A) Linear array rectangular display,  
(B) Phased array sector display

#### *Focal Depth*

Focal length can be altered electronically by the operator. Greater the time delay in energizing successive pair of elements, shorter the focal length. Beam is focused in the azimuthal plane (plane parallel to the length of the transducer). Focusing in the elevation plane (thickness) is done with an acoustic lens. Electronic focusing is also possible with 1.5 D transducer. Focusing make the image good in the focal region, and worsen the image beyond.

This can be overcome by multiple zone focusing facility. Multiple pulses (3–4) are sent in each scan line and phase delays are altered to have different focus, both at transmission and reception mode. Each time, the transducer is gated to receive only from one focal zone. Echoes arriving from nearer or distant points are blocked.

#### **Real-time Scanning**

Real-time scanning is one in which the image is constructed instantaneously and renewed 30 times per second. It updates the position of the structures that is displayed as it happens, including anatomical motion. Both linear array and sector probes can be used for real-time scanning (Table 14.4). The real-time scanning image quality depends on (i) field of view (FOV), (ii) number of A lines per image (N): high resolution require large number of lines (100) per frame, (iii) line density (LN), (iv) depth (D), and (v) frame rate: to follow motion large number of frames/second is required.

Frame repetition frequency depends on N and PRF, whereas PRF in turn depends frame rate and lines per frame (PRF = frame rate  $\times$  lines per frame). Hence, higher PRF must be used to have higher

frame rate. However, this will decrease the depth selection (Depth  $\times$  PRF = 0.5  $\times$  sound velocity). Both high frame rate and large depth is not possible. In a real-time scanning, the following relationship is true:

$$\text{Depth} \times \text{frame rate} \times \text{lines per frame} = \text{constant}$$

**TABLE 14.4 Sector vs linear scan**

Sector scanning	Linear scanning
Narrow field near skin, wider at depth	Wide field near skin
Easy to manipulate	Require larger patient contact area
Require smaller acoustic window	Better image quality
Heart imaging through intercostal spaces	Imaging of whole abdomen, liver, superficial vessel, thyroid
Infant brain imaging through fontanel	Obstetrics and gynecology
Used in Intracavitory probes	Can be made as curved format, to give sector image

## DOPPLER ULTRASOUND

Doppler ultrasound is based on the shift of frequency in an ultrasound wave, caused by a moving reflector, e.g. blood cells. It is similar to a siren on a fire truck; the sound is high pitched as the truck approaches the listener ( $\lambda$  is compressed), and shifted to a lower pitch as it moves away from the listener ( $\lambda$  is expanded).

Comparison of incident US frequency with the reflected US frequency from the blood cells, gives the velocity of blood. It also helps measurement of blood flow (indirect), create color blood flow maps of vasculature, etc. The change of frequency is proportional to the velocity of the interface. Higher the transducer frequency or faster the interface moves, greater the change in frequency.

## DOPPLER SHIFT

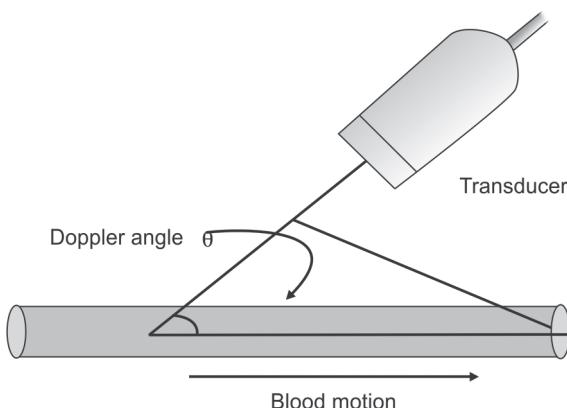
Doppler shift ( $f_d$ ) is the difference between the incident frequency and reflected frequency. If a reflector is moving away from the source,

$$f_d = f_i - f_r = \frac{\text{Reflector speed} \times 2 \times f_i}{\text{Reflector speed} + \text{speed of sound}}$$

where,  $f_i$  is the frequency of the incident sound, and  $f_r$  is the frequency of the reflected sound. Thus, blood moving away from the transducer, expand the sound waves, and produces lower frequency. Similarly, blood moving towards the transducer compresses the sound waves and produces higher frequency echoes.

When the sound and blood are not in parallel, the Doppler shift is less, and the equation needs modification (Fig. 14.20). If  $\theta$  is the angle between the US beam and direction of moving blood, then

$$\cos \theta = \text{Adjacent side}/\text{Hypotenuse}$$



**FIG. 14.20:** The ultrasound and blood flow directions are not parallel; it makes an angle  $\theta$  with blood flow direction

In the Doppler shift equation, the reflector speed is much smaller (200 cm/s) compared to sound speed (154,000 cm/s), hence, the equation may be written as

$$f_d = \frac{\text{Reflector speed (v)} \times 2 \times f_i \times \cos \theta}{\text{Speed of sound (c)}}$$

Thus, the Doppler frequency shift,  $f_d = 2 f_i v \cos \theta / c$ , where,  $v$  is the velocity of moving blood,  $c$  is the speed of sound,  $f_i$  is the incident sound frequency, and  $\theta$  is the Doppler angle.

The blood velocity,  $v = f_d c / (2 f_i \cos \theta)$ . If  $\theta = 0, 30, 45, 60$  and  $90$ , then  $\cos \theta = 1, 0.87, 0.7, 0.5$  and  $0$ , respectively. The maximum shift is found at  $0$  angle, that is  $\theta = 0$ ,  $\cos \theta = 1$ . At large angle ( $> 60$ ), the shift is small, minor errors in angle accuracy can result in large errors in velocity. Doppler frequency shift of the moving blood occurs in the audible range, which can be converted into audible signal

through loudspeaker. It can be heard by the operator, to help in positioning and diagnosis.

#### *Worked Example 14.3*

In a Doppler examination  $f_i = 5 \text{ MHz}$ ,  $v = 25 \text{ cm/s}$ ,  $\theta = 45 \text{ deg.}$ , calculate the Doppler shift?

Doppler shift,  $f_d$

$$= 2 f_i v \cos \theta/c,$$

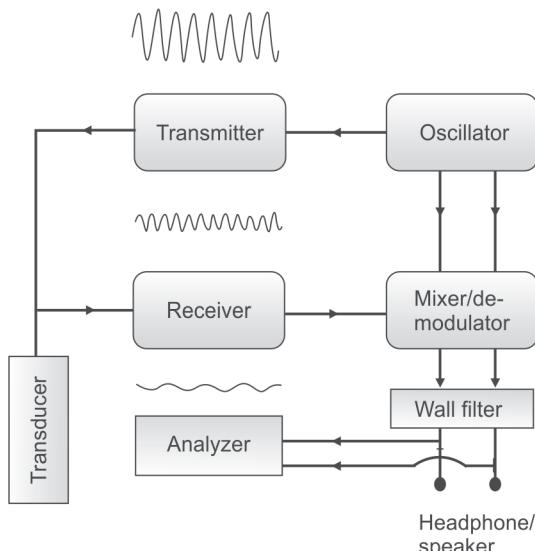
$$= \frac{2 \times 5 \times 10^6 \times 25 \text{ cm/s} \times 0.707}{1540,00 \text{ cm/s}} = 1.14 \times 10^3 = 1.14 \text{ kHz}$$

## DOPPLER SYSTEMS

The Doppler systems used in medicine may be divided as (i) continuous Doppler, (ii) pulsed Doppler, (iii) duplex scanning (iv) color flow imaging and (v) power Doppler.

### Continuous Doppler

It has two transducers, one for transmission and other for detecting echoes. An oscillator gives resonant frequency to the transmitter and demodulator, which compares the returning frequency to the incident frequency (Fig. 14.21). Receiver amplifies the returning signal. Mixer/demodulator extract the Doppler shift frequency by using low pass filter. The low pass filter removes superimposed high frequency oscillations. Wall filter is used to remove low frequency signals from vessel wall



**FIG. 14.21:** Continuous wave Doppler system

and moving specular reflectors. An audioamplifier amplifies the Doppler signal. A recorder tract the spectrum changes as a function of time for analysis.

### *Advantages*

High accuracy of Doppler shift measurement is possible, since narrow bandwidth is used. High velocities are measured without aliasing effects. Demodulator does not reveal, whether the flow is toward or away from transducer. Hence, it uses a method of signal processing called quadrature detection, to determine direction of flow.

### *Disadvantages*

It suffers from depth selectivity and accuracy is affected by object motion within the beam path. Multiple overlying vessels result in superimposition, difficult to distinguish. Spectral broadening of frequencies occurs with a large sample area across the vessel profile, resulting in high velocity at center and lower at the edges of the vessel.

### **Pulsed Doppler**

Pulsed Doppler combines continuous wave Doppler and pulse echo imaging. The former gives velocity information and the later gives depth information. In pulse echo format, long SPL is used, to get accuracy of frequency shift. Depth selection is obtained by an electronic gate, which will reject all signals except those falling within the gate window. Alternatively, multiple gates are also used to get velocity profile across the vessel.

The phase of the returning echoes from a stationary object, relative to the phase of the transmitted sound, does not change with time. However, the phase of returning echoes from moving object does vary with time. Hence, sample of the shifted frequencies are measured as phase change. As per sampling statistics theory, PRF must be at least twice the Doppler frequency shift, otherwise aliasing may occur. This may leads to error in velocity estimation.

$$\text{That is, } \text{PRF} = 2 f_d \text{ (max)} = 2 \times 2 f_o v_{(\text{max})} \cos \theta / c \\ \text{or } v_{\text{max}} = c \times \text{PRF} / (4 f_o \cos \theta)$$

Thus, the maximum velocity,  $V_{\text{max}}$  depends on PRF,  $f_o$  and  $\cos \theta$  (larger angles). If 2 kHz is the maximum Doppler shift frequency, then the minimum PRF should be equal to  $2 \times 2 \text{ kHz} = 4 \text{ kHz}$ .

### Duplex Scanning

Duplex scanning consists of both B-mode real-time US and pulsed Doppler. Since pulsed Doppler does not provide visual display, which is overcome by duplex scanner. Initially, it operates in 2D, B-mode to create an real-time image, to select a Doppler gate window position and then switched over to pulsed Doppler mode. Electronic array transducers are used in a group for B mode, then one or more is used for Doppler information. Velocity and Doppler angles are obtained from Doppler frequency shift and B-mode scan, respectively. The information of flow is obtained by the relation:

$$\text{Flow (cm}^3/\text{sec)} = \text{vessel's cross-sectional area (cm}^2\text{)} \times \text{velocity (cm/sec)}$$

There may be error in flow volume due to: (i) vessels axis might not lie totally within the scanned plane, (ii) curved vessel, and (iii) altered flow. This gets exaggerated, if Doppler angle  $> 60^\circ$ , misposition of Doppler gate overestimate velocity and non-circular cross-section of vessel cause errors in area estimation. However, Duplex scan has the following features: (i) velocity profile can be obtained with multiple gate, (ii) mapping of velocity in color and flow in gray scale, and (iii) direction of flow by real-time color flow, etc.

### Color Flow Imaging

Color flow imaging gives 2D visual display of moving blood over a gray scale image (Fig. 14.22). Blood moving towards the transducer is coded with red, and blood moving away from the transducer is coded with blue.

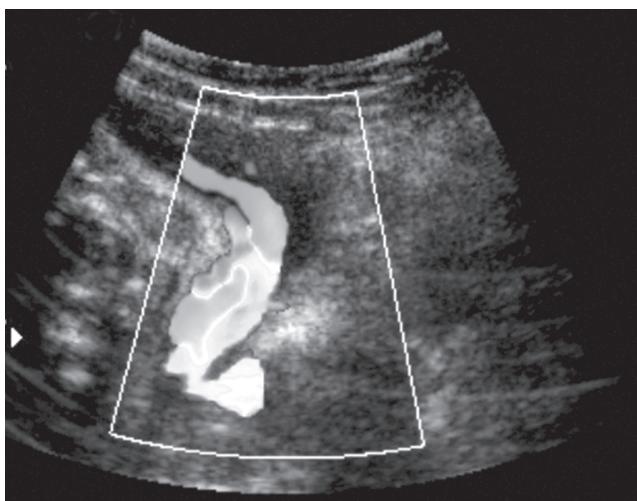


FIG. 14.22: Color Doppler of umbilical cord

Thus, artery and veins are coded with red and blue color, respectively. There are two methods of color flow imaging, namely, (i) phase shift autocorrelation and (ii) time domain correlation.

In phase shift autocorrelation, the similarity of one scan line to another, when maximum overlap exists, is measured. In time domain correlation, the reflector motion over a time period,  $\Delta T$ , between consecutive pulse echo amplitude is measured. The degree of similarity between the two is mathematically correlated. Velocity measurement accuracy is limited in time domain correlation.

### Doppler Spectrum

A Fourier transform mathematically analyze the US signal and gives amplitude vs frequency profile. This is known as Doppler spectrum which consists of variety of frequencies (due to velocity) within a sampling gate, over a time period (Fig. 14.23). It is continuously updated as real-time.

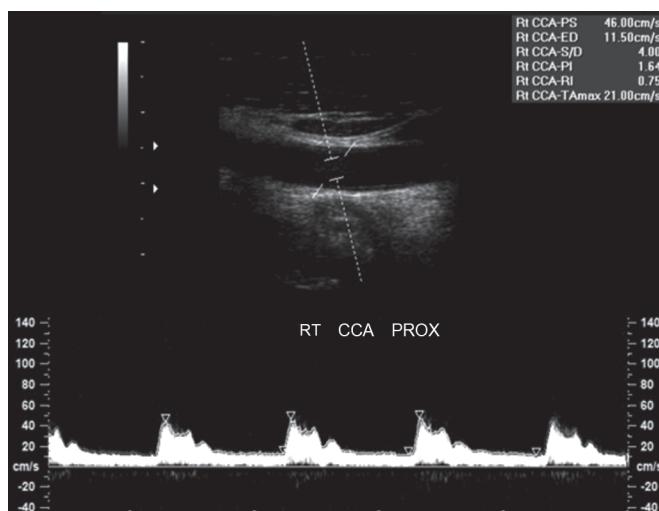


FIG. 14.23: Doppler spectral display

The blood flow may be laminar, blunt or turbulent depending on the vessel wall characteristics, size, shape, and flow rate. Fast laminar flow, present at the center of a large smooth vessel walls, and slower flow present near the walls due to frictional forces. The turbulent flow is caused by plaque and stenosis. Hence, a large Doppler gate positioned to encompass the entire lumen will contain large range of blood velocities. A smaller gate positioned at the center of the vessel will have smaller, faster range of velocities. A Doppler gate positioned near a stenosis in the turbulent flow pattern will measure the largest range of velocities.

### *Display*

The Doppler spectrum is displayed below the 2D B-mode image as a moving trace, on a monitor. The flow velocity (a frequency) is in Y axis and the time in X axis. Intensity of the Doppler signal at a particular frequency and moment in time is displayed as brightness at that point. Velocities in one direction are displayed as positive, and other direction as negative in a vertical scale. As new data arrive, the information is updated and scrolled from left to right. It appears as choppy sinusoidal wave.

### *Spectral Interpretation*

Spectrum reveals presence of flow, direction of flow and its characteristics. Doppler angle of  $30^\circ$  gives best result for direction of flow. It is very difficult to identify lack of flow, that is due to noise. Normal blood flow is represented by a specific characteristic spectrum. Disturbed and turbulent flow alters the spectrum proportional to disease pattern. Pulsatility and resistivity are quantified as follows:

$$\text{Pulsatility index (PI)} = \frac{\text{Maximum flow} - \text{Minimum flow}}{\text{Average flow}}$$

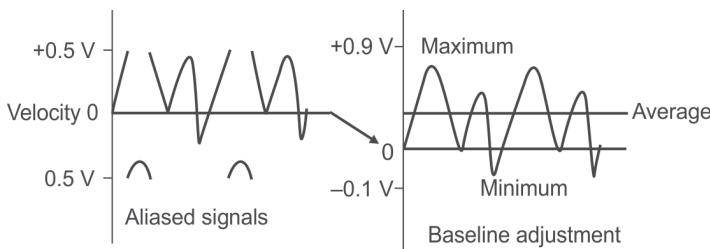
$$\text{Resistive index (RI)} = \frac{\text{Maximum flow} - \text{Minimum flow}}{\text{Maximum flow}}$$

### *Aliasing*

Aliasing is an error caused by an insufficient sampling rate (PRF), relative to high frequency of Doppler signals, generated by fast moving blood. Minimum two samples per cycle of Doppler shift frequency are required to determine velocity unambiguously. The aliased signals wrap around to negative amplitude, masquerading as reverse flow. To avoid aliasing, (i) the velocity scale may be adjusted to wider range or (ii) the spectral base line (0 velocity) may be lowered (Fig.14.24). However, the minimum and maximum Doppler shift cannot exceed  $\pm \text{PRF}/2$ .

### **Power Doppler**

Power Doppler is a signal processing method that relies on the total strength of the Doppler signal (amplitude) and ignores directional (phase) information. The power mode of signal acquisition is dependent on the amplitude of all Doppler signals, regardless of frequency shift. This improves sensitivity to motion (e.g. slow blood flow), at the expense of directional and quantitative flow information. Images are more sensitive to motion and are not affected by the Doppler angle.



**FIG. 14.24:** Aliasing of signals and remedial measures

Aliasing is not a problem as only the strength of the frequency shift is analyzed, not the phase. Greater sensitivity allows detection and interpretation of very subtle and slow blood flow. Frame rates are slower and flash artifacts may occur, which are related to color signals arising from moving tissues, patient motion, or transducer motion. Power Doppler uses the same levels of power as standard color flow procedure (no increased transmit power).

## ULTRASOUND IMAGE QUALITY, ARTIFACTS AND BIOEFFECTS

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### IMAGE QUALITY

Ultrasound image quality depends on (i) spatial resolution, (ii) temporal resolution, (iii) contrast resolution, and (iv) noise.

#### Spatial Resolution

Spatial resolution is the ability to resolve two closely placed objects. It is divided into axial resolution, lateral resolution and elevational resolution.

##### *Axial Resolution*

Axial resolution is the ability to separate two closely spaced interfaces, along the direction parallel to the beam. It is equal to  $1/2$  spatial pulse length. Higher the frequency, shorter the pulse, and greater the damping and better the axial resolution.

##### *Lateral Resolution*

Lateral resolution is the ability to separate two objects at same depth side by side, i.e. across the beam. It depends on beam width, shape and varies with distance from transducer. It decreases with increasing beam width, and increases with increase of frequency (Table 14.5).

**TABLE 14.5** The axial and lateral resolution variation with frequency

Frequency (MHz)	Image depth (cm)	Axial resolution (mm)	Lateral resolution (mm)
2.0	30	0.7	3.0
3.5	17	0.4	1.7
5.0	12	0.3	1.2
7.5	8	0.2	0.8
10.0	6	0.15	0.6

For a given axial resolution, the lateral resolution is 3–5 times larger. Focused transducer with acoustic lens decreases beam diameter and improves lateral resolution at short depth, with increased far field beam divergence. Phased array focusing reduce effective beam width and improve lateral resolution. Increase of focal zones, improve lateral resolution, but increases time and reduces frame rate.

#### *Elevational Resolution*

The elevational resolution is in the slice thickness direction which is perpendicular to the image plane. It depends on transducer element height. It is significant in volume averaging, near to transducer surface and in far field. It is worst in array transducer, but improved with fixed focal length lens. Elevational resolution is good in multiple linear arrays with 5–7 rows, known as 1.5 D transducer. Phased excitation from outer to inner arrays to minimize slice thickness at a given depth and improves elevational resolution.

#### **Temporal Resolution**

Temporal resolution is the ability to resolve events in time. It is very important in examination of highly moving organs, e.g. echocardiography. Temporal resolution is dependent on pulse repetition frequency (PRF).

#### **Contrast Resolution**

Contrast is generated by differences in signal amplitude, depends on acoustic impedance difference, density and size of scatterer, signal processing, attenuation difference, and area of low and high attenuation (fluid-filled cysts, gallstones). Microbubble contrast agent improves the visualization of vasculature, helps tissue perfusion. Harmonic imaging improves image contrast, by degrading signals from low frequency echoes. Pulse inversion harmonic imaging enhances contrast resolution.

### Noise

The sources of noise are electronic amplifiers, environmental sources (power fluctuation, equipment malfunction) and TGC which reduces contrast and increases noise with depth. A low noise, high gain amplifier is required for optimal low contrast resolution. Image processing reduces noise. Contrast to noise ratio can be increased by image processing, but decreases frame rate and spatial resolution. Low power operation ultrasound systems (OG) needs higher electronic amplification, increases noise and decreases contrast to noise ratio. This can be improved by high transmit power, but limited by bioeffect.

## ARTIFACTS

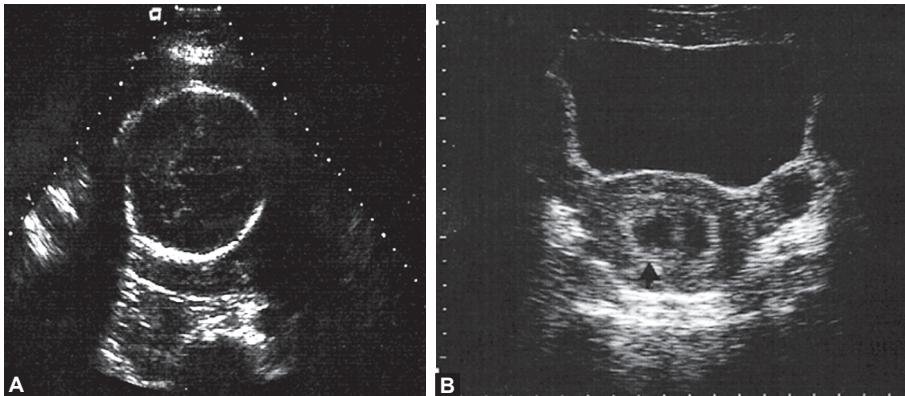
Artifact is the image formed by the echoes, which do not correspond in location, intensity or to actual interfaces in the patient. It appears in multiples, and the sources are propagation, attenuation, equipment's malfunction or design and operator error. Artifacts gives misleading information, but useful sometimes. The different types of artifacts are (i) refraction, (ii) reverberation/ring down, (iii) multiple path reflection, (iv) speckle, (v) slice thickness/width, (vi) beam lobes, and (vii) acoustic shadowing and enhancement.

### Refraction Artifacts

Refraction is the change of direction of the transmitted pulse. Refracted beam causes misregistration of echo, resulting in misplaced anatomy, e.g. eye (lens-vitreous humor) and fatty tissues. Misplacement depends on the position of the transducer and angle of incidence with tissue boundaries. It appears as misregistration, defocusing, and ghost images. Misregistration causes improper placement, and distortion of size or shape. Defocusing is due to loss of beam coherence, and cause shadowing at the edge of large curved structures (Fig. 14.25A). Ghost images are due to altered sound beam path (Fig. 14.25B).

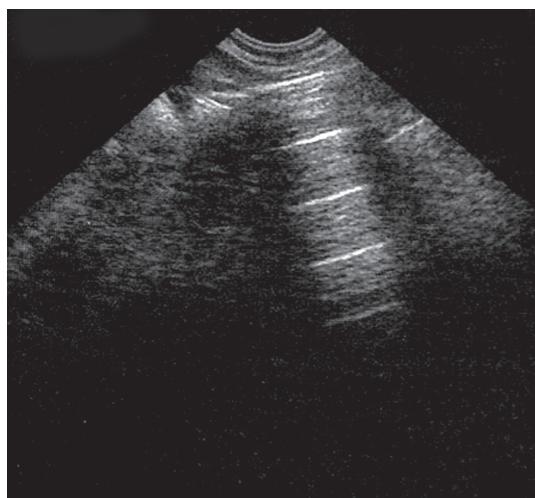
### Reverberation or Ring Down Artifacts

The reverberation artifacts are due to multiple echoes generated between two closely spaced interfaces. As a result, the US beam gets reflected back and forth and the image contains the object at regular interval. But, amplitude reduces due to reverberations. It is usually caused by reflections between transducer and strong reflector such as (i) metallic objects, e.g. bullet fragments and (ii) air pockets/partial liquid anatomy.



**FIG. 14.25:** (A) Refraction artifacts, defocusing type, cause shadowing at the edges, (B) Refraction artifacts appear as Ghost images: Single gestational sac – duplicated. Second copy of the reflector, which is side-by-side at the same depth as the true reflector

The reverberation artifacts appear as series of bright bands, parallel to sound beam's main axis with decreasing intensity and equidistant from each other. It is also called ring down or comet tail artifacts and commonly seen in images of bowel gas or sections of the bladder wall (Fig. 14.26.).



**FIG. 14.26:** Ring down artefact (reverberation) seen in bowel gas

### Multiple Path Reflection

Whenever there is oblique reflection, the beam reaches a second reflector and echo return via first reflector to the transducer. It causes ghost or apparent image, and anatomy is misplaced on the beam axis distal

to the actual position. It usually occurs near highly reflective surface, e.g. interface of the liver and the diaphragm.

### **Speckle**

If scattering surfaces are spaced at distances less than the axial resolution, echoes undergo constructive and destructive interference, that increases the noise and seen as speckle (textured appearance). The echo pattern is random, unrelated to the scattering structure, but useful to differentiate tissues. Higher frequency transducers create finer speckle patterns than lower frequency transducers. Speckle is typically associated with liver parenchyma, thyroid, heart muscle, skeletal muscle, spleen and kidney.

### **Slice Thickness**

The off axis echoes from outside the section thickness appear in regions that is echo free. Loss of signal from objects that are smaller than the volume element due to partial volume averaging, inclusion of signals from objects not in the imaging plane are the main causes. It results in false echo, not controllable by the user, e.g. lumens of large blood vessels and large cysts.

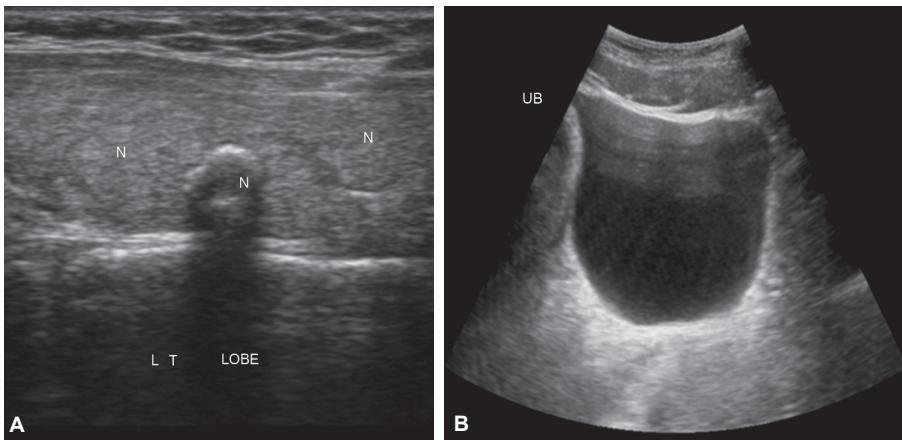
### **Beam Lobes**

Emissions of US in the off axis direction are called side lobes or grating lobes. These lobes with a strong specular reflector, add their energy to the main beam, resulting in misplaced artifacts. It is displaced laterally, and real reflector may not be seen. It may mask weak echoes that are within the main beam's axis and redirect diffuse echoes from adjacent soft tissues into an organ that is normally hyperechoic. Typical example is gallbladder, that produce artificial pseudo sludge.

### **Acoustic Shadowing and Enhancement**

Shadowing is a hypointense signal area distal to an object or interface, caused by objects with high attenuation or reflection of the incident beam. Highly attenuating objects reduce the intensity of the transmitted beam, and induce low intensity streaks in the image, e.g. bowel gas, lung, bone, gallstones, kidney stones. As a result, anechoic or hypoechoic region is seen deeper to a highly attenuating medium (Fig. 14.27).

It may prevent visualization of true anatomy and considered as a beneficial artifact. It may be divided into clean and dirty artifacts. The clean artifact may appear posterior to calcification or bone due to high percentage of absorption and reflection with no transmission. Whereas dirty artifacts appear posterior to air filled structures due to high percentage of reflection and small percentage of transmission.



**FIG. 14.27:** (A) Acoustic shadowing (thyroid),  
(B) Acoustic enhancement (bladder)

## BIOEFFECTS OF ULTRASOUND

Diagnostic ultrasound is a safe tool. deleterious bioeffects on either patients or operators have not been reported. However, any harm can be caused by diagnostic US intensities. Hence, the user has to consider issues of benefit versus risk. The main bioeffects are thermal effects, cavitations and acoustic power.

### Thermal Effects

Thermal effects are caused by absorption of ultrasound energy. This depends on the heat deposition in tissue, which will vary on rate of deposition on the focal zone, how fast the heat is removed by blood flow, etc. Absorption increases with frequency and varies with tissue type. Bone has higher attenuation coefficient, and may cause significant heat deposition in a tissue-bone interface. Heat indicators are  $I_{SPTA}$  ( $SPTA=$ Spatial peak temporal average intensity), and thermal index (TI).

Thermal index is the acoustic power produced by the transducer to the power required to raise tissue in the beam area by  $1^{\circ}\text{C}$ . It is estimated by the algorithm that takes into account the frequency (f), beam area, and acoustic output power. TI values are associated with intensity,  $I_{SPTA}$  and TI for soft tissue (TIS), bone (TIB) and cranial bone (TIC), which can be quantified. This is useful in obstetrics scanning of late term pregnancies and Doppler ultrasound.

In diagnostic US, the rise in temp is  $1\text{--}2^{\circ}$ , which is well below the potential damage level. However, Doppler study may approach these levels with high PRF and long pulse duration. The nonthermal effects

is the mechanical movement of the particles of the medium due to radiation pressure and acoustic streaming, resulting a steady circulatory flow.

### Cavitations

Cavitations are highly compressible bodies of gas/vapor, generated by the sound. It results from negative pressure, which induces bubble formation in the medium. It can be subtle, readily observable, unpredictable, and sometimes violent.

Stable cavitations refer the pulsation of persistent bubbles in the tissue that occur at low/medium US intensities. Transient cavitations may occur at high intensity, bubbles respond nonlinearly to the driving force, causing a collapse, approaching speed of sound (bubbles may dissolve, and disintegrate or rebound). Free radicals such as H, OH may perform chemical changes in the DNA. Short pulses are capable of forming transient cavitations, however, diagnostic intensity is below the threshold limit of 1kW per cm<sup>2</sup>.

The Mechanical index (MI) is a value that estimates the likelihood of cavitations by the ultrasound beam. MI is directly proportional to the peak rarefactional pressure, and inversely proportional to the square root of the frequency (MHz). An attenuation of 0.3 (dB/cm) is assumed for the algorithm that estimate the MI. As US output power is increased, the MI increases linearly. An increase in the transducer frequency decreases the MI by the square root of 4 or by a factor of 2. MI values are associated with  $I_{SPTA}$ , the measure of intensity.

### Acoustic Power

Acoustic power is the rate of energy production, and the unit is Watt (W). Acoustic intensity is the rate in which sound energy flows through unit area, and the unit is W cm<sup>-2</sup>. Acoustic power level depends on operational characteristics of the system, namely, transmit power, PRF, f, and operation mode. In diagnostic radiology, the intensity levels are kept below the threshold for established bioeffect.

Biological effects are demonstrated at higher US power levels. It includes macroscopic damage (rupturing of blood vessels, breaking up cells) and microscopic damage (breaking of chromosomes, changes in cell mitotic index). No bioeffect is seen below an  $I_{SPTA}$  of 100 mWcm<sup>-2</sup>. The levels and durations for typical imaging and Doppler studies are below the threshold for known adverse effects. Though it is safe, it should be used only in patients for whom definite benefit will be obtained.

# 15

# Magnetic Resonance Imaging

## **INTRODUCTION**

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The magnetic resonance imaging (MRI) was discovered in 1970, by Paul C Lauterbur, Stony Brook, at New York. He jointly used a radiofrequency (RF) and spatial magnetic field gradients to generate images that display magnetic properties of the proton, reflecting clinically relevant information. Basically, it is a nuclear magnetic resonance (NMR) technique, applied for human imaging. Nobel prize in medicine (2003), was awarded for the above discovery, which was shared by Sir Peter Mansfield, and Paul C Lauterbur.

Nuclear magnetic resonance (NMR) is the spectroscopic study of the magnetic properties of the nucleus of the atom (1940). Protons and neutrons of the nucleus have a magnetic field associated with their nuclear spin and charge distribution. Resonance is an energy coupling that causes the individual nuclei, when placed in a strong external magnetic field, to selectively absorb, and later release, energy unique to those nuclei and their environment.

The special feature of MRI includes (i) high contrast sensitivity to soft tissues differences, (ii) inherent safety to the patient (non-ionizing radiation), (iii) to examine anatomic and physiologic properties of the patient, and (iv) imaging of blood flow without contrast. The limitations includes (i) high equipment cost, (ii) scan acquisition complexity, (iii) long imaging time, (iv) image artifacts, and (v) patient claustrophobia.

## **MRI: THE BASICS**

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### **MAGNETISM**

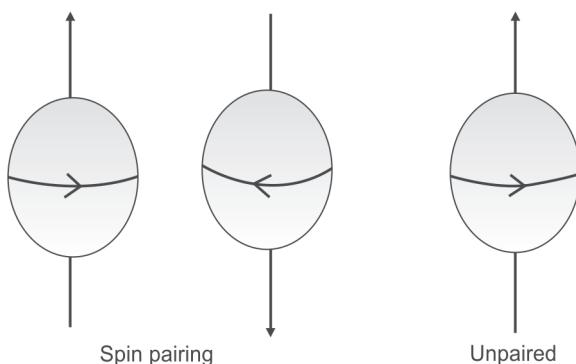
Magnetism is the fundamental property of matter, generated by moving charges. Every matter possesses magnetic susceptibility. Based on susceptibility, material may be divided into (i) diamagnetic, (ii) paramagnetic,

and (iii) ferromagnetic. Diamagnetic materials have negative susceptibility, which will oppose the applied magnetic field, e.g. calcium, and water. Paramagnetic material have slightly positive susceptibility, that will enhance the applied magnetic field, e.g. gadolinium contrast agent. Ferromagnets have higher positive susceptibility, that will enhance the external field significantly, e.g. iron, cobalt, and nickel.

The magnetic field has direction that depends on the sign and the direction of the charge motion. The magnetic field strength ( $B$ ) or magnetic flux density is the number of magnetic lines of force per unit area and its SI unit is tesla (T). The alternate unit is Gauss (G) and  $10000\text{ G} = 1\text{ T}$ .

### HYDROGEN CHARACTERISTICS

Human body has 70% water and hydrogen is abundant. Hydrogen nucleus consists of protons and neutrons. They spin continuously like a top, their spin and charge distribution gives magnetic properties. Proton has positive charge, nuclear spin, behaves as magnetic dipole and has magnetic moment. The charge in-homogeneities on subnuclear scale, gives magnetic field to the neutron. The neutron magnetic field is equal to that of proton, but in opposite direction. Hence, pairing occurs within nucleus, and cancels their magnetic moment. The unpaired spins determines nuclear magnetic moment of the nucleus (Fig. 15.1).



**FIG. 15.1:** Spin pairing in the nucleus, unpaired spins gives magnetic moment

If the total number of protons ( $P$ ) and neutrons ( $N$ ) in the nucleus is even, the magnetic moment is zero. If  $N$  is even and  $P$  is odd or

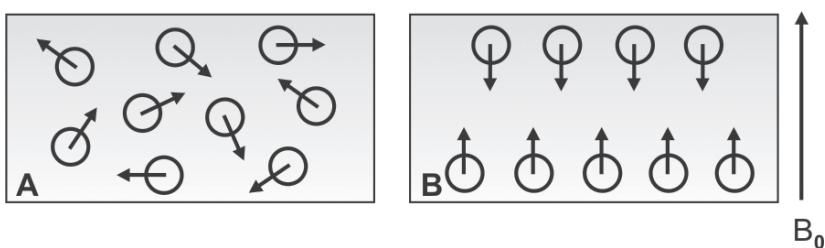
**TABLE 15.1** Physical properties of various nuclei suitable for MR imaging

Nucleus	% Isotopic abundance	Magnetic moment	Relative physiological concentration	Relative sensitivity	Gyromagnetic ratio, $\gamma/2\pi$ , (MHz/T)
$^1H$	99.98	2.79	100	1	42.58
$^{31}P$	100	1.13	$7.5 \times 10^{-2}$	$6 \times 10^{-5}$	17.2
$^{23}Na$	100	2.22	0.08	$1 \times 10^{-4}$	11.3
$^{16}O$	99.0	0.0	50	0	5.8

N is odd and P is even, the noninteger nuclear spin generates a magnetic moment. Hence,  $^1H$ ,  $^{16}O$ ,  $^{23}N$ , and  $^{31}P$  are the nuclei in the human body suitable for MR imaging. However, hydrogen scores over the other nuclei in terms of its isotopic abundance, magnetic moment, physiologic concentration and relative sensitivity (Table 15.1).

### PROTON AND MAGNETIC FIELD

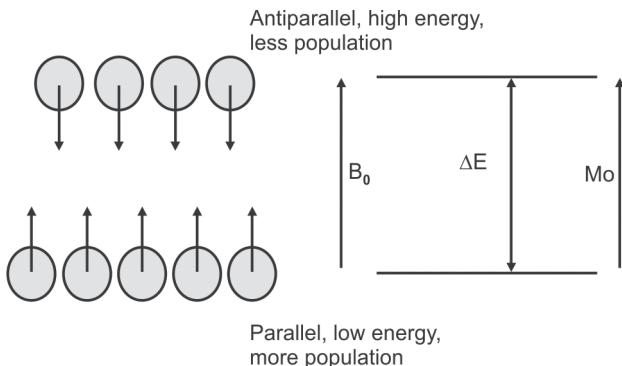
The spinning proton (spin) is considered to be like a bar magnet. The magnetic moment of a single proton is very small, not detectable, whereas billions of atoms gives measurable MRI signal. Generally, spins are randomly distributed in the tissue, which are due to thermal energy agitation, results in no tissue magnetization. Under the influence of external magnetic field ( $B_0$ ), the spins get aligned and have orderly orientation (Fig. 15.2).



**FIG. 15.2:** (A) Random orientation of spins and their magnetic moment, (B) Orderly orientation of spins in the presence of external magnetic field

Thus, spins are distributed in two energy states, namely, low energy level and high energy level. Spins having low energy align parallel to the external magnetic field and spin up wards. Spins having high energy oppose the external magnetic field and spin downwards. However, there is a slight majority of spins existing in the low energy state due

to thermal energy of the sample (Fig. 15.3). Higher the applied magnetic field, greater the energy separation, greater the excess number in low energy state.



**FIG. 15.3:** Orderly orientation and energy separation of spins, under the external magnetic field

The number of excess proton is about 3 spins per million at 1.0 Tesla magnetic field strength. In a typical MRI voxel volume, there are about  $10^{21}$  protons, resulting in  $3 \times 10^{15}$  more spins ( $3 \times 10^{-6} \times 10^{21} = 3 \times 10^{15}$ ) in the low energy state, that gives observable magnetic moment and MR signal.

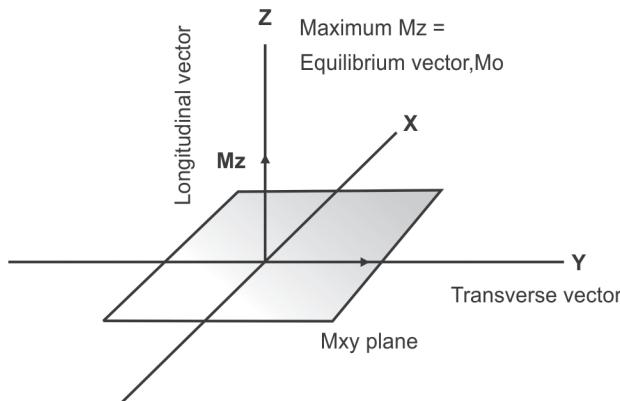
## MAGNETIZATION VECTOR

The net magnetization vector ( $M$ ) has three components, namely, (i) longitudinal ( $M_z$ ), (ii) equilibrium ( $M_0$ ), and (iii) transverse ( $M_{xy}$ ). The  $M_z$  is the component of the magnetic moment parallel to the applied magnetic field.  $M_{xy}$  is the component of the magnetic moment perpendicular to the applied magnetic field (Fig. 15.4).

Equilibrium is a condition, in which  $M_z$  is maximum and the transverse magnetization is zero and  $M_z$  is said to be equal to  $M_0$ . This is determined by the excess number of protons that are in the low energy state. At equilibrium, the vector component of the spins is oriented randomly in the XY plane, and cancel each other's magnetic moment, and hence transverse is zero.

## PRECESSION

External magnetic field not only separates the spin energy states, it also exerts a force on the proton that undergoes precession (Fig. 15.5). The direction of the spin axis tilts and rotates around the external magnetic



**FIG. 15.4:** Longitudinal magnetization vector and transverse magnetization vector of a proton

field, with fixed frequency. This precession occurs at an angular frequency ( $\omega_0$ ) that is proportional to magnetic field strength ( $B_0$ ). Larmor equation gives the relation between magnetic field strength and angular frequency

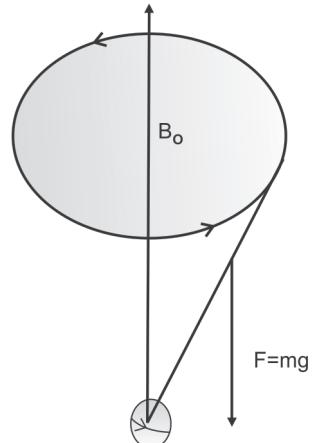
$$\begin{aligned}\omega_0 &= \gamma B_0 \\ 2\pi f &= \gamma B_0 \\ f &= (\gamma/2\pi) B_0\end{aligned}$$

where,  $\gamma$  is the gyromagnetic ratio (MHz/Tesla), and  $f$  is the linear frequency in MHz. The energy separation  $\Delta E$  is proportional to the precession frequency. Larger the magnetic field, higher the precessional frequency, and higher the energy separation. The gyromagnetic ratio is unique to each element as shown in Table 15.1.

## RADIOFREQUENCY AND RESONANCE

If a radiofrequency (RF) pulse having frequency (42.58 MHz) equal to Larmor frequency of tissue is applied perpendicular to the magnetic field, then it is absorbed by the proton nuclei and resonance occurs. At resonance, RF does two things:

1. Converts spins from low energy—parallel direction to the high energy—anti-parallel direction, and spins are said to be excited.



**FIG. 15.5:** Proton undergoes precession with Larmor frequency ( $\omega_0$ ), under the influence of external magnetic field

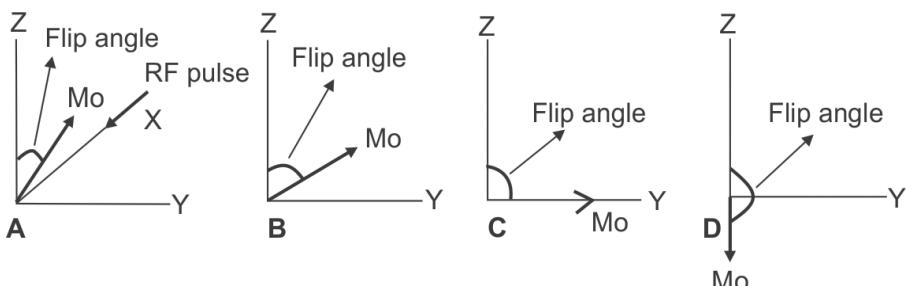
2. Resonance pulls the protons and make them in phase, but they still continue precession.

Two types of RF pulses are used in MRI, namely,  $180^\circ$  RF pulse and  $90^\circ$  RF pulse. The  $180^\circ$  RF pulse has total energy, so that it gives required energy to each proton, which tilt them by  $180^\circ$ . This reverses the magnetic vector  $M_z$  to  $-M_z$  direction. A  $90^\circ$  RF pulse is one which has energy equal to half of the total energy, which tilts half of the dipoles. Thus, application of  $90^\circ$  RF pulse will bring equal number of protons in spin up and spin down position, still in phase and continue precession. Now, the  $M_z$  is reduced to zero and the phase coherence of the dipole produces a transverse magnetism,  $M_{xy}$ , which is equal to tilting the  $M_z$  to  $90^\circ$ .

The strength of the magnetic field determines the tissue resonant frequency. The frequency increases or decreases linearly with increase or decrease of magnetic field strength. Typical magnetic field strength range for imaging is 0.1 to 4.0 T. In the case of hydrogen, the protons precessional frequency is 21.29, 42.58, 63.87 and 127.74 MHz, for 0.5 T, 1.0 T, 1.5 T and 3 T magnetic field strength, respectively.

### FLIP ANGLE

Flip angle describes the rotation through which the longitudinal magnetization vector is displaced to generate the transverse magnetization (Fig. 15.6). Common flip angles are  $90^\circ$  and  $180^\circ$  (proportional to time). A  $90^\circ$  angle (10–100 ms) provides largest possible transverse magnetization. Fast MRI technique uses smaller flip angle ( $30^\circ$ ), and generates lesser signal in the transverse. But, greater transverse magnetization per excited time is obtained. Instead of  $90^\circ$ ,  $45^\circ$  needs only half the time, but  $\sin 45 = 0.707$ , that means it gives 70% of the signal, which is quite useful.



**FIG. 15.6:** (A) Small flip angle, (B) Large flip angle, (C) Flip angle 90 degree, and (D) Flip angle 180 degree

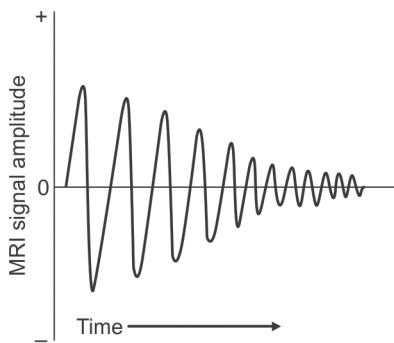
## MRI SIGNAL

As the  $90^\circ$  RF pulse is withdrawn, the perturbed system goes back to its equilibrium state. The transverse vector continues to rotate in the M<sub>xy</sub> plane, and induces an AC voltage in the receiver coil. This is the MR signal called as free induction decay (FID). This signal is also a RF having voltage in the order of mV.

The MR signal is greater when the  $90^\circ$  RF is switched off. After that, individual spins go out of phase, and return to their original orientations. As a result, M<sub>z</sub> grows and M<sub>xy</sub> decreases. Hence, induced MR signal undergoes decay, but frequency remains the same (Fig. 15.7).

Thus, MR signal from each voxel can be identified to produce grey shades in the final image. The signal produced by the  $90^\circ$  pulse depends upon the M<sub>z</sub> immediately before the pulse is applied. The signal is proportional to (i) proton density ( $\text{p/mm}^3$ ), (ii) gyromagnetic ratio of the nucleus, and (iii) magnetic field strength ( $B_0$ ).

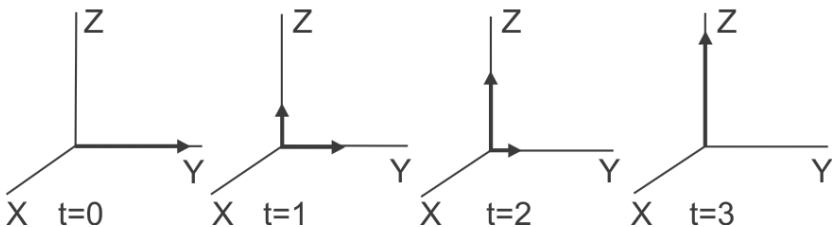
Only mobile protons give MR signal, and major part of the signal is due to water. Large molecule or bound molecule (bone) do not give signal. The air in sinuses does not have hydrogen, hence appear black. The proton density is higher in fat than soft tissue, higher in gray matter than white matter.



**FIG. 15.7:** MRI signal: Free induction decay (FID) after the withdrawal of  $90^\circ$  RF pulse

## RELAXATION TIMES

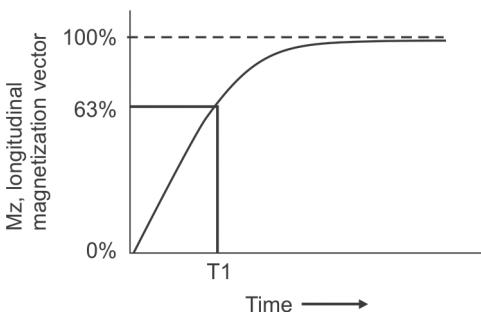
When the RF pulse is switched OFF, the magnetization vector returns to equilibrium position and the spins undergo loss of phase coherence. As a result, transverse magnetization vector decays and longitudinal magnetization vector grows (Fig. 15.8). Hence, two signals are obtained, namely, T<sub>1</sub>, the longitudinal time constant and T<sub>2</sub>, the transverse time constant. Both are affected by tissue molecular structure and chemistry. Normal and abnormal tissue can alter T<sub>1</sub> and T<sub>2</sub>.



**FIG. 15.8:** Decay of transverse magnetization with time, simultaneously longitudinal magnetization grows

### T1 RELAXATION TIME

Return of  $M_z$  to the equilibrium value  $M_0$ , require exchange of energy between spin and tissue lattice, which is called spin-lattice relaxation. It is an exponential event and measured by a time constant,  $T_1$ . The  $T_1$  is a time required to recover 63% of the longitudinal magnetization,  $M_z$  (Fig. 15.9).



**FIG. 15.9:** Spin-lattice relaxation time. Longitudinal magnetization undergoes exponential growth with time.  $T_1$  corresponds to the time, that have 63% growth of  $M_z$

The recovery of  $M_z$  versus time is given by the relation:

$$M_z(t) = M_0(1 - e^{-t/T_1})$$

where,  $M_z$  is the longitudinal magnetization that recovers after a time  $t$ , with relaxation constant  $T_1$ . When  $t = T_1$ , then  $1 - e^{-1} = 0.63$  and  $M_z = 0.63 M_0$ . Full longitudinal recovery depends on the  $T_1$  time constant. Time equal to  $3 \times T_1$ , after  $90^\circ$  pulse, 95% of the equilibrium magnetization is established. It takes  $5 \times T_1$  time, to return back to the full magnetization.

$T_1$  time of the sample can be measured by using various delay times (DT), between two  $90^\circ$  RF pulses. The longitudinal magnetization that has recovered during the delay is converted to transverse magnetization. The maximum amplitude of the FID is recorded as a function of delay time.

### T1 Relaxation and Tissue

Protons lose energy to the surrounding tissue and the rate of energy loss depends on tissue composition. T1 depends on the dissipation of absorbed energy into the surrounding molecular lattice. The relaxation time varies substantially for different tissue structures and pathologies. Energy loss is more rapid in tissues that are more complex, and T1 is short. Energy loss is slower in simple molecule like water, and T1 is long.

The energy transfer is efficient, if the precession frequency of excited proton overlaps the vibration frequency of molecular lattice. Large and stationary molecules have low vibration frequencies, little overlap, results longest T1. Small size, aqueous molecules have wide vibration frequencies (low, medium and high), little overlap, long T1. Moderately sized molecules like lipids, proteins and fats have more structured lattice. They have increased overlap, and most conducive for spin-lattice relaxation with short T1.

In general, the inability to release energy to the lattice, results in a relatively long T1 relaxation. Water has extremely long T1, addition of water soluble proteins (hydrogen layer bound the molecule), slows the molecular motion. The vibrational frequency changes from high to low, increases the overlap, and shortens the T1. Biologic tissues T1 ranges from 0.1 to 1 second in soft tissues and 1–4 seconds in aqueous tissues (CSF) and water.

Free water, urine, amniotic fluid, CSF and salt solutions have long T1. Tissue having higher % of water, always have long T1. Compact bone, teeth, calculi and metallic clips have long T1, since they are solid and rigid. T1 relaxation increases with higher field strengths. However, this increases Larmor frequency, and reduces the spectral overlap, resulting in longer T1 time (Table 15.2).

**TABLE 15.2** Variation of T1 time with magnetic field strength, for various biological tissues

Tissue	T1, ms @0.5T	T1, ms @1.5T
Fat	210	260
Liver	350	500
Muscle	550	870
White matter	500	780
Gray matter	650	900
Cerebrospinal fluid	1800	2400

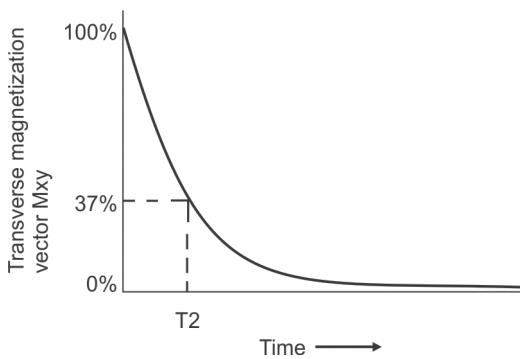
Contrast agent decrease T1 relaxation, by allowing free protons to become bound. This creates a hydrogen layer, which is known as spin lattice energy sink. Even a small amount of gadolinium contrast in pure water has a dramatic effect on T1 decrease from sec-ms.

## T2 RELAXATION TIME

Decay of transverse magnetization,  $M_{xy}$ , requires exchange of energy between spin and spin. Due to loss of phase coherence, some spins travel faster and some slower. This is called *spin-spin relaxation*, which is an exponential decay. It is measured by a time constant,  $T_2$ . It is the time taken to reduce the transverse magnetization vector to 37% of the peak value (Fig.15.10). The transverse and equilibrium vector is given by the relation,

$$M_{xy}(t) = M_o e^{-t/T_2},$$

where,  $M_{xy}$  is the transverse magnetic moment at time  $t$ . When  $t = 0$ ,  $M_o = M_{xy}$ , the transverse magnetization. When  $t = T_2$ , then  $e^{-1} = 0.37$ , and  $M_{xy} = 0.37 M_o$ .



**FIG. 15.10:** Spin-spin relaxation: Transverse magnetization undergoes exponential decay.  $T_2$  corresponds to time that have 37% of  $M_{xy}$

## T2 Relaxation and Tissue

Intrinsic magnetic inhomogeneities are a tissue character and are patient related. External magnetic inhomogeneities due to imperfections in the magnet, which accelerate the dephasing process is machine related. T2 mechanisms are determined by the molecular structure of the sample and  $T_2$  is few tens of milliseconds.

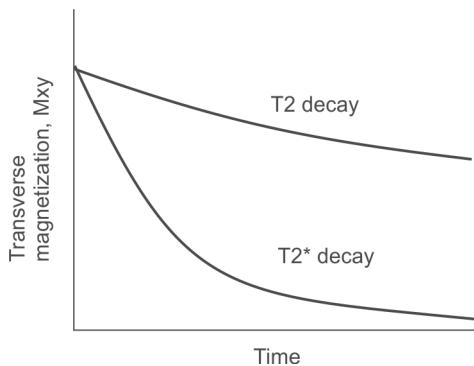
Small, mobile molecules in amorphous liquids exhibit long  $T_2$  (e.g. free water). Their fast and rapid molecular motion reduces or cancels intrinsic magnetic inhomogeneities of the spins. Lighter molecules are in rapid thermal motion, it smoothes the local field variation, hence

long T2. Thus, free water, urine, amniotic fluid, CSF, solutions of salt has long T2.

Greater the percent of free water in tissue, longer the T2, e.g. spleen > liver, renal medulla > cortex. Water bound to the surface of proteins or other large molecules, moves slowly, shorter the T2, e.g. hydrogen in fat. As the molecular size increases, its motion is constrained, intrinsic inhomogeneity is higher, and T2 decay is more rapid. Large, non moving structures with stationary inhomogeneities have a very short T2, e.g. compact bone, teeth, calculi, and metallic clips.

### T2\* Relaxation Time

Extrinsic magnetic inhomogeneities make loss of phase coherence more rapidly than from spin-spin interactions. When external magnet's ( $B_0$ ) inhomogeneity is considered, the spin-spin decay constant T2 is reduced to T2\* (Fig.15.11). The T2\* depends on the homogeneity of the main magnetic field and presence of susceptibility agents (contrast material) in the tissue.

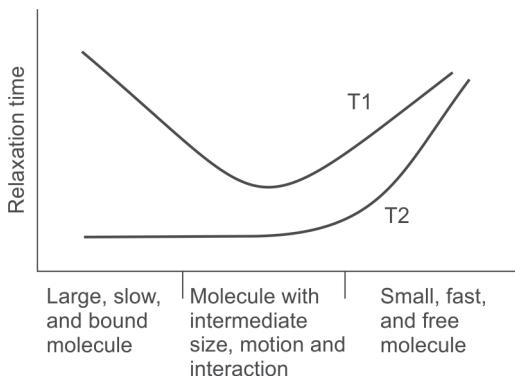


**FIG. 15.11:** T2 decays with intrinsic magnetic in-homogeneities and T2\* decays with both intrinsic and extrinsic magnetic in-homogeneities

### COMPARISON OF T1 AND T2

T1 is significantly longer than T2, and T1 time of 500 ms has a corresponding T2 time of 50 ms (5–10 times shorter). Molecular motion, size and interactions influence T1 and T2. Small molecules exhibit long T1 and long T2 (Fig.15.12). Intermediate sized molecules have short T1 and short T2. Large, slowly moving, or bound molecules have long T1 and short T2 times. Most tissues of interest in MRI involve intermediate and small sized molecules.

Hence, a long T1 usually refers a long T2, and short T1 infers a short T2. The differences in T1, T2 and T2\* provides high contrast in MRI. Magnetic field strength influences T1 (Larmor frequency), but has



**FIG. 15.12:** Molecular size, motion, interaction affects the relaxation times T1 and T2

**TABLE 15.3** T1 and T2 time for various body tissues

Tissue	T1, ms	T2, ms
Water	3000	3000
CSF	2000	150
Gray matter	800	100
White matter	650	90
Spleen	400	60
Kidney	550	60
Liver	400	40
Fat	250	80

an insignificant impact on T2 decay. Abnormal tissue tends to have higher proton density (PD), T1 and T2, than normal tissue, due to increased water content or vascularity. T1 and T2 show greater variation than PD. The range of T1 and T2 values are higher for brain, compared to CT, hence superior soft tissue contrast in MRI (Table 15.3).

Agents that disturb the local magnetic field, are either paramagnetic blood degradation products, or elements with unpaired electron spin (e.g. gadolinium) or any other ferromagnetic materials, cause significant decrease in T2\*.

## MRI SIGNAL LOCALIZATION

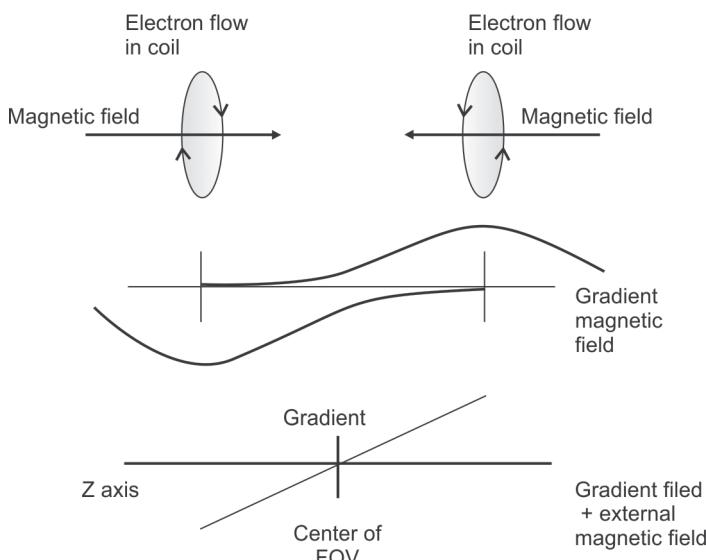
### GRADIENT FIELDS

The production of image involves: (i) collection and analysis of signal, in terms of amplitude, frequency, and phase, and (ii) make them Fourier

transformed to get individual pixels in the image. To distinguish the positions of the signal in a 3D patient, magnetic field gradients are used to localize MRI signal. These field gradients are obtained by superimposing the magnetic fields of one or more coils. Hence, RF excitations + gradient magnetic field are used to localize signal from individual voxel in patients.

Usually, gradients are sequenced in a specific order and sometimes gradients overlap partially or completely. A positive gradient field increases the external magnetic field ( $B_0$ ), and negative gradient field reduces  $B_0$  (Fig. 15.13). Three sets of gradients are used along z, y and x axis and they are named as slice selection gradient (SSG), phase encoding gradient (PEG) and frequency encoding gradient (FEG). When independently energized, each gradient produces a linearly variable magnetic field across FOV. Gradient polarity reversal is also possible.

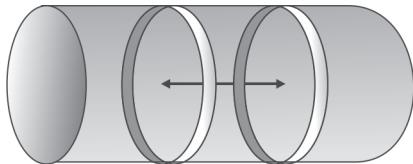
The peak amplitude of the gradient field determines the steepness of the gradient magnetic field strength and it varies from 1 to 50 mT/m. The slew rate is the time required to achieve the peak field amplitude. Shorter slow rate is always better and it ranges from 5 to 250 mT/m/ms.



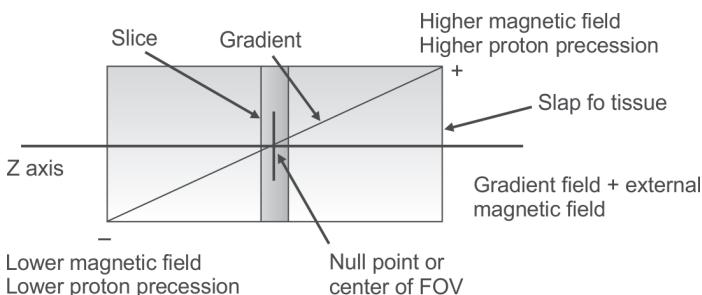
**FIG. 15.13:** A pair of coil in the Z axis is energized with DC current to produce gradient magnetic field

## SLICE SELECTION

MR image is made up of series of parallel slices, e.g. transverse slices, which are imaged in turn. Along with 90° pulse, the pair of Z coil is energized with DC supply (Fig. 15.14). This produces a controlled magnetic field gradient (SSG) along z axis (cranial-caudal). The total magnetic field decreases at the head side and increases at the toe side, remains the same at the isocenter. It varies from head to toe with constant increment of mT per m. Protons at the head side precess slowly, and are faster at the toe side, and have moderate precession at the middle (null point) (Fig. 15.15). The protons in the selected slice precess with narrow range of frequency.



**FIG. 15.14:** Z axis gradient coil, for slice selection



**FIG. 15.15:** Gradient magnetic field makes the proton faster at the toe side and slower at the head side and moderate precession at the middle (null point)

A narrow band RF pulse is applied to the whole volume and only protons in the thinner slice are excited. The spins along the gradient that have a precessional frequency equal to RF will absorb energy due to resonance. The magnetic vectors of the above spins will tip and gives MR signal.

The slice thickness depends on (i) RF bandwidth, and (ii) gradient strength across FOV. Slice thickness is reduced either by (i) increasing the gradient magnetic field or (ii) decreasing the RF band width. Thinner slice gives better anatomical detail with lesser partial volume effect, but takes longer time and typical slice thickness range is 2–10 mm. The RF may excite tissues on either side of the selected slice, since it has higher or lower frequency about the band width and generate signal. This is called cross talk, which affects the image slice. Hence, a gap equal to 10% of slice thickness should be kept in between slices. This is not required in interleaved scanning.

## PHASE ENCODING GRADIENT

The protons in the slice are excited initially by  $90^\circ$  pulse, which are in phase coherence. Now, the Y gradient coil is switched on for few ms by DC voltage (Fig.15.16). It produces magnetic field-gradient, along y direction (front to back). Spins in the upper (front) voxel precess slowly, and that are in the bottom (back) precess faster. Thus, spins in the upper voxel lags behind that of bottom voxel. Even if the gradient pulse is over, all precess at the same rate, but phase differences exist, depending on the position. Thus, MR signal has phase variation from different pixel of tissue, in the selected slice.

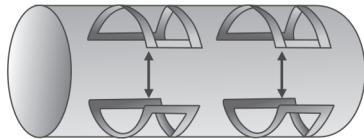
Steep gradient gives even distribution of spins, and the total signal is zero. No gradient means, all spins are in phase, which gives maximum signal. This is known as zero spatial frequency, and data are stored at the center of K space. A  $512 \times 512$  matrix may have equal number (512) of spatial frequencies. Phase encoding must be repeated in steps (256) of gradient increments, which will step up the phase shifts. The pixel size = FOV/number of phase encoding gradients.

## FREQUENCY ENCODING GRADIENT

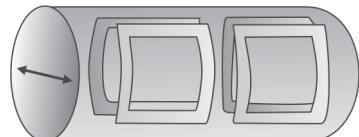
When the PEG is applied, X-gradient coil (Fig.15.17) is energized, applied in orthogonal direction (patient side-side). Protons in a vertical column experience same magnetic field and emit MR signal of same frequency. Spins in the left side precess slowly, and that in right precess faster, resulting in a frequency gradient from left to right. The MR signal from a given slice consists of range of RF frequencies, on either side of the applied pulse.

## COMPOSITE SIGNAL

The composite signal emitted by a whole slice comprise of spectrum of phases and frequencies. Computer has to analyze the signal for frequency and phase. The signal is amplified, digitized, and decoded by Fourier transform. Signal amplitude is dependent on spin density and T1 and T2 times. Fourier transform is a mathematical process



**FIG. 15.16:** Y-axis gradient coil, for phase encoding



**FIG. 15.17:** X-axis gradient coil, for frequency encoding

which converts a signal (amplitude vs distance) into a signal of amplitude vs frequency. This step makes the signal easy to store and digitize.

### SINC PULSE, K-SPACE, IMAGING TIME

Tissue slice is considered as rectangular shape. To excite a rectangular slice, RF pulse requires synthesis of a specialized waveform called *sinc pulse*. Its width determines the output frequency bandwidth (BW). A narrow sinc pulse and high frequency produce a wide BW. A broad and low frequency sinc pulse gives narrow BW, resulting in an increased SNR. Narrow BW is not always good, it results in chemical shift artifact and other undesirable image characteristics. There is trade-off in image quality and hence optimal RF bandwidth and gradient field strength is required.

The data in the signal is stored in the K-space, which is a spatial frequency domain in the computer. K-space stores signal spatial frequencies and their origin. Spatial frequency corresponds to image brightness. The number of lines filled in K-space is equal to the number of encodings in the sequence. The center K-space consists of shallow gradient data with low spatial frequency, less details, and strong signal. Upper and lower K-space stores steeper gradient data with high spatial frequency, better details, and weak signal.

Imaging time is equal to number of signal averages (excitations)  $\times$  number of phase encoding steps  $\times$  pulse repetition time (TR). Increase of excitation reduces noise, and increases the imaging time.

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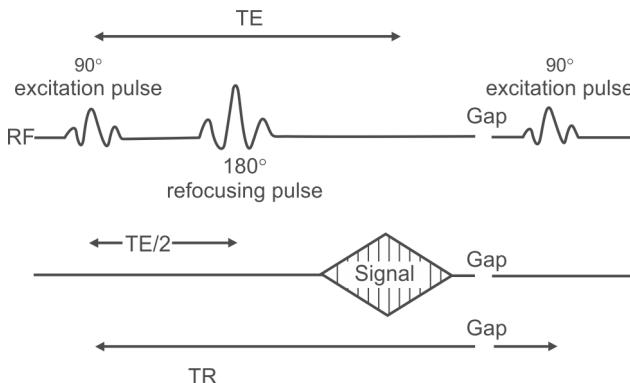
## MRI IMAGING SEQUENCES

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Sequence in MRI refers timing, order, polarity and repetition frequency of RF pulses and applied magnetic field gradients. It makes the signals dependent of T1, T2 or spin density relaxation characteristics. The various sequences include (i) spin echo pulse sequence, (ii) inversion recovery, and (iii) gradient recalled echo.

### SPIN ECHO SEQUENCE

Spin echo pulse sequence nullify the external magnetic field variations, due to imperfection of magnet, distortion due to patient (susceptibility), and gradient across voxel. As a result, some dipoles precess faster than others, after 90° RF removal. In this sequence, 90° RF pulse is followed by a 180° pulse with time gap (TE/2) and the signal is measured after a time, TE (time of echo). The above is said to be one cycle, hundreds of such repeated cycles gives one MR image (Fig. 15.18).



**FIG. 15.18.** Pulse echo sequence principle and one cycle events

### Imaging Cycle

The events that are happening in 1 cycle are: (i) Just after the 90° pulse, dipoles are in phase, the transverse magnetization is maximum, and the FID signal is not measured. (ii) Dephasing occurs, spins precessing faster and slower, and the FID signal decays with  $T2^*$ . (iii) When 180° pulse is applied, dipoles tip from spin up to spin down position. It turns the individual vector to rotate by 180° in X direction. Fast spin become slow and slowspins become fast, and rephasing takes place and signal grows. (iv) After a time (TE/2), spins are in phase, MR signal is in peak and spins goes out of phase and MR signal decays.

### Rephasing Pulse and TR

The 180° RF pulse is called as rephasing or refocusing pulse and it reverses and eliminates effect of external magnetic field inhomogeneities. The MR signal appears as an echo of initial FID, and longer the TE, smaller the MR signal. The TR is called Time of repetition. The pulse echo sequence uses a series of 90° pulses separated by a period known as the time of repetition (TR). The TR ranges from 300–3000 ms. Subsequent 90° pulse is applied before the complete longitudinal magnetization recovery of the tissues.

### TE and MR Signal

The TE (Time of echo) is the time between 90° pulse and production of spin echo signal. The time delay (TE/2) is the time between 90° and 180° RF pulse and TE ranges from 5–140 ms. The MR signal

depends on (i) spin density ( $\rho H$ ), (ii) signal from fluid flow,  $f(v)$ , (iii) T1 and T2 (tissue properties), and (iv) TR and TE (machine properties). In general, the MR signal (S) is given by the equation:

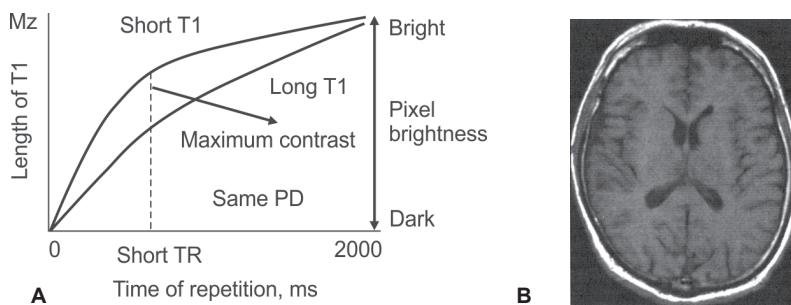
$$S \propto \rho H f(v) \{1 - e^{-TR/T1}\} e^{-TE/T2}$$

### Image Contrast

MR image consists of three tissue properties, namely, PD, T1, and T2. Whereas TE, TR are the machine parameters, that weigh the contrast in the image. Brightness of the pixel depends on (i) proton density, (ii) recovery of  $M_z$  (length of T1, compared to TR), and (iii) decay of  $M_{xy}$ , (length of T2, compared to TE). The selection of TR and TE is critical, so that the image brightness depends on any one of the tissue parameter T1, T2, PD. Images can be obtained as T1 weighted image, or T2 weighted image and or proton density weighted image.

#### T1 Weighted Image

T1 weighted image produce contrast based on T1 characteristics of tissues by de-emphasizing T2. It employs short TR, (300–800 ms) to maximize the contrast and short TE, (15 ms) to minimize T2 dependency. The image contrast is due to recovery properties of T1 and shorter the T1, brighter the image. In brain, cerebral tissues, fat, white matter, gray matter, and CSF are well distinguished in T1 weighted image (Fig.15.19). Fat is most intense and appear as white, but CSF lowest signal and appear as black.

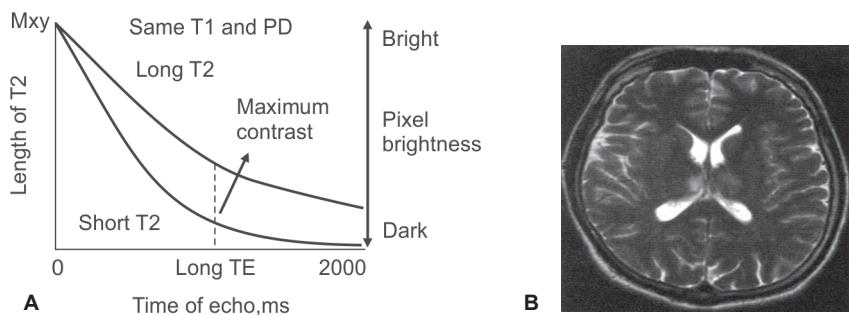


**FIG. 15.19:** (A) T1 contrast vary with TR, maximum contrast for short TR, for a constant PD, (B) T1 weighted image of brain, short TR = 500 ms, and short TE = 8 ms

#### T2 Weighted Image

T2 weighted image produces contrast based on T2 characteristics of tissues, by de-emphasizing T1. It employs long TR (1000–2000 ms)

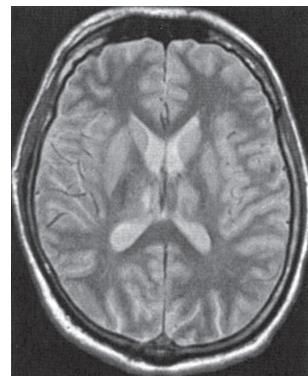
to reduce T1 contrast and long TE (90–140 ms), to maximize T2 contrast. The image contrast is due to recovery properties of T2, and longer the T2, brighter the signal. In T2 weighted brain image, CSF is brighter than Fat (Fig. 15.20).



**FIG. 15.20:** (A) T2 vary with time of echo, long TE gives maximum contrast, for constant T1 and PD, (B) T2 weighted image of brain, long TR = 2400 ms, and long TE= 90 ms

#### *Proton Density Weighted Image*

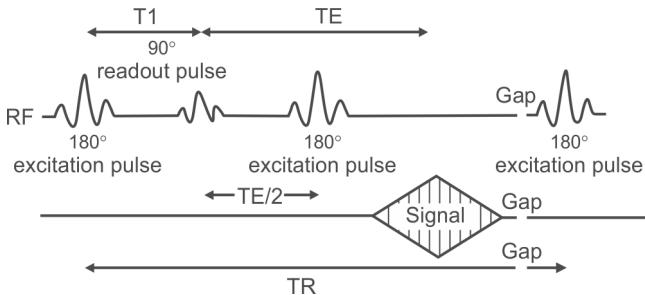
Spin density weighting mainly relies on differences in the number of protons per cc. Greater the spin density, larger the longitudinal magnetization (e.g. lipids, fats). It employs long TR (1000–3000 ms), to minimize T1 contrast and short TE (15 ms), to minimize the T2 effects. The signal strength is greater with less noise. Higher the PD brighter the image, and hence CSF appears white, but white matter appears black. Though the signal to noise ratio (SNR) is higher, the image contrast is poor (Fig. 15.21).



**FIG. 15.21:** Proton density weighted brain image with long TR = 2400 ms and short TE = 30 ms

#### **INVERSION RECOVERY**

Inversion recovery emphasizes T1 weighting and the amplitude of  $M_z$  is greater than by 2. The use of initial  $180^\circ$  pulse, tips the spins anti-parallel to z axis and inverts the  $M_z$  to  $-M_z$ . The spins return to parallel, by spin-lattice relaxation, and  $M_z$  recovers after a time  $0.69 \times T_1$ . After a delay time  $T_I$  (Time of inversion, 500 ms), a  $90^\circ$  RF pulse is applied, which tilt the available  $M_z$  to the transverse plane and FID signal is produced. A second  $180^\circ$  pulse at time  $TE/2$ , produces an echo signal at  $TE$  (Fig. 15.22).



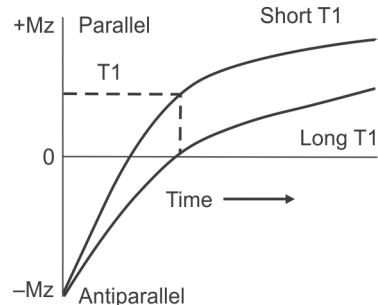
**FIG. 15.22:** Inversion recovery imaging sequence principle and one cycle events

One cycle in inversion recovery is  $180^\circ$ ,  $90^\circ$ , and  $180^\circ$  RF pulses and the cycle is repeated after TR (1000 ms). Echo amplitude depends on TI, TE, TR and  $M_z$  magnitude. It employs short TE (20 ms), to minimize  $T_2$  dependency. Basically, inversion recovery gives  $T_1$  weighted image and tissue with longer  $T_1$  is suppressed (Fig. 15.23). Longer the TI or shorter the T1, greater the MR signal and thus TI controls the tissue contrast. This technique is time consuming, but gives better gray and white matter discrimination.

### STIR and FLAIR

The Short tau inversion recovery (STIR) is mainly used for fat suppression. Fat produces bright intense signal which obscure contrast in other tissues. It employs short TI (125 ms) and TR (2500 ms) and eliminates chemical shift artifacts.

The Fluid attenuated inversion recovery (FLAIR) uses longer TI (2400 ms) and TR (7000 ms), to reduce the signal levels of CSF and other tissues with long  $T_1$  relaxation constants. It reduces CSF signal and other water-bound anatomy in the MR image by using a TI selected at or near the bounce point of CSF.

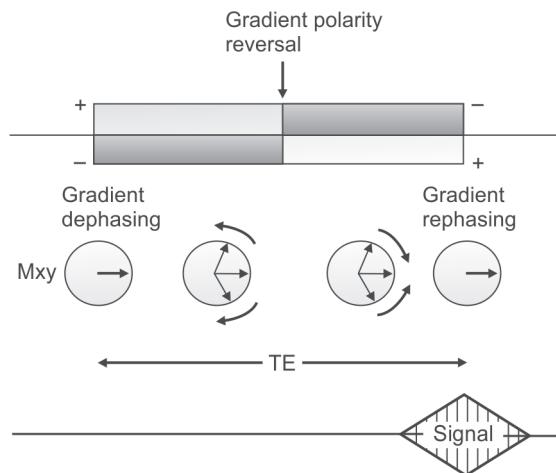


**FIG. 15.23:** Tissue with short  $T_1$  gives large MR signal for a given TI. Tissue with long  $T_1$  yet to recover, and get suppressed

### GRADIENT RECALLED ECHO

The gradient recalled echo (GRE) uses a magnetic field gradient to induce echo, instead of  $180^\circ$  pulse. The magnetic field gradient is reversed

after a delay time of TE/2 (Fig. 15.24). The GRE is not a true spin echo technique but a purposeful dephasing and rephasing of the FID. The magnetic field inhomogeneities and tissue susceptibilities are emphasized in GRE and hence the image is T2\* weighted.



**FIG. 15.24:** Gradient recalled echo imaging principle

Depending on the desired contrast, flip angles of 10–90° are applied. With short TR (< 0.2 seconds), smaller flip angles are used and long TR (> 0.2 seconds) uses flip angle > 45°. More transverse magnetization is generated with smaller flip angles, due to quick build up of magnetization in tissue. Tissue contrast in GRE pulse sequences depend on TR, TE, and the flip angle. In this, RF pulse of smaller strength is used. It is useful with shorter TR times, but long TR imaging will differentiate T2 and T2\* and moving blood appears bright. GRE has number of specialized sequences, that includes:

1. GRASS: Gradient Recalled Acquisition in the Steady State.
2. FISP: Fast Imaging with Steady State Precession.
3. FAST: Fourier Acquired Steady State.
4. SPGR: SPOiled Gradient Recalled Echo.

## SPECIALIZED MR SEQUENCES

In addition to spin echo, inversion recovery, and GRE sequences, there are few sequences that can generate specific information about the tissue, molecular nature and structure. They include (i) magnetic

resonance angiography, (ii) perfusion imaging, (iii) diffusion imaging, (iv) MR spectroscopic imaging.

## **MAGNETIC RESONANCE ANGIOGRAPHY**

The magnetic resonance angiography (MRA) employs *blood flow enhancement* as the basis. The flow of blood depends on velocity, flow profile, direction relative to slice, pulse sequence and its parameters and slice acquisition and direction of flow. Signal from flow can be used to produce MR angiographic images.

The relative saturation of the surrounding tissue and the blood entering the slice volume decides the signal. Blood entering the slice contain unsaturated spins and blood within the slice contains saturated spins. Blood leaving the slice removes spins of different saturation conditions. The blood outside the volume does not interact with RF field and enter the volume and produce large signal, compared to spins within the volume. This effect is higher in slow laminar flow and hence veins appear bright. Since aorta has fast flow, it appears dark or void. Turbulent flow cause rapid loss of coherence and appear as dark.

Sometimes, flow related enhancement is undesirable. Presaturation pulses are applied to volumes above and below the imaging volume. This can be achieved by GRE sequence and hence veins, artery and CSF appear bright. This is helpful to avoid motion artifacts and depends on blood velocity, slice thickness and TR. MRA technique can be done either by phase contrast angiography or time of flight angiography.

## **PERFUSION IMAGING**

Perfusion facilitates delivery of oxygen, nutrients and removal of waste such as CO<sub>2</sub>. Perfusion measurement reveals the rate of blood delivery to the capillary bed. It is a measure of metabolic activity and it can be done either by using bolus of contrast or arterial spin labeling. The first method uses paramagnetic contrast agent, such as diethylenetriaminepenta-acetic acid (Gd-DTPA) to carry out measurements. The contrast agent modify the relaxation of protons in the blood and shorten T2\*, which causes changes in signal. Thus, pre- and post-contrast images reveal the level of perfusion. This procedure is contraindicated in patients with renal dysfunction.

In arterial spin labeling perfusion techniques, two images are obtained. The first image is obtained by spin inversion technique (spin labeled image) and the second image without inversion (control image).

Subtraction of the two images gives perfusion image, which is a measure of blood flow.

### **Functional Imaging**

Functional imaging (fMRI) is based on increased blood flow in the brain area due to natural activity. This reduces deoxyhemoglobin levels, which is a paramagnetic agent that alters T2\*. Hence, blood oxygen level is compared between stimulus and rest, to study brain function. The effects are short lived, and require rapid imaging sequences, e.g. EPI, fast GRE.

At rest oxyhemoglobin and deoxyhemoglobin levels are equal. During activity, more oxygen is extracted from the capillary, resulting increased blood flow that causes change in deoxyhemoglobin. Oxyhemoglobin (fully oxygenated blood) is a diamagnetic and has no effect on signal. Deoxyhemoglobin (reduced hemoglobin) is a paramagnetic agent, due to unpaired electrons and produce magnetic inhomogeneities in tissues and increases T2\*. Thus, deoxyhemoglobin serve as an *in vivo* positive contrast agent in functional MR study. Its concentration is influenced by variation in oxygenation and tissue metabolism. The acquisition is named as blood oxygen level dependent (BOLD).

In BOLD technique, multiple T2\* weighted images of head are produced at rest. Later, the patient is subjected to stimulus and again multiple images are obtained. The rest image data set is subtracted from the stimulus data, by voxel by voxel. The brain activity is represented by change in signal in a specified area of the brain.

The stimulus may be finger movement, light flashes or sound. The areas of activity in the brain are statistically analyzed and color coded. The other areas are not color coded. The resultant image is superimposed on a gray scale brain image, to obtain functional map of the patient.

### **DIFFUSION WEIGHTED IMAGING**

Water mostly gives MRI signal from normal and diseased tissue. In normal state, water has random motion due to thermal energy (Brownian motion). In tissue, restriction of molecular motion is there and hence the damping effect called diffusion exist. In many tissues, diffusion is isotropic. In some tissues, the diffusion is in preferred direction, and hence, it is anisotropic (e.g. muscle, white /gray matter). In diffusion weighted imaging (DWI) strong MR gradients are applied to produce signal differences based on the mobility and directionality of water diffusion. Normal tissue water molecule has more mobility, have greater

loss of signal. Abnormal or injured tissue water molecules have lesser mobility, resulting in less signal loss. Thus, DWI technique can estimate diffusion coefficient, which is a measure of molecular motion. Diffusion imaging techniques includes (i) echo planner imaging and (ii) navigator imaging techniques.

Echo planner imaging (EPI) acquires a complete image in a single shot and it is sensitive to magnetic field inhomogeneities. Image distortion artifacts are present due to susceptibility variation at interfaces, e.g. air, bone and soft tissue. The image is noisy and spatial resolution is limited, and hence signal averaging is required.

Navigator methods acquire images in multiple shots, uses navigator MR signals for each shot, to detect and correct the bulk motion. It gives improved spatial resolution with minimal image distortion artifacts and SNR. The required acquisition time is about 10 minutes and mainly used in cardiac study with ECG gating. It is less prone to ghosting artifacts, due to patient motion.

The perfusion imaging distinguishes areas of rapid and slow proton diffusion. Volume in which protons are more mobile will show increased signal, compared to that of less mobile. It is a valuable diagnostic tool in stroke patients, neurology and the patient is held by straps firmly during imaging.

## **MR SPECTROSCOPIC IMAGING**

MR spectroscopic imaging (MRS) is a method of measuring tissue chemistry. It provides frequency spectrum of the tissue based on the molecular motion and composition. The peak intensities and position in the spectrum indicate how the atom is bounded to a molecule. The metabolites peaks, caused by frequency shifts are analyzed.

The electron cloud shielding around a nuclei causes slightly different resonance frequencies. This is compared with standard frequency and the shift, known as chemical shift is obtained. The chemical shift lies between water and fat. Since the amplitudes of water and fat are greater, they need to be suppressed. Then, the signal is Fourier transferred and plotted as frequency spectrum.

MRS can be performed either with single voxel or multiple voxel with a minimum volume of about  $1 \text{ cm}^3$ . In a single voxel, stimulated echo acquisition mode (STEAM) or point resolved spectroscopy sequence (PRESS) is used. The study reveals the presence of pathology, relationship between choline (Cho) concentration, creatine (Cr) and

N-acetylcysteate (NAA) as ratios. Higher Cho concentration with depressed NAA and Cr reveals presence of tumor. Lipid peak indicates hypoxia and high grade tumor.

Thus, MRS is useful to identify metabolic disorders, infections, and treatment evaluation. To study the individual metabolism, sequential imaging with phase encoding gradient is used. High magnetic field (3 T) is required for good spectral resolution and the field must be uniform, about 1 ppm. To reduce imaging time, larger pixels (1 cm) are used.

## OTHER IMAGING TECHNIQUES

The other MR imaging techniques includes (i) multislice technique, (ii) multiecho technique, (iii) fast spin echo, (iv) echo planner imaging, (v) 3-D Fourier imaging and (vi) parallel imaging.

In a multislice technique, a successive  $90^\circ$ ,  $180^\circ$  pulses of differing frequency, excite a series of up to 32 separate slices, before repeating the first slice. In multiecho technique, after  $90^\circ$  pulse, 2 or more  $180^\circ$  pulse produce successive echoes with increasing TE, at the rate of two images per slice. In fast spin echo  $90^\circ$ ,  $180^\circ$ ,  $180^\circ$  can be modified to produce 4–16 echoes by different phase encoding gradients. It reduces the imaging time by a factor of 4–16 and fat produces high intense signal and muscle often darker. Increased matrix size is used, to improve resolution.

Echo planner imaging (EPI) is a fast form of GRE (50 ms). After a  $90^\circ$ ,  $180^\circ$  spin echo sequence and slice selection, polarity of the frequency encoding gradient is reversed fast, and the phase encoding gradient also switched on and off. Multiple echoes are obtained, before the decay of Mxy, and take 2–3 s for brain.

In 3-D Fourier imaging, shallow z gradient is used to select a thick slice, in order to cover the whole volume. Frequency encoding is done along one axis and phase encoding is done in other two axes. The data processing and scan time is too long and can be reduced by GRE with short TE. The 3-D FT is used to decode information and it involves motion and wraparound artifacts.

Parallel imaging replaces multislice and high gradient strength MR techniques, to give high temporal and spatial resolution. There are two techniques, namely, SMASH (simultaneous acquisition of spatial harmonics) and SENSE (sensitivity encoding). Both use an array of RF detection coils, to perform phase encoding. Each coil in an array is connected to a separate RF receiver, giving parallel set of data,

and each one produces separate image. Images are obtained with fewer phase code steps. EPI can be used with SMAH or SENSE, to reduce the imaging time to half.

## **MRI INSTRUMENTATION AND BIOSAFETY**

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### **MAGNET**

The magnet is the heart of the MR system and the patient is placed inside the magnet, surrounded by set of coils connected to RF generator. The type of magnet includes (i) permanent magnet, (ii) resistive electromagnet and (iii) superconducting electromagnet.

#### **Permanent Magnet**

The permanent magnet consists of two flat opposing pole pieces (iron, alloys Al, nickel, cobalt). It is expensive, but cheaper in running cost. It requires no power, uses low strength, vertical magnetic field up to 0.3 T. No claustrophobia issue, suitable for children, aged, and interventional work.

#### **Resistive Electromagnet**

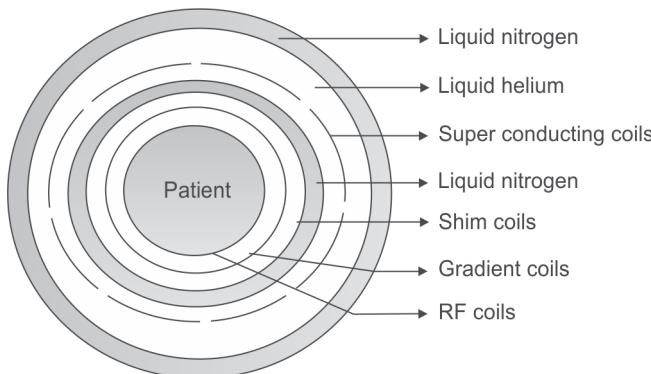
Resistive electromagnet has set of coils run by direct current with 50–100 kW (Al or copper). It produces heat, and require water cooling. It can provide both vertical and horizontal magnetic field up to 0.5 T, and has no fringe field. It can be switched off during emergency, cheapest, smaller, and weighs 2 ton.

#### **Superconducting Magnet**

Super conducting magnet is made by a direct current solenoid (niobium-titanium alloy in copper matrix). It is basically an air core cylinder of 1 m diameter and 2–3 m depth (Fig. 15.25). It is cooled by a cryogen, liquid helium at 4 K ( $-269^{\circ}\text{C}$ ). It has negligible resistance, and large current can be used without overheating. It provides horizontal fields up to 3.0 T with high field uniformity (1 ppm in a  $40\text{ cm}^3$ ). It is large in size, expensive, claustrophobia to patients and weighs about 6 ton. It takes hours to cool and current build up. Current flows, even with no power, but consume cryogen liquid.

To shut down, the stored electromagnetic energy in the coil has to be removed carefully, to avoid quench. The liquid helium is kept in a cryostat, replenished periodically. Refrigerator system is used to reduce helium losses. The air entering the system may solidify

and the coolant level must be checked daily. If level is too low, quench occurs, temperature rises, and superconductivity will be lost. As temperature rises, liquid boils rapidly, vented outside the building causing hazards. As superconductivity fails, copper do the conduction of current.



**FIG. 15.25:** Cross-sectional view of MR gantry degin

The disadvantages of super conducting magnet includes (i) high initial capital and sitting cost, and cryogen cost, (ii) difficulty in turning off the field, (iii) extensive fringe field, and (iv) uncontrolled quenching due to boiling of helium.

### Magnetic Field

To have good performance, the magnetic field strength, temporal stability and its field homogeneity are very important. Field should be uniform up to 5 ppm and 1 ppm for MR spectroscopy over a large volume. The fringe fields are negligible in permanent magnet, and greater in resistive and superconducting magnets. Higher the field, larger the MR signal, and greater the SNR. But, higher magnetic field increases T1, and require longer TR and imaging time.

### Coils

MRI machine has three important coils, namely, (i) shim coils, (ii) gradient coils, and (iii) RF coils. Shim coils works with direct current, and make the main magnetic field uniform throughout the imaging volume.

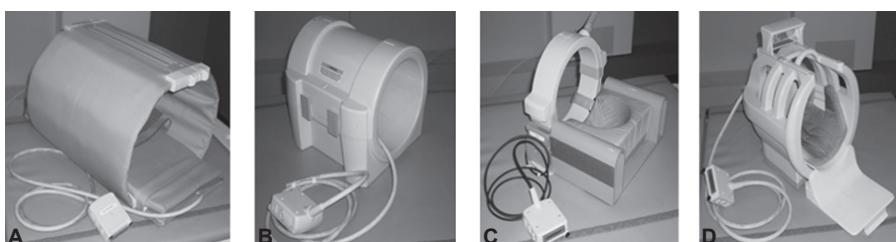
Gradient coils provide gradient magnetic field up to 20 mT/m. There are 3 sets of gradient coils, carrying direct current. All coils are connected to amplifiers, which control the rise time and maximum value of the

gradient. These coils are switched off rapidly (about 1 ms), which makes loud noise.

RF coils are made for either (i) transmit and receive or (ii) receive only coils. RF produces magnetic field perpendicular to the main magnetic field. It should be close to the imaging part. Types of RF coils includes (i) body coil, (ii) head coil, (iii) surface coil, (iv) phased array coils, and (v) transmit phased array coils (Fig. 15.26). Body coil is a standard one permanently fixed in the gantry. It transmits RF and receive MR signal, e.g. chest, abdomen. Head coil is used for brain imaging, and it transmits and receives the signal.

Surface coils are basically receiver only coils, used closer to the imaging part, e.g. lumbar spine, knee, and orbit. They receive signals effectively from a depth provides smaller voxel, better resolution, but have smaller FOV and less uniformity. Receiver coils provide larger signal, lesser noise, and improve SNR. In phased array, four or more receiver coils are used and they receive signal individually and then combine the signal. This makes the signal less noise, high SNR with large FOV.

Transmit phased array coils produce current on each element and have special amplifiers. The control of amplitude and phase make them independent. It employs reduced pulse duration, with higher SNR, improved field homogeneity, reduced specific absorption rate and it is useful in parallel MRI (SENSE).



**FIG. 15.26:** (A) Body coil, (B) Knee coil, (C) Head and shoulder coil, and (D) Head coil

### Biosafety

In MRI, ionizing radiation is not used. Hazards arise from static magnetic field, gradient field and RF field, cryogenic liquids, claustrophobia, and noise.

### **Static Magnetic Field**

Static magnetic field induces voltage in blood, resulting in depolarization in moving heart muscle. Pregnant patients should not be exposed > 2.5 T. The limit for hospital staffs are (i) whole body < 2T, (ii) limbs < 5 T, and in 24 hours they should not be exposed to more than 0.2 T.

Patients with implants, prostheses, aneurysm clips, pacemakers, heart valves, etc. should be away from MRI area, where fringe fields are > 0.5 mT. Ferromagnetic material brought into the imaging room are attracted to the magnetic field (e.g. IV pole), and can become a deadly projectile to the patient.

### **Gradient Magnetic Field**

Gradient magnetic fields produce eddy current in conductive tissue, causing nerve stimulation, involuntary muscular contraction, breathing difficulties, and ventricular fibrillation (< 60 T/s). Patient with heart disease should be careful. Non-metallic implant materials can lead to significant heating, due to rapidly changing gradient fields.

Induced current produces flashes of light on the retina, vertigo, which results in nausea, sensation of metallic taste. Hence, MRI is not recommended during first trimester. Induced voltages may affect implanted devices and monitoring instruments, e.g. cochlear implants, cardiac pacemaker and electrocardiography monitor. Therefore, MR compatible devices should be used. Noise may bring irreversible damage and the limit is 140 dB. However, for hearing protection, it should be < 90 dB. Ear plugs may reduce noise by 10–30 dB. Low frequency sound may transmit to the fetus, and damage hearing, but it is not proven.

### **Fringe Field**

Tiny magnetic field around the magnet are called fringe fields, e.g. 1.5 T magnet has 1 mT fringe field at 9.3 m, and 0.5 mT at 11.5 m. Extensive fringe fields create hazardous conditions in adjacent areas and cause disruption of electronic signals and sensitive electronic devices. Gamma cameras, image intensifiers, color TVs are severely impacted by fringe fields < 0.3 mT. Areas above 1.0 mT require restricted access with warning signs. Disruption of the fringe fields can reduce the homogeneity of the imaging volume (automobile, elevator, etc.). Fringe field can affect watches, erase CD data and credit cards, distort video display, and photo multiplier tubes (PMT).

### **RF Field**

Heating may occur at high frequencies with strong magnetic field, resulting in rise of temperature of skin and rectum. Heating may affect

metallic implants, cornea and testes. The RF heating effect is quantified by specific absorption ratio (SAR), which is the RF energy deposited per unit mass. SAR limit for the whole body is  $1 \text{ Wkg}^{-1}$ , it restricts temperature within  $0.5^\circ$ . SAR is higher for large body site, higher magnetic field and high conductivity tissue (brain, blood, liver and CSF). It is also higher for a  $180^\circ$  pulse, than  $90^\circ$  pulse, spin echo sequence than GRE.

Similarly, RF signals entering the room or leaving the room may also create a hazardous condition. This can be avoided by construction of Faraday cage. The MRI room walls are shielded with copper sheet and windows with copper wire mesh. This will prevent entry of RF signal from outside.

### **Projectile Effect**

Attraction of ferromagnetic object varies with square of the magnetic field and inversely with cube of the distance. Scissors, scalpels become lethal projectiles in MRI room. Oxygen cylinder, patient bed, and fire fitting apparatus have caused major injuries in the past.

### **Safety Limit**

MRI is extremely safe when used within the regulatory guidelines of FDA, USA. The guidelines recommend the limit for static magnetic field as 2.0 T, changing magnetic field as 6T per second, RF power deposition as  $0.4 \text{ Wkg}^{-1}$  and acoustic noise levels as 200 Pascal.

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## **MRI IMAGE QUALITY AND ARTIFACTS**

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### **MRI IMAGE QUALITY**

The image quality in MRI is controlled by (i) spatial resolution, (ii) contrast, (iii) signal to noise ratio, and (iv) scan time.

#### **Spatial Resolution**

Spatial resolution is controlled by the pixel size, which depends on matrix size and FOV, e.g. a 25 cm FOV and  $256 \times 256$  matrix, will have the pixel size of 1 mm. Use of larger matrix size, or smaller FOV, use of local coil, reduce pixel size and improve resolution. Use of thinner slice and 3D data acquisition, compared to 2D will also improve resolution. Higher magnetic field will provide better resolution, it is partially true. Higher magnetic field, larger the SNR, thinner the slices and reduction of partial volume effects, resulting in improved

resolution. However, higher magnetic field increases RF heating create artifact and lengthens T1.

### **Contrast**

Contrast is the difference in SNR between adjacent tissues. It can be enhanced by image weighted technique (T1, T2, and PD) or contrast. Off-resonance RF frequency can be used, to get signal from free protons. It suppresses the signal from proton bound to macromolecules. Fat suppression with STIR, will enhance contrast between lesions and adjacent fatty tissue. T2 weighted image enhances the contrast between normal and abnormal tissue (bright).

Paramagnetic gadolinium shortens T1, in the adjacent tissue, increases inherent contrast. Super paramagnetic, iron oxide ( $\text{Fe}_3\text{O}_4$ ), and dysprosium ( $\text{Dy}^{3+}$ ) produce large local magnetic field and shorten T2 and  $\text{T2}^*$  and the areas appear black. Hyperpolarized (xenon-129 exposed to laser) gas as contrast, dissolves in blood and shows large chemical shift. It can be used with low magnetic field, to give large SNR, e.g. lung, low field angiography and spectroscopy imaging.

### **Noise**

Noise is a random variation of the MR signal, present in all frequencies, in all the time. Noise reduces contrast and it is worst with low proton density and low signal. SNR can be increased either by increase of signal, or reduction of noise. The factors that influence noise are: (i) patient, scanner, and environment voxel volume, (ii) signal average, (iii) RF bandwidth and RF quality factor, (iv) magnetic field strength, (v) cross excitation and, (vi) acquisition and reconstruction algorithm.

Increase of signal can increase SNR and can be achieved as follows. Large voxel by increasing FOV or slice thickness or reducing phase encoding steps, that will improve SNR but reduces resolution. Decrease of TE, increase of TR or flip angle, spin echo sequence than GRE, and higher magnetic field gives large signal and increases SNR.

Reduction of noise can also improve SNR, that can be achieved as follows. Increasing the number of excitations, reducing the RF receiver bandwidth, reducing cross talks by having larger gaps between slice, reducing volume of tissue with surface coils (phased array type) and 3D imaging than 2D (lengthen scan time) are the noise reduction methods in MRI.

### **Scan Time**

Scan times control motion artifacts, and image quality, which is a function of TR and number of excitations. Shorter TR will decrease SNR, decrease number of slices and increases T1 weighting. Reduction of number of excitations will decrease SNR, but increase motion artifacts.

## **MRI ARTIFACTS**

MRI artifacts are present as positive or negative intensities that do not represent the imaged anatomy, which can limit the diagnostic potential. Knowledge about impact of acquisition protocols, etiology of artifact production will enhance the goal of achieving good diagnostic images. Artifacts may be machine dependent, patient related, and signal processing. It can be classified as (i) susceptibility artifacts, (ii) gradient field artifacts, (iii) motion artifacts, (iv) chemical shift artifacts, (v) wrap around artifacts, (vi) RF artifacts, (vii) K-space errors, (viii) ringing artifacts, and (ix) partial volume artifacts.

### **Susceptibility Artifacts**

Susceptibility artifacts are due to change in magnetic susceptibility of the tissues, which will distort the magnetic field, cause signal loss, due to rapid de-phasing ( $T2^*$ ) at the tissue-air interface, e.g. lungs, and sinuses. It is helpful in diagnosis, e.g. age of hemorrhage, since hemoglobin contain iron, which is ferromagnetic. This artifact is more in GRE than Spin echo sequence.

### **Gradient Field Artifacts**

Gradient field artifacts arise from gradient field nonuniformity or gradient failure. The field strength of the periphery is less than the ideal, which cause artificial compression of the anatomy, resulting in barrel distortion of image.

### **Motion Artifacts**

Motion artifacts are due to long scan time, cyclic heart motion or breathing, and blood flow. Motion artifacts cause blurring, reduce contrast, produce ghost images and appear in the phase encoding (slow event) direction. Motion artifact suppression technique such as cardiac and respiratory gating is used.

### **Chemical Shift Artifacts**

Chemical shift refers changes in resonance frequency and appear as displacement in the frequency encoding direction. Stronger the magnetic field, higher the chemical shift. In the case of water and

fat, water is placed few pixels higher in the gradient, e.g. fat around optic nerve, vertebral bodies.

### **Wrap Around Artifacts**

Wraparound artifacts are mismapping of anatomy that lies outside of the FOV but within the slice volume. The RF coil pick up signal from outside FOV and allot space in the matrix. Anatomy is displaced to the opposite side of the image, in the phase encoding direction. The causes are nonlinear gradients or under sampling of the frequencies in the returned signal. The remedial measures includes, circuits to suppress aliasing in FEG (low pass filter) direction, doubling the number of phase encoding steps, doubling the FOV in PEG, reducing the number of excitations, moving the ROI to the center or displaying only central half of the image and use of surface coil that matches FOV.

### **RF Artifacts**

Radiofrequency artifacts arise from surface coil's variation in uniformity due to RF attenuation, mismatching, sensitivity fall off with distance, resulting in shading and loss of image brightness. Stray RF signals (TV, radio, motor, fluoro light, computer) reaching the MRI antenna can also create artifacts.

The type of RF artifacts includes (i) center point artifact: due to RF amplifiers (two) imbalance, resulting in a bright spot in the centre of the image, (ii) zipper artifacts: due to narrow band RF noise, and (iii) herringbone artifacts: due to broad band noise (diffused and contrast reduced image).

Non-rectangular RF pulses received from adjacent slices excite and saturate protons in adjacent slices. It degrades the SNR on T2 weighted image and reduces image contrast on T1 weighted image. Hence, slice interleaving technique is used.

### **K-space Errors**

Error in K space encoding affect the reconstructed image, causes artificial superimposition of wave patterns across the FOV. This is called K space error artifact and the remedial measure is identification of bad pixels, and averaging the signals in adjacent pixels.

### **Ringing Artifacts**

Truncation or ringing artifacts appear as multiple, regularly spaced parallel bands of alternating bright and dark signal that slowly fades with distance. It is described as a diminishing hyper and hypointense signal oscillation from transition, known as Gibbs phenomenon. It occurs

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near sharp boundaries and high contrast interfaces, e.g. skull/brain interface, fat and muscle. This is caused by insufficient sampling of high frequencies and likely for smaller matrix sizes, in PEG direction. The remedy is increased matrix size or reduced FOV.

### **Partial Volume Artifacts**

Partial volume artifact arises from finite size of the voxel over the averaged signal and it results in loss of detail and spatial resolution. Use of smaller pixel size or smaller slice thickness will reduce partial volume artifact. However, it will reduce SNR, which can be maintained with longer imaging time.

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