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Evaluation of Image quality and Patient Radiation Dose in Chest X-ray Imaging

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Amin abdelhameed Amin

Abbreviations

a-Se amorphous selenium

a-Si amorphous silicon

CR computed radiography

CSI (Tl) cesium iodide doped with thallium

DR digital radiography

FFD focus and film distance

ICRU International Commission on Radiation Units and Measurements

ICRP International Commission on Radiation protection

MTF modulation transfer function

PACS picture archiving and communication system

TBCTA Tuberculosis Coalition for Technical Assistance

WHO World Health Organization

Units

A ampere

cm centimeter

kV kilovolt

mA milliampere

mAs milliampere second

mm millimeter

Glossary

Amorphous selenium (a-Se): A material used in a direct digital X-ray detector.

Amorphous silicon (a-Si): A material used in a direct digital X-ray detector.

Anode: A part of the inside the X-ray tube which generates the X-ray beams.

Artifacts: Marks on a radiograph that is foreign to the real image, such as scratches, fingerprints, static and etc.

Bucky: A device that is a part of the X-ray equipment unit which contains the moving grid system for reducing unnecessary scattered radiation. Bucky is a commonly used word of abbreviation of the Potter-Bucky moving grid system.

Cassette: A light-tight holder that contains a pair of intensifying screens and an X-ray film.

Cesium iodide thallium [CsI (Tl)]: The material used in an indirect digital X-ray detector.

Computed radiography (CR): Digital radiography technology which uses a phosphor-based plate.

Contrast: The difference between the light and dark areas of a radiograph.

Density: The degree of blackening of an X-ray film caused by X-ray exposure.

Exposure: The amount of radiation produced from the X-ray tube by pre-set factors such as kV, mA and seconds.

Grid: A device for absorbing unnecessary scattered radiation.

High-frequency inverter generator: A generator which produces a high frequency Output of AC voltage through the inverter from the rectified DC power input.

Modulation transfer function (MTF): The absolute value of spatial resolution which quantifies image sharpness.

Quality assessment: A method for measuring the efficacy of the chest radiograph used for improving imaging quality.

X-ray tube: A device that is a part of an X-ray equipment unit that contains an anode and a cathode for generating the X-ray beam.

X-ray tube current: The current flowing in the X-ray tube during an exposure. It is one of the radiographic exposure factors which controls the intensity of radiation and affects image density. The unit of X-ray tube current is the mA.

X-ray tube voltage: The electric voltage to the X-ray tube during an exposure. It is one of the radiographic exposure factors which controls the quality of the X-ray radiation and affects image contrast. The unit of X-ray tube voltage is the kV.

Conventional x-ray system = Photofluorography it's an x-ray system used only for chest imaging and use film /screen combinations.

Digital x-ray system =conventional upgraded using CCD technology it's an x-ray system used only for chest x-ray and use CCD technology to produce digital imaging

The body mass index (BMI) is defined as the individual's body mass divided by the square of his or her height. The formulae universally used in medicine produce a unit of measure of kg/m^2.

Body Mass index = weight (kg) / $(\text{height (m)})^2$.

KeV: A unit of energy useful with x-rays. It is equal to the energy supplied to an electron when accelerated through 1 kV.

Antiscatter grid: A thin structure made of alternating strips of lead and material transmissive to x-rays. Strips are oriented so that most scattered x-rays go through lead sections and are preferentially absorbed, while unscattered x-rays go through transmissive sections.

Dose: An estimate of the amount of ionizing radiation actually absorbed by a patient or attendant medical worker.

ALARA stands for As Low As Reasonably Achievable.

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Abstract

The aim of this study was to evaluate the image quality and Patient radiation dose in chest imaging using a Charge-Coupled Device (CCD) detector system with the image quality and doses delivered by a state-of-the-art conventional film-screen radiography system. The X-ray system used in this study was special system used only in Posteroanterior chest radiographs, it is a photofluorographic system " conventional screen-film system " and the same system upgraded to digital with CCD technology. Image quality was evaluated to ensure that the potential reduction in radiation dose did not result in decreased image quality. Two groups of fifty patients referred to the radiology department for routine posteroanterior chest radiography in National Center of Diseases Control (Tripoli-Libya). At random, images obtained with the film/screen system and CCD detector system. All patient groups were matched for body mass index. To measure effective dose which represents the risk of late radiation-induced effects, we used Fluke Victoreen 4000 M to measure Air Kerma then use PC-based Monte Carlo program for calculating patient's effective dose. Image quality of two systems was evaluated by three experienced radiologists. The CCD detector radiography system allowed a reduction in effective dose compared with the film-screen radiography. In addition, image quality produced by CCD camera detector radiography system was significantly better than the image quality produced by the film-screen radiography systems.

Arabic Summary

الملخص العربي

تعتبر الأشعة السينية من أهم طرق التصويرالطبى المعروفة التي استخدمت لأكثر من 100 سنة والتى تم تطويرها خلال هذة الفترة وقد ادى هذا التطور الى ظهور انواع مختلفة من التكنولوجيا المستخدمة فى التصوير بالاشعة السينية ، ان التطور الهائل فى مجال الحاسب الالى وتكنولوجيا الاتصالات وتخزين الملفات وارشفة الصور ادى الى ظهور الانظمة الرقمية فى جميع مجالات التصويرالطبى كما ادى الى ظهور التصويرالرقمى بالاشعة السينية .

الانظمة الرقمية التى بدأت باستخدام الواح تخزين من الفوسفور(CR) ثم تطورت الى ظهور الالواح المسطحة (Flat Panel Detector) التى تقوم بتحويل الاشعة الى صورة رقمية مباشرة وكذلك (CCD Detector) التى تقوم بتحويل صورة الاشعة الى صورة رقمية ايضا .

نتيجة لهذا التطور الكبير كان من الضرورى عمل تقييم لهذه التكنولوجيا وقياس مدى تاثيرها بالسلب او الايجاب على الصورة الطبية ومدى جودتها وتاثيرها على التشخيص مقارنة بافلام الاشعة و الجرعة الاشعاعية التى يتعرض لها المريض .

لذلك قامت العديد من الدراسات بعمل هذا التقييم وكانت معظم هذه الدراسات موجهة الى الواح التخزين الفوسفورية (CR) و الالواح المسطحة (Flat panel detector) وكانت النتيجة دائما ان التكنولوجيا الحديثة تؤدى الى تخفيض الجرعة الاشعاعية وتحسين جودة الصورة .

Arabic Summary

الهدف من هذه الرسالة هو عمل بحث لتقييم جودة الصورة والجرعة الاشعاعية التى يتعرض لها المريض اثناء تصوير الصدر على اساس استخدام التكنولوجيا التقليدية والتكنولوجيا الحديثة باستخدام (CCD Detector).

نظام الأشعة السينية المستخدمة في هذه الدراسة هو نظام خاص يستخدم فقط في الصور الاشعاعية الخاصة بالصدر " photofluorographic" وتم ايضا تطوير النظام ورفع مستواه لتكنولوجيا رقمية باستخدام CCD.

تم استخدام جهاز "Fluke Victoreen 4000 M" وذلك لقياس الجرعة الاشعاعية فى الهواء والتى يتعرض لها المريض ثم استخدام البرنامج الطبى "PCXMC" الخاص بحساب الجرعة الاشعاعية الفعالة ،كما تم قياس الجرعة الفعالة لمائة مريض من مرضى الشهائد الصحية فى المركز الوطنى للوقاية من الامراض فرع طرابلس وتسجيل هذه القيم فى جداول وتقييمها و عمل تقييم لصور الاشعه بواسطة ثلاثة من الاطباء العاملين فى مجال الاشعة.

ان النظام الذى تم تطويره ليعمل بالنظام الرقمى أدى الى إنخفاض الجرعة الاشعاعية التى يتعرض لها المريض مقارنة مع النظام التقليدى كما أدى الى رفع جودة الصورة وتحسين التشخيص الطبى من خلال الصورة الرقمية.

Introduction & Aim of the Work

1.1 General Introduction

Images of the human body are derived from the interaction of energy with human tissue. The energy can be in the form of radiation, magnetic or electric fields, or acoustic energy. The energy usually interacts at the molecular or atomic levels [1].

On Friday evening, 8 November 1895 Röntgen, Wilhelm Conrad (also sometimes spelled Roentgen) discovered a “new kind of ray” that penetrated matter. Röntgen professor of physics at Julius Maximilian University of Wurzburg, named the new kind of ray X-strahlen “X-rays” (“X” for unknown). [2]

X-rays are electromagnetic waves having energy in the general range of approximately one to several hundred kilo electron volts (**keV**). In medical X-ray imaging, the X-ray energy typically lies between 5 and 150 keV, with the energy adjusted to the anatomic thickness and the type of study being performed [3].

X-ray technology are divided into two main groups digital and conventional film-screen system, digital radiography is a form of X-ray imaging, where digital X-ray sensors are used instead of traditional photographic film.

X-rays are known to cause malignancies, skin damage and other side effects and they are thus potentially dangerous. Therefore, it is essential and in fact mandatory to reduce the radiation dose in diagnostic radiology as far as possible. This is also known as the ALARA (as low as reasonably achievable) principle. However, the dose is linked to image quality and the image quality may not be lowered so far that it jeopardizes the diagnostic outcome of a

radiographic procedure. The process of reaching this balance between dose and image quality is called optimization. Radiation dose will be measured as kerma area-product (KAP), entrance surface dose (ESD) and effective dose.

Because computer technology and storage capacity have developed rapidly during recent years, PACS (picture archiving and communication systems) have become more important, and the accurate implementation of a PACS depends on digital radiography techniques. Digital radiography systems offer an instant image display, a wide dynamic range, and digital images allow flexibility in processing and archiving, [6]

The first step in digital chest radiography was the use of storage phosphor plates. Introduced some 20 years ago [4], these computed radiography systems are widely used because of their compatibility with existing radiography equipment. However, conflicting results have been reported concerning comparisons of the image quality and radiation dose delivered by computed radiography systems with those of conventional film-screen radiography combinations.

Recently, full-field digital X-ray detector radiography systems based on cesium iodide and amorphous silicon or CCD become commercially available. These systems combine all advantages of digital radiography with higher quantum detection efficiency than is attainable with film screen [5].

1.2 History and Scope of X-ray in medical field

Until the beginning of the last century, the investigative tools available to clinical medicine amounted to the microscope, the thermometer, the knife and the stethoscope. Our understanding of the gross functions of the visually discernible organs of the human body evolved from centuries of careful anatomical studies of the dead and observations of the functional deficits imposed by injury and disease on the living. Until 1896 no means existed to examine or measure the hidden internal world of the living human body. [7]

One evening in December 1895, in an attempt to understand the glow produced in such a tube, he covered it with opaque paper. Then one of those events that trigger great minds to discovery happened: he had darkened his laboratory to better observe the glow produced in the tube, and in the dim light he noticed flashes of light on a barium-platinocyanide screen that happened to be near the apparatus. Because the tube (now covered with black paper) was obviously opaque to light emitted from the tube, he realized that the flashes on the screen must be due to emissions from the tube because they disappeared when the electric potential was disconnected.

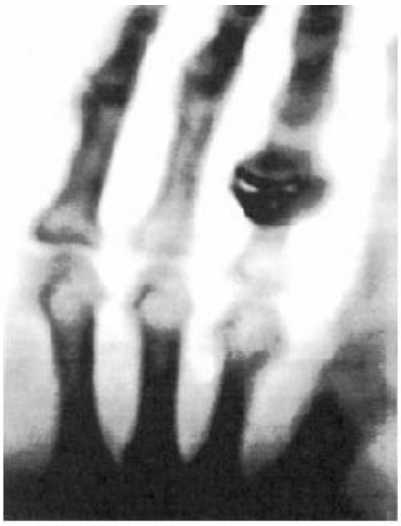


Figure [1-1] first X-ray image [8]

In 1901 Röntgen received the Nobel Prize for Physics, which was the first Nobel Prize in physics ever awarded. Unfortunately, Röntgen, his wife, and his laboratory workers all died prematurely of cancer. On January 18, 1896 an *X-ray machine* was formally displayed by H.L. Smith. Upon discovery in 1895, X-

Rays were advertised as the new scientific wonder and seized upon by entertainers. Circus patrons viewed their own skeletons and were given pictures of their own bony hands wearing silhouetted jewelry. While many people were fascinated by this discovery, others feared that it would allow strangers to look through doors and invade people's privacy.

In the 1940s and 50s, (real time) X-ray machines were used in stores to help sell footwear. These were known as fluoroscopes. However, as the harmful effects of X-ray radiation were properly considered, they finally fell out of use. Shoe-fitting use of the device was first banned by the state of Pennsylvania in 1957. (They were more a clever marketing tool to attract customers, rather than a fitting aid.) [2].

1937-The Fifth International Congress of Radiology accepts the roentgen as an international dosage unit for x- and γ-radiation and in 1940 first application of Angiography. At 1946—Advisory Committee on X-ray and Radium Protection reorganized to the National Committee on Radiation Protection (United States).[9]

1.3 Previous researches

Several studies have confirmed the excellent image quality and low dose provided by the digital radiography system. Some of these studies were listed.

Study was done by (Department of Diagnostic Radiology, Ruprecht-Karls-University of Heidelberg, and Radiological University Hospital. Germany) and publish at American Roentgen Ray Society, Received April 19, 2002; accepted after revision December 17, 2002. Under title (**Comparing Image Quality of Flat-Panel Chest Radiography with Storage Phosphor Radiography and Film-Screen Radiography) [**10].

Study was done by (Department of Medical Physics and Radiation Protection, Ghent University, Proeftuinstraat 86, Gent B-9000, Belgium. And Department of Radiology, Ghent University Hospital, De Pintelaan 185, Gent B-9000, and Belgium.) And publish at American Roentgen Ray Society, Received February 11, 2003; accepted after revision April 17, 2003. Under title (**Dose Reduction in Patients Undergoing Chest Imaging:** Digital Amorphous Silicon Flat-Panel Detector Radiography Versus Conventional Film-Screen Radiography and Phosphor-Based Computed Radiography) [11].

Study was done by (Department of Medical Physics and Radiation Protection, Ghent University, Proeftuinstraat 86, Gent B-9000, Belgium. And Department of Radiology, Ghent University Hospital, De Pintelaan 185, Gent B-9000, and Belgium.) And publish at American Roentgen Ray Society .Received March 7, 2005; accepted after revision July 20, 2005. Under title (**Image Quality and Radiation Dose on Digital Chest Imaging:** Comparison of Amorphous Silicon and Amorphous Selenium Flat-Panel Systems) [12].

Study was done by (Medical Physics Department and Accident and Emergency Department, S. Orsola-Malpighi Hospital, Bologna, Italy) and publishes at The British Journal of Radiology, Vol. 79 (2006). Under title (**Comparison of radiation doses to patients undergoing standard radiographic examinations with conventional screen–film radiography, computed radiography and direct digital Radiography)**[13].

Study was done by (Department of Biomedical Imaging, University of Malaya Medical Centre, Kuala Lumpur, Department of Diagnostic Imaging, Paediatric Institute, Kuala Lumpur Hospital, Kuala Lumpur, Malaysia)and publish at Biomedical Imaging and Intervention Journal, Received 3 February 2006; received in revised form 7June 2006; accepted 19 June 2006, under title (**Evaluation of radiation dose and image quality following changes to tube potential (kVp) in conventional paediatric chest radiography**) [14].

Study was done by (Departments of Radiology and Medical Statistics, Leiden University Medical Centre, Albinusdreef 2, 2333 ZA Leiden, The Netherlands) and publish at The British Journal of Radiology,Received 29 September 2006 Revised 7 February 2007 Accepted 22 February 2007. Under title **(Dose reduction in digital chest radiography and perceived image quality)** [15].

Study was done by (Zahedan Health Promotion Research Center, Zahedan University of Medical Sciences, Zahedan, Fasa University of Medical Sciences, Fasa ,National Radiation Protection Department, Tehran, Iran) and publish at Indian Journal of Science and Technology, Vol. 4 Nov 2011, under title (**Evaluation of image quality and patient dose in conventional**

radiography examinations in radiology centers in Sistan and Baluchestan, Iran and comparing with that of international guidelines levels)[16].

1.4 Objective

The aim of this research project was to evaluate the patient radiation dose and image quality delivered by conventional film/screen upgraded to digital system using CCD technology with those of state-of-the-art conventional film-screen in clinical chest imaging. Image quality performance was evaluated to ensure that the potential dose reduction was not detrimental to quality.

1.5 Subjects and Methods

Material and reference was collected and characteristic of the two systems were studied. Two groups of 50 patients each were examined (chest radiography) using the conventional film-screen upgraded to digital radiography and film-screen system. All patient groups were matched for body mass index. To measure the entrance skin dose, measuring instruments were used to every patient. The calculation of the effective doses, which represents the risk of late radiation-induced effects, and records all reading in tables and compare the results. Image quality of all two systems was evaluated by experienced radiologists.

1.6 Research Structures

This thesis consist from four chapters, chapter two focuses on history of X-ray , X-ray machine digital machine and film /screen component and describes the difference between two systems. Chapter three focuses on radiation, radiation units, effects and medical image quality and its properties. Chapter four focuses on experimental work, methodology of work, how the result was collected and discussion.

Film /screen X-ray systems

&

Film /screen upgraded to digital x-ray system

2.1 Introduction

X-ray imaging is a well-known imaging modality that has been used for over 100 years, since Roentgen discovered X-rays based on his observations of fluorescence. His initial results were published in 1895, and reports of diagnoses of identified fractures shortly followed. A year later, equipment manufacturers started selling X-ray equipment. Today, X-ray and its three-dimensional (3D) extension, computed tomography (CT), are used commonly in medical diagnosis. [17]

X-rays are high-energy photons. Their generation creates incoherent beams that experience insignificant scatter when passing through various media. As a result, X-ray imaging is based on through transmission and analysis of the resulting X-ray absorption data. Typically, X-rays are detected through a combination of a phosphor screen and a light-sensitive film, the current system, which has been used for mammography and radiography for many years, provides a good-quality analog image that is not compatible with digital storage and transmission requirements of the modern digital era. A slight variation of this common technique is used in fluoroscopy where image intensifier is used as transition stage to supply signals to CMOS cameras producing an analog image directly on a TV screen. Multiple conversions steps in this case from X-rays to electrons to light to camera display lead to poor image quality. [18]

An alternative to the conventional detection technique uses a digital detector that converts X-ray photons directly into an electrical signal of digital nature. Digital image leads to lower storage cost and ease of electronic transmission in a future of Healthcare era.

There are some important differences in characterization of film-based imaging and digital imaging. In this chapter the focus will be on the two technologies because the replacement of the film-based imaging the same way that digital cameras have displaced analog films in consumer cameras. In digital imaging we use terms such as brightness, dynamic range, linearity, or signal-to-noise ratio instead of density, latitude, film speed, or image sharpness, the terms associated with film-based technology.

2.2 Production of X-ray

X-rays are produced when highly energetic electrons interact with matter and convert their kinetic energy into electromagnetic radiation. A device that accomplishes such a task consists of an electron source, an evacuated path for electron acceleration, a target electrode, and an external energy source to accelerate the electrons. Specifically, the X-ray tube insert contains the electron source and target within an evacuated glass or metal envelope, the tube housing provides shielding and a coolant oil bath for the tube insert, collimators define the X-ray field; and the generators the energy source that supplies the voltage to accelerate the electrons. The generator also permits control of the X-ray output through the selection of voltage, current, and exposure time. These components work in concert to create a beam of X-ray photons of well-defined intensity, penetrability, and spatial distribution [19].

2.2.1 Bremsstrahlung Spectrum

The conversion of electron kinetic energy into electromagnetic radiation produces x-rays. A simplified diagram of an X-ray tube was shown in Fig [2-1] illustrates the minimum components. A large voltage is applied between two

electrodes (the cathode and the anode) in an evacuated envelope. The cathode is negatively charged and is the source of electrons; the anode is positively charged and is the target of electrons. As electrons travel from the cathode to the anode, they are accelerated by the electrical potential difference between these electrodes and attain kinetic energy. The electric potential difference, also called the voltage.

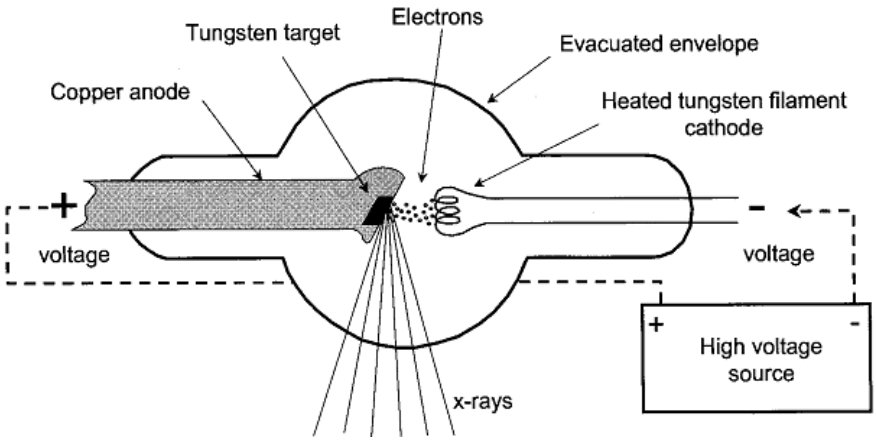


Figure [2-1] Illustrates the minimum components required for X-ray production

On impact with the target, the kinetic energy of the electrons is converted to other forms of energy. The vast majority of interactions produce unwanted heat by small collisional energy exchanges with electrons in the target. This intense heating limits the number of X-ray photons that can be produced in a given time without destroying the target. Occasionally (about 0.5% of the time), an electron comes within the proximity of a positively charged nucleus in the target electrode. Attraction forces attract and decelerate the electron, causing a significant loss of kinetic energy and a change in the electron's trajectory as shown in Fig [2-2]. An X-ray photon with energy equal to the kinetic energy lost by the electron is produced (conservation of energy). This radiation is termed bremsstrahlung, a

German word meaning "braking radiation"[20].

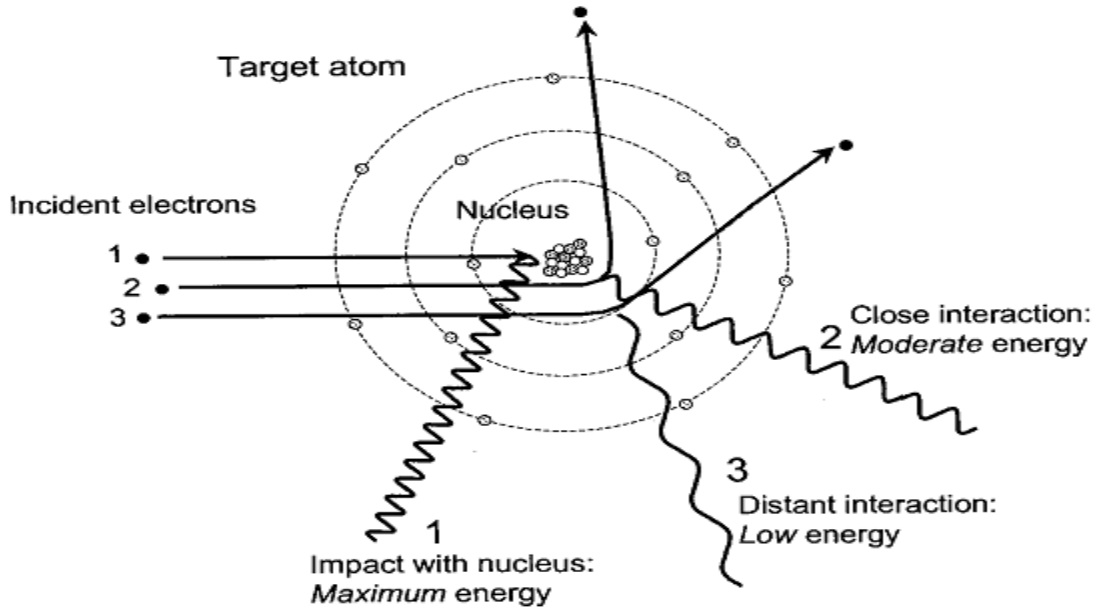


Figure **[2-2]** Bremsstrahlung radiation arises from energetic electron interactions with an atomic nucleus of the target material.

2.2.2 Characteristic X-Ray

Each electron in the target atom has a binding energy that depends on the shell in which it resides. Closest to the nucleus are two electrons in the K shell, which has the highest binding energy. The L shell, with eight electrons, has the next highest binding energy, and so when the energy of an electron incident on the target exceeds the binding energy of an electron of a target atom, it is energetically possible for a collisional interaction to eject the electron and ionize the atom. The unfilled shell is energetically unstable, and an outer shell electron with less binding

energy will fill the vacancy. As this electron transitions to a lower energy state, the excess energy can be released as a characteristic X-ray photon with energy equal to the difference between the binding energies of the electron shells [20] as shown in Fig [2-3].

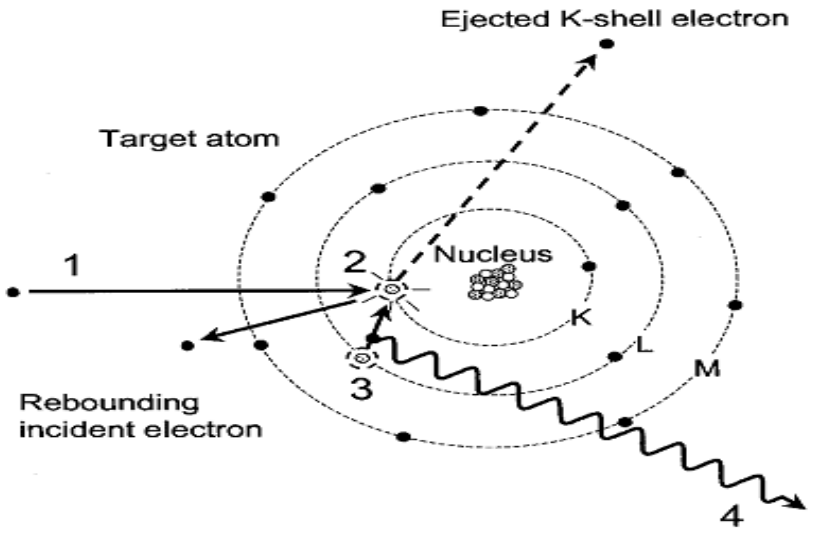


Figure [2-3] shows the characteristic X-ray Generation methods.

2.3 Conventional X-ray System

Conventional X-ray radiography produces images of anatomy that are shadowgrams based on X-ray absorption. The x-rays are produced in a region that is nearly a point source and then are directed on the anatomy to be imaged. The X-rays emerging from the anatomy are detected to form a two-dimensional image, where each point in the image has a brightness related to the intensity of the X-rays at that point. Image production relies on the fact that significant numbers of X-rays penetrate through the anatomy and that different parts of the anatomy absorb different amounts of X-rays. In cases where the anatomy of interest does

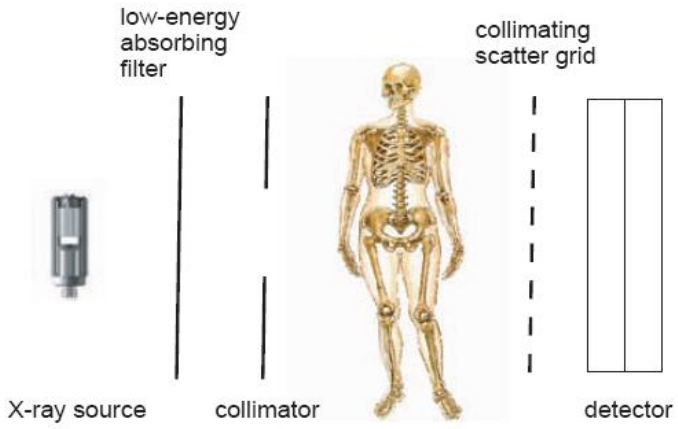
not absorb x-rays differently from surrounding regions, contrast may be increased by introducing strong X-ray absorbers. For example, barium is often using to image the gastrointestinal tract. [21]

X-rays are electromagnetic waves (like light) having an energy in the general range of approximately one to several hundred kiloelectronvolts (keV). In medical X-ray imaging, the X-ray energy typically lies between 5 and 150 keV, with the energy adjusted to the anatomic thickness and the type of study being performed. X-rays striking an object may either pass through unaffected or may undergo an interaction. These interactions usually involve either the photoelectric effect (where the X-ray is absorbed) or scattering (where the X-ray is deflected to the side with a loss of some energy). X-rays that have been scattered may undergo deflection through a small angle and still reach the image detector; in this case they reduce image contrast and thus degrade the image. This degradation can be reduced by the use of an air gap between the anatomy and the image receptor or by use of an antiscatter grid [22].

2.3.1 System Component

The complete radiographic imaging system was illustrated schematically in Fig [2.4] it consists of the following elements;

Figure [2.4 a] shows the X-ray system parts.



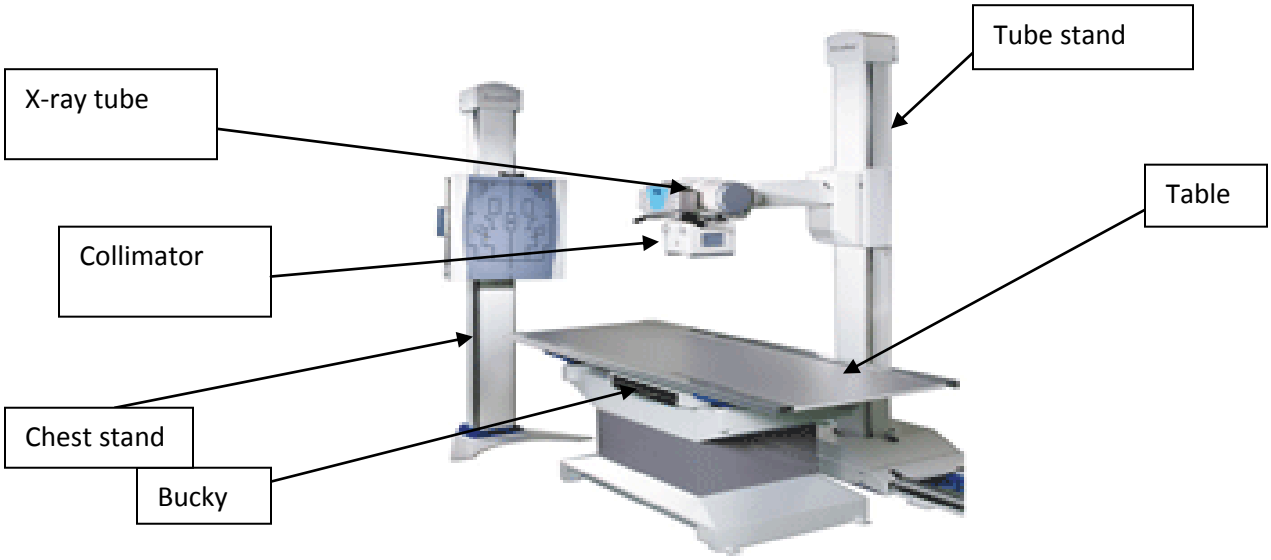


Figure [2.4 b] shows the X-ray system parts.

X-ray Tubes

The modern X-ray tube can produce very large beam intensity, but it uses the same principle that Roentgen employed in 1895. The kinetic energy of accelerated electrons is converted into heat and a broad spectrum of electromagnetic radiation that includes X-rays. The same principle is used to produce visible light photons on the screens of the older TV sets, desktop computer monitors and oscilloscopes. They are basically cathode ray tubes shown schematically in Fig [2.5][23].

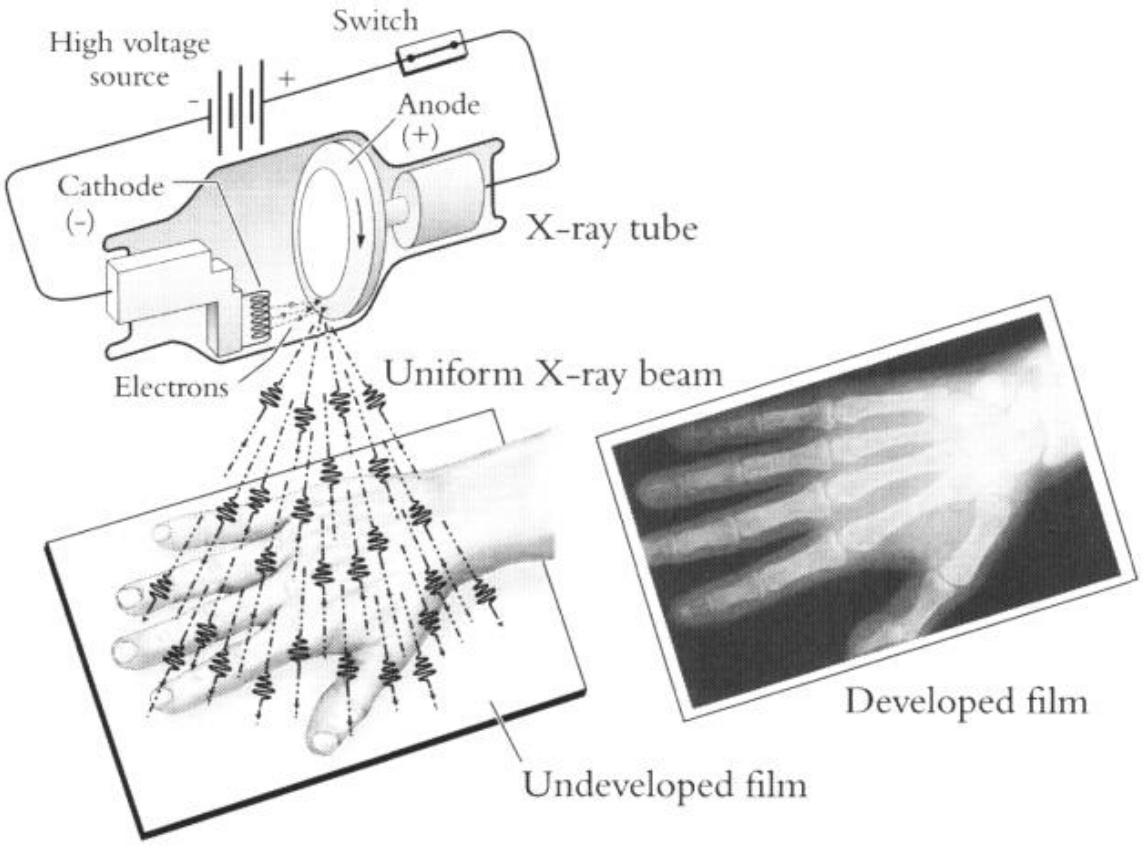


Figure [2.5] X-ray Tube

In a typical medical X-ray tube, the accelerating voltage or HT is set at 30–150 kV so that electrons acquire a kinetic energy of 30–150 keV just before they hit the target anode. Typically, electron currents will be in the range 10–100 milliamps and thus the electrical power consumed by the tube will be 100 X 10^3 V X 0.1 A = 10 kW. The electrically-charged electrons interact very strongly with the target atoms and are brought to rest within a few mm of its surface. By far the largest part of the electron beam energy is converted into heat and this is deposited within a few mm of the surface of the target. Only about 1% of the available electron energy is actually converted into useful X-ray radiation, which escapes

from the tube. Without special precautions, the electron beam would rapidly melt the surface of the target, eventually drilling a hole straight through the metal. Thus, the most important problem for the tube designer is to cope with the 10 kW of heat energy that the electron beam deposits in a very small volume of the anode.

The heat problem is dealt with in four ways. The electron beam is shaped into a long thin rectangle oriented radially with respect to the anode disc. The anode is made to rotate at high speed (2000–3000 RPM) underneath the beam. Both of these ensure that the beam energy is spread over as large an area of target anode as possible. Even then the anode will get very hot. Very high melting point metals such as tungsten and ruthenium are used as targets. Finally, many modern tubes are pulsed and the radiographer is only allowed to take a certain number of exposures before a heat sensor detects the raised temperature of the anode and shuts the tube down until the anode has cooled off sufficiently to start again. Cooling of the anode target can be accomplished by circulating water or oil but many modern tubes just rely on radiative cooling. [7, 24].

An aluminum filter, often complemented by a copper filter. This filter removes low-energy photons, thus increasing the mean energy of the photon beam. Low-energy photons deliver doses to the patient but are useless for the imaging process because they do not have enough energy to travel through the patient and never reach the detector. Because low-energy photons are called soft radiation and high-energy photons hard radiation, this removal of low-energy photons from the beam is called beam hardening. Collimator to limit the patient area to be irradiated .This is a collimator that absorbs scatter photons. It stops photons with large incidence angle, whereas photons with small incidence angle can pass right

through the grid. The grid can be made of lead, for example. Detector can be a screen–film combination in which a film is sandwiched between two screens. [25]

Photon Detectors

The problems of X-ray detection bring together medical science and particle / nuclear physics. For the particle physicist, the problem is to detect and localize the results of rare fleeting complex particle interactions. The medical physicist has a similar problem of detecting and localizing enough photons to produce an image, but at the same time, minimizing the radiation dose administered to the patient. In both cases the aim is to make the best possible use of the photon flux available. Both communities are concerned with the efficiency of a device and its spatial resolution.

The majority of X-ray images are still recorded on photographic film. Digital radiography and X-ray CT both have to make use of electronic detectors that produce numbers which are stored in a computer and then subsequently manipulated to create the required two-dimensional image. Gamma imaging similarly depends on the efficient electronic detection of photons. J-ray energies are between two and ten times higher than X-rays but the same physical principles and many of the methods used for X-rays are also employed to detect and count J-rays.

Detection requires that an X-ray hits an atom or atoms of the detector and leaves a permanent, measurable trace of its passage. In order to be able to localize the point of impact of an X-ray, most detectors have to be thin and thus many photons simply pass straight through the detector without leaving any trace

whatsoever. The term quantum efficiency describes the number of incident photons required to produce a measurable trace of an impact in the detector. [7]

2.3.2 X-ray film / screen

The modern screen-film detector system used for general radiography consists of a cassette, one or two intensifying screens, and a sheet of film. The film itself is a sheet of thin plastic with a photosensitive emulsion coated onto one or both sides. The cassette has several functions. Most importantly, the inside of the cassette is light tight to keep ambient light from exposing the film .after a new sheet of film has been placed in the opened cassette (in a darkroom or other dark environment), the two halves of the cassette as shown in Fig. [2-6] are closed. During the closing process, the screens make contact with the film, at first near the cassette hinge; then, as the cassette continues to close, air is forced out of the space between the screens and the film the screens are permanently mounted in the cassette on layers of compressible foam. The foam produces force, which holds the screens tightly against the sheet of film. Physical contact between the screens and the film is necessary to avoid artifacts on the radiographic image. By avoiding air trapping and by applying pressure on the screens, the cassette maintains good screen-film contact, which is essential for good image quality. [26]

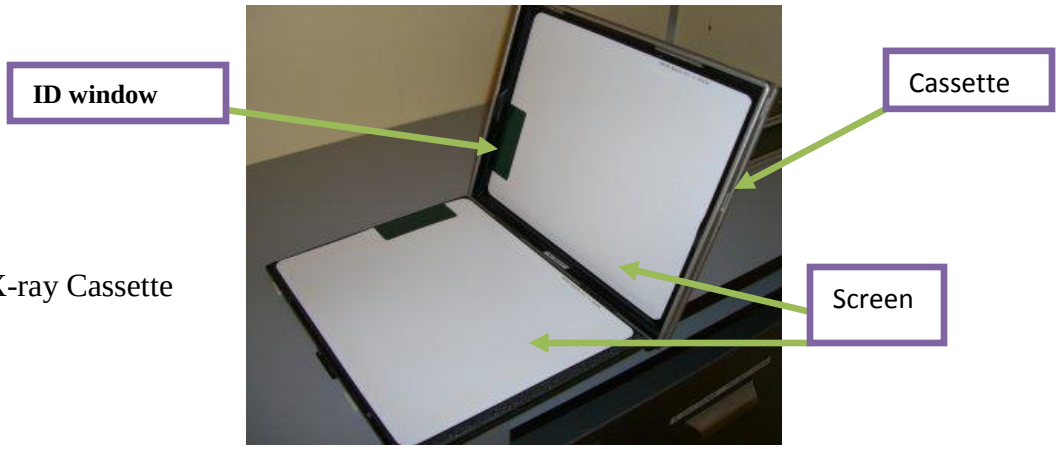


Figure [2.6] X-ray Cassette

2.4 Conventional upgraded to digital

2.4.1 System component

Conventional upgraded to digital is the same component as Conventional system except image detector in digital system is digital detector. Digital image receptors are gradually replacing screen-film cassettes as radiology departments convert to an all-digital technology. Radiography is really the last modality to make the transition to digital acquisition. The reasons for this are clear: Screen film is a tried and true detector system that produces excellent image quality under most circumstances [27, 28], and therefore the motivation for change is low. In addition, the large field of view and high spatial resolution of radiography require digital radiographic images to contain large amounts of digital data. For example, a typical high-resolution digital chest radiograph ranges from 4 megabytes (MB). After upgrading system we can use the latest technology in radiology known as the Picture Archival and Communication System (PACS) is a system for storage of images and transferring images between computers in different facilities through networks. This system consists of devices to produce and store digital images electronically, workstations to view and interpret images, and a network linking computers from different sites. Appropriate PACS software allows the interpreter to manipulate images as needed, at his own location, by retrieving images from other locations via PACS. The introduction of PACS can virtually eliminate the use of X-ray films or Polaroid films for routine interpretation of images, making the imaging centers filmless. [29]

2.4.2 Image detectors

To produce an image from the attenuated X-ray beam, the X-rays need to be captured and converted to image information. Some detectors for digital radiography are relatively recent developments. Conventional X-ray can be upgraded to digital using different technology, computed radiography (CR), Charged-coupled devices (CCD), Flat panel detector [30].

Computed radiography

Computed radiography (CR) is a marketing term for photostimulable phosphor detector (PSP) systems [17]. Phosphors used in screen-film radiography, when struck by an X-ray beam. When X-rays are absorbed by photostimulable phosphors, some light is also promptly emitted, but much of the absorbed X-ray energy is trapped in the PSP screen and can be read out later. For this reason, PSP screens are also called storage phosphors or imaging plates. CR was introduced in the 1970s, saw increasing use in the late 1980s [31], and was in wide use at the turn of the century as many departments installed PACS, often in concert with the development of the electronic medical record.

CR imaging plates are made of BaFBr and BaFI. Because of this mixture, the material is often just called barium fluorohalide. A CR plate is a flexible screen that is enclosed in a cassette similar to a screen-film cassette. One imaging plate is used for each exposure. The imaging plate is exposed in a procedure identical to screen-film radiography, and the CR cassette is then brought to a CR reader unit. The cassette is placed in the readout unit, and several processing steps then take place [32]:

1. The cassette is moved into the reader unit and the imaging plate is mechanically removed from the cassette.
2. The imaging plate is translated across a moving stage and scanned by a laser beam
3. The laser light stimulates the emission of trapped energy in the imaging plate, and visible light is released from the plate.
4. The light released from the plate is collected by a fiber optic light guide an strikes a photomultiplier tube (PMT), where it produces an electronic signal.
5. The electronic signal is digitized and stored.
6. The plate is then exposed to bright white light to erase any residual trapped energy.
7. The imaging plate is then returned to the cassette and is ready for reuse.

The digital image that is generated by the CR reader is stored temporarily on a local hard disk. Many CR systems are joined ("docked") directly to laser printers that make film hard copies of the digital images. CR systems often serve as entry points into a PACS, and in such cases the digital radiographic image is sent to the PACS system for interpretation by the radiologist and long-term archiving. The imaging plate is a completely analog device, but it is read out by analog and digital electronic techniques, as shown in Fig [2.7]. The imaging plate is translated along the readout stage in the vertical direction (the y direction), and a scanning laser beam interrogates the plate horizontally (the x direction). The laser is scanned with the use of a rotating multifaceted mirror. As the red laser light (approximately

700 nm) strikes the imaging phosphor at a location (x, y), the trapped energy from the X-ray exposure at that location is released from the imaging plate.

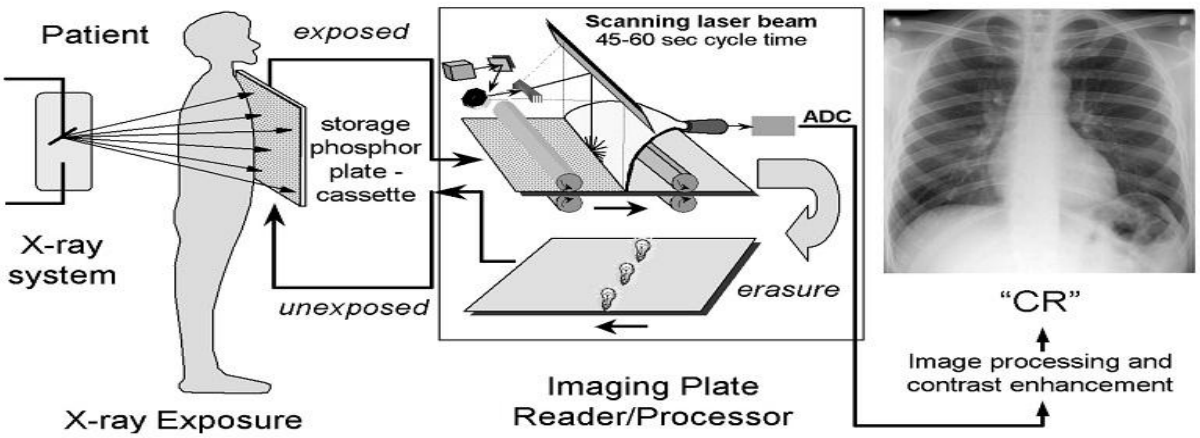


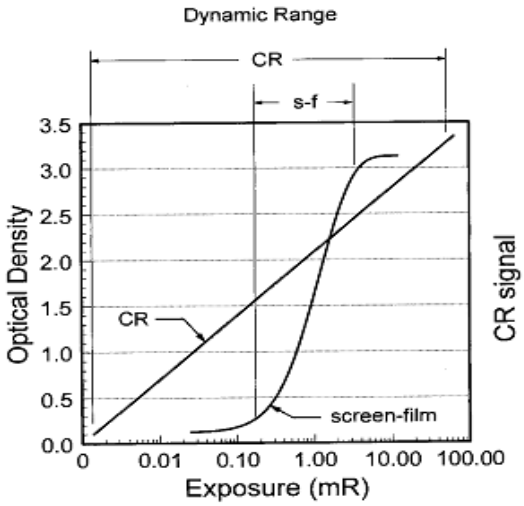
Figure [2.7] Schematic of a CR imaging system, screen, and scanner.

A fraction of the emitted light travels through the fiber optic light guide and reaches a PMT .The electronic signal that is produced by the PMT is digitized and stored in memory. Therefore, for every spatial location (x, y), a corresponding gray scale value is determined, and this is how the digital image I(x, y) is produced in a CR reader.

CR systems operate with a workflow pattern very similar to screen-film radiography. CR cassettes are exposed just like screen-film cassettes, and they are then brought to a reader unit, just as film-screen cassettes are brought to the film processor. The similarity in handling of CR and screen-film cassettes contributed to the initial success of CR. One of the advantages that CR over screen-film radiography is its much larger dynamic range Fig [2.8]; in other words, the exposure latitude with CR is much wider than with screen-film systems. Consequently, retakes due to overexposure or underexposure are rare with CR. Because of the difficulty of obtaining precise exposures in the portable

radiographic setting, where phototiming usually is not available, CR is often used for portable examinations. Although CR is capable of producing images with proper gray scale at high and low exposure levels, quality assurance efforts should still be in place to ensure proper exposure levels. CR images exposed to low exposure levels, while maintaining proper gray scale in the image, have higher levels of X-ray quantum noise ("radiographic mottle"). CR images produced at high exposures have low quantum noise but result in higher X-ray dose to the patient.Computed tomography (CR) system consist from CR Reader, Computer, dry printer as shown in fig (2.9). [33]

Figure [2.8] the exposure range (film latitude) in which screen-film cassettes produces usable optical densities is illustrated..



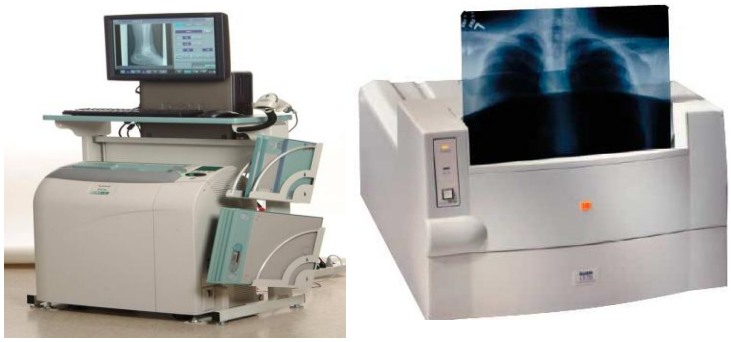


Figure [2-9] Computed Radiography (CR) system

Charged-coupled devices (used in X-ray machine in experimental part)

Charged-coupled devices (CCDs) form images from visible light. CCD detectors as shown in Fig [2.10] are used in most modern video cameras and in digital cameras. The principal feature of CCD detectors is that the CCD chip itself is an integrated circuit made of crystalline silicon, similar to the central processing unit of a computer. A CCD chip has discrete pixel electronics etched into its surface; for example, a 2.5 X 2.5 cm CCD chip may have 1024 X 1024 or 2048 X 2048 pixels on its surface. Larger chips spanning 8 X 8 cm have been produced. The silicon surface of a CCD chip is photosensitive- as visible light falls on each pixel, electrons are liberated and build up in the pixel. More electrons are produced in pixels that receive greater light intensity. The electrons are kept in each pixel because there are electronic barriers (voltage) on each side of the pixel during exposure. Once the CCD chip has been exposed, the electronic charge that resides in each pixel is read out [34].

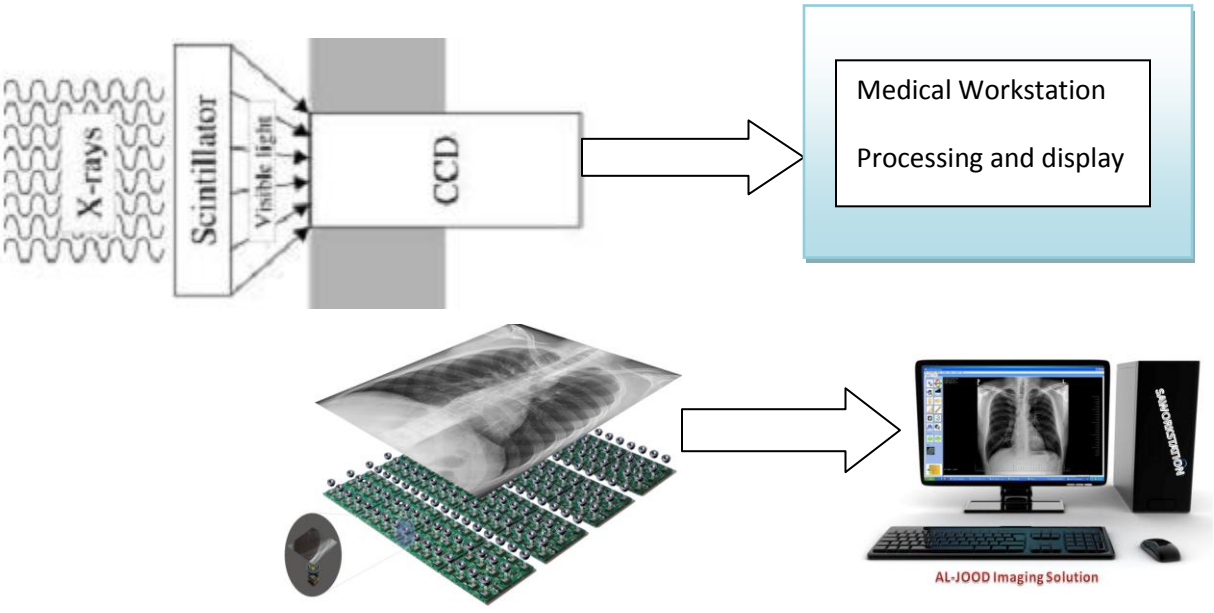


Figure [2.10 a] CCD detector

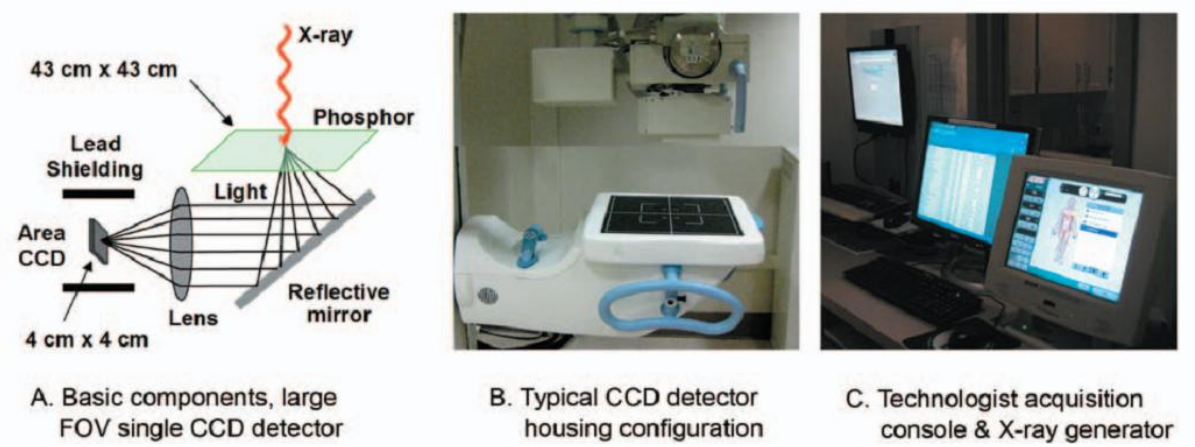


Figure [2.10 b] CCD detector

Flat panel detector systems

Flat panel detector systems make use of technology similar to that used in a laptop computer display, and much of this has to do with wiring the huge number of individual display elements. Instead of producing individual electrical connections to each one of the elements in a flat panel display, a series of horizontal and vertical electrical lines are used which, when combined with appropriate readout logic, can address each individual display element. With this approach only 2,000 connections between the imaging plate and the readout electronics are required for a 1,000 X 1,000 display, instead of 1,000,000 individual connections. For a flat panel display, the wiring is used to send signals from the computer graphics card to each display element, whereas in a detector the wiring is used to measure the signal generated in each detector element [34].

Indirect Detection Flat Panel Systems

Indirect flat panel Fig [2.11] detectors are sensitive to visible light and an X-ray intensifying screen is used to convert incident X-rays to light, which is then detected by the flat panel detector. The term "indirect" comes from the fact that x-rays are absorbed in the screen, and the absorbed X-ray energy is then relayed to the photo detector by visible light photons. This indirect detection strategy is analogous to a screen-film system, except that the electronic sensor replaces the light-sensitive film emulsion. Dual-emulsion film is thin and X-rays penetrate it easily; therefore, in a screen-film cassette, the film is sandwiched between screens to reduce the average light propagation path length and improve spatial resolution. Flat panels are thicker than film and do not transmit X-rays well; consequently, a sandwiched design is not possible. Instead, the intensifying screen is layered on the front surface of the flat panel array. This means that the light emanating from the *back* of the intensifying screen strikes the flat panel and much of the light that is released in the screen has to propagate relatively large distances through the screen, which results in more blurring. To improve this situation, most flat panel detector systems for general radiography use CsI screens instead of GdzOz5. CsI is grown in columnar crystals and the columns act as light pipes to reduce the lateral spread of light [19, 35].

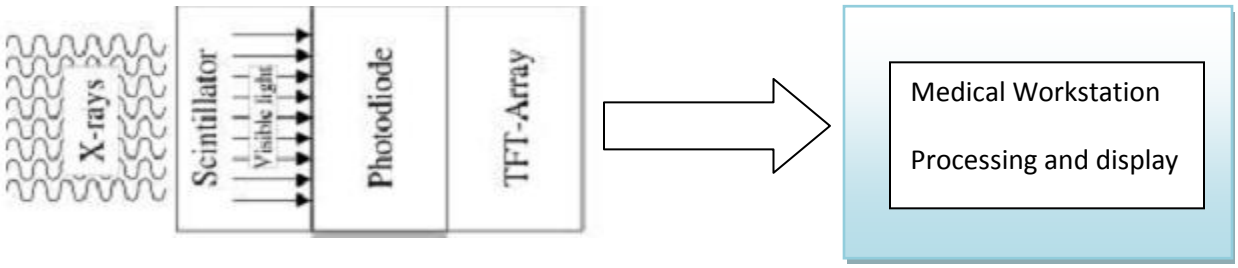


Figure [2.11 a] Indirect Detection Flat Panel Systems

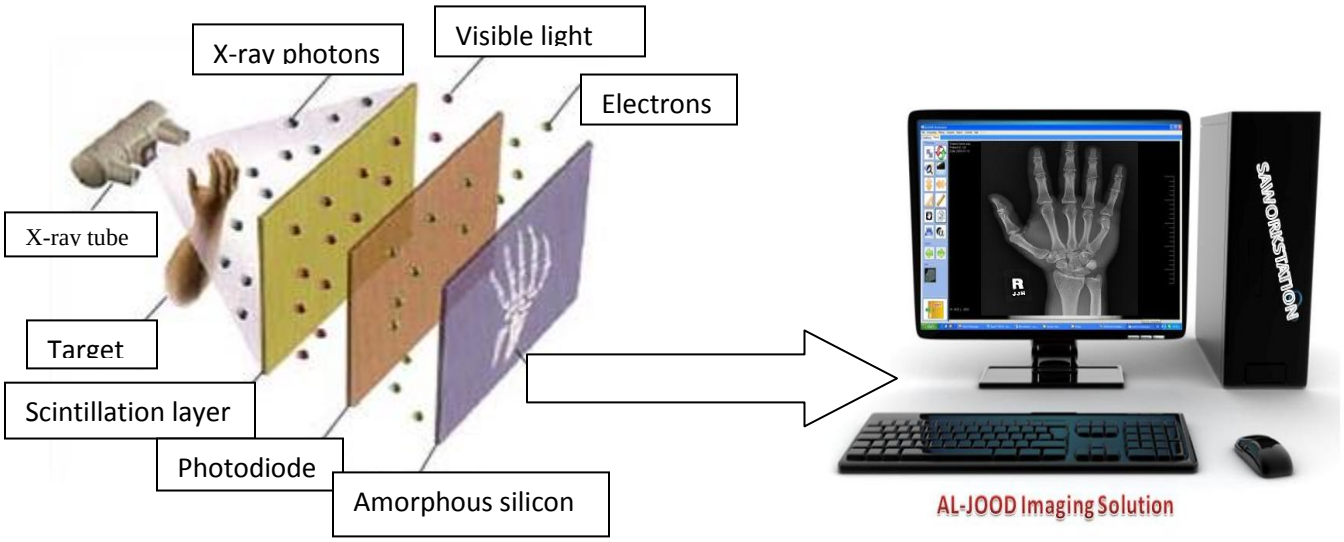


Figure [2.11 b] Indirect Detection Flat Panel Systems

Direct Detection Flat Panel Systems

Direct flat panel detectors as shown in Fig [2.12] are made from a layer of photo conductor material on top of a TFT array. These photoconductor materials have many of the same properties as silicon, except they have higher atomic numbers. Selenium is commonly used as the photoconductor. The TFT arrays of these direct detection systems make use of the same line-and-gate readout logic, as described for indirect detection systems. With direct detectors, the electrons released in the detector layer from X-ray interactions are used to form the image directly. Light photons from a Scintillator are not used. A negative voltage is applied to a thin metallic layer (electrode) on the front surface of the detector, and therefore the detector elements are held positive in relation to the top electrode. During X-ray exposure, X-ray interactions in the selenium layer liberate electrons that migrate through the selenium matrix under the influence of the applied electric

field and are collected on the detector elements. After exposure, the detector elements are read out as described earlier for indirect systems [19].

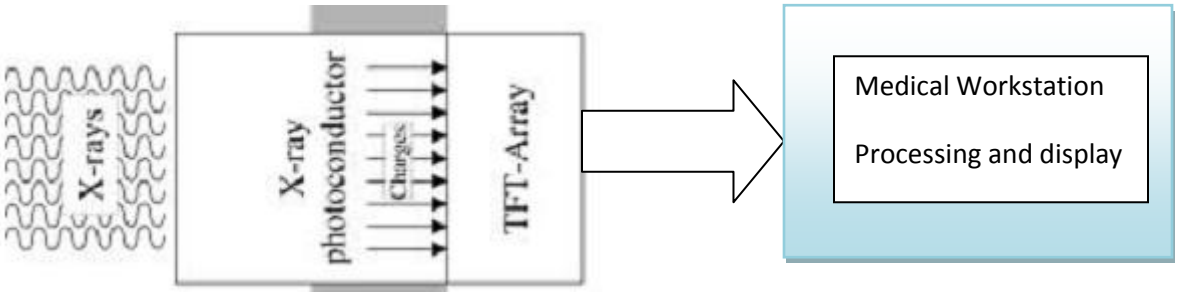


Figure [2.12] direct flat panel detectors

2.5 Advantages and Disadvantages for Conventional and Digital X-ray

2.5.1 Advantages of digital system

- Digital imaging can achieve higher spatial and contrast resolution and dose efficiency.
- Digital images can be adjusted electronically at a workstation to optimize the quality of X-ray of the desired anatomy.
- Digital radiography can achieve rapid processing and fewer repeat examinations by digital processing technology. And it can increase room use and reduce the cost through curtailment of darkroom and film processing procedures.
- Digital radiography system can efficiently archive and retrieve massive amounts of imaging data.
- Digital imaging data can be transmitted onto a workstation monitor and printed on film, as well as for electronic storage.
- Digital images can be sent by electronic media such as email or mobile phones, to a remote reader for diagnosis Pacs and teleradiology [36, 37].

2.5.2 Disadvantages of digital X-ray imaging system

- Costly initial investment
- Advanced maintenance technology and costly running expenses
- Vulnerable to unreliable electric power supply

2.5.3 Advantages of conventional X-ray

- Low cost initial investment compared with digital
- Easy maintenance and not costly running expenses
- invulnerable to unreliable electric power supply

2.5.4 Disadvantages of conventional X-ray

- Fixed spatial and contrast resolution
- Costly to archive films and patient history
- High patient dose

X-ray physics

&

Radiography image quality

3.1 Introduction

The benefits of the penetrating power of X-rays were obvious from the moment of their discovery in 1895[39], and medicine was quick to develop X-rays into the most important diagnostic tool of the last century. Acute biological effects, similar to sunburn, of direct exposure to high levels of X-ray radiation also became apparent rather quickly to both researchers and medical practitioners. The International Commission on Radiological Protection (ICRP) was set up in 1928 to evaluate the risks to man of all exposures to ionizing radiation and to set limits on maximum permissible levels of exposure. The exposure limits decreed by this and other similar bodies have dropped continuously over the past eighty years, as the more insidious effects of exposure to radiation have been revealed both by basic biological research and statistical studies. The most important of the latter arose from the long-term effects on the survivors of the Hiroshima and Nagasaki atomic explosions at the end of the Second World War.

The biological effects of exposure to ionizing radiation can be divided into two main regimes that depend on the level of exposure and the time interval between the radiation insult and the appearance of symptoms. At the very high doses received by early radiation workers and many of the Japanese atomic bomb victims, death and/or acute biological effects appear or reappear in seconds to weeks. At very much lower doses other effects such as genetic damage, life shortening and carcinogenesis only appear after many years or decades. At whole body doses in excess of 2–5 Sv, deterministic link between the level of exposure and a spectrum of illnesses (radiation sickness). There is a relatively simple relationship between the severity of an illness, its prognosis

without medical intervention and dose level. In addition there appears to be a fairly well-defined exposure threshold below which these symptoms never appear. At whole body doses in excess of 20 Sv, death is almost always the outcome in a matter of weeks to months. The acute effects of high radiation dose are not a matter for concern for the general public since, barring nuclear accident or nuclear war, the public does not come into contact with sufficiently powerful sources of radiation. They are potentially a matter of extreme concern for many workers in both the medical and industrial areas and radiation protection measures are universally in place to prevent accidental exposure to high levels of radiation [40].

The beneficial use of ionizing radiation, particularly in cancer therapy, necessarily exposes parts of the patient to lethal doses of radiation. Side effects in healthy tissue are an inevitable consequence. Of more concern, both to the general public and to radiologists, are the long term effects of an accumulated exposure to very low levels of radiation that are routinely used in, for example, dentistry. This is an extremely difficult problem. It is now assumed, although it took fifty years to demonstrate and become accepted, that there is no lower exposure threshold for long-term effects such as carcinogenesis. Rather there is a very small but calculable chance of even a single, trivial X-ray examination producing a fatal cancer sometime in the patient's lifetime. The International Commission on Radiological Protection (ICRP) has, in over fifty reports and publications, stressed that all exposure to radiation should be justified and guided by the three basic tenets of radiological protection. First, that the benefits of any radiation examination or procedure should outweigh the likely detriment. Second, that in any procedure the ALARA principle should be adopted. ALARA

stands for ‘As Low as Reasonably Achievable’. Finally only in exceptional circumstances should the dose limits set by the ICRP ever be exceeded [41].

Although it is generally accepted that the benefits of ionizing radiation in medicine have consistently outweighed long-term harm, there is continuing pressure to further reduce medical exposure and a developing culture of tighter voluntary control.

The purpose of medical imaging is to reveal and record the structural or functional state of the body. Mostly we want to see what is going on inside the body – to check that all is well, or to find out why all is not well. Sometimes we want a record of the current state of the body to be referred to at some future time to monitor the progress, or absence of progress, of a disease or a treatment. [42]

3.2 Radiation

The terms ‘dose’ and ‘exposure’ are widely used but often misunderstood. ‘Doses’ maybe measured for particular tissues or organs (e.g. skin, eye, bone marrow) or for the whole body, while ‘exposure’ usually refers to equipment settings (time, mA, kV). A commonly used measure of dose in surveys is ‘entrance dose’, measured in milligrays (mGy).This has an advantage of being fairly easily measured by placing dosemeters on the patient’s skin. [43, 44, 48]

3.2.1 International SI Units of Exposure and Dose

The term radiation dose, or simply dose, is defined carefully in terms of two key concepts: (a) the energy deposited per gram in an absorbing medium, principally tissue, which is the absorbed dose, and (b) the damaging effect of the

radiation type, which is characterized by the term effective dose equivalent. A related term is radiation exposure, which applies to air only and is a measure of the amount of ionization produced by x-rays and gamma radiation in air. Each of these is defined in the conventional system of units, which in various forms and refinements has been used for several decades, and in the newer SI system, which is gradually re placing the conventional units. This presentation uses both sets, but for the most part emphasizes conventional units, which are firmly embedded in governmental standards and regulations, at least in the USA. [8]

The five basic concepts that relate a radiation worker to a source of radiation are (1) activity (A), measured in becquerels (Bq) or in curies (Ci); (2) exposure (X), measured in Coulombs/kg of air or in roentgens (R); (3) absorbed dose (D), measured in grays or in rads, (4) equivalent dose (HT), measured in sieverts or in rems, and (5) effective dose (E), also measured in sieverts or in rems.

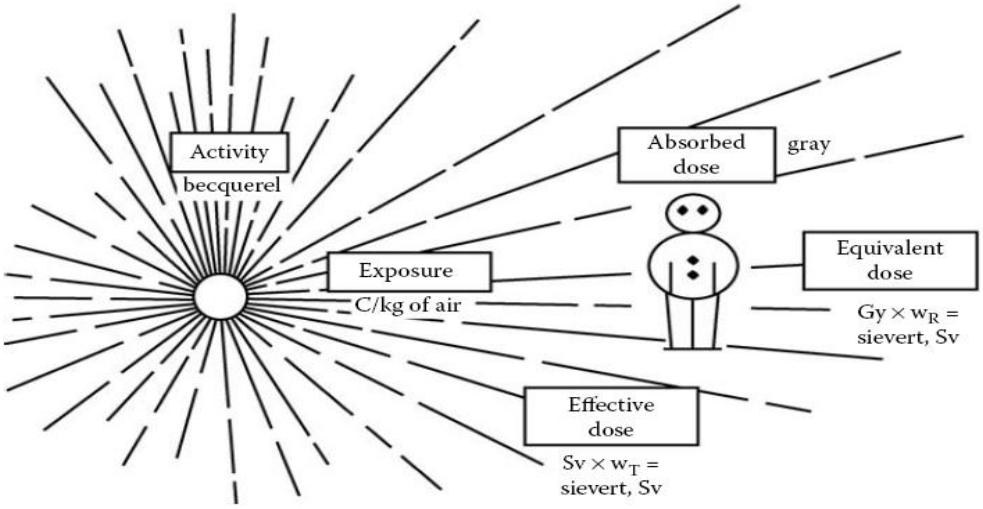


Figure [3.1] the five concepts that relate a radiation worker to a radioactive source

I. Exposure (X)

Exposure is the sum of all the electrical charges (Q) of one sign produced by x- or gamma rays in a given mass (m) of dry air at standard temperature and pressure (STP). The formal expression for exposure is

$$X = Q/m \tag{3.1}$$

The S.I. unit of exposure is the Coulomb per kg (C/kg) of dry air at STP. The special unit of exposure is the roentgen (R), defined as 2.85×10^{-4} C/kg of dry air at STP. This definition applies only to X- ray or gamma rays under 3 MeV of energy. One C/kg is equal to 3876 roentgens.

II. Activity (A)

Activity is an abbreviation of radioactivity. Activity is the number of radioactive atoms (N) decaying per unit time (t). The formal expression is

$$A = N/t \tag{3.2}$$

The S.I. unit of activity is the Becquerel (Bq), which is equal to one nuclear transformation per second (1 nt/s) Quantification (standardization) of a radioactive source, can be accomplished by relative and absolute methods of standardization.

III. Absorbed dose (D)

Absorbed dose (D) is the radiation energy (E) transferred to a unit mass (m) of any material. [47, 49] The formal expression is

$$D = E/m \tag{3.3}$$

The S.I. unit of absorbed dose is the gray (Gy), which is equal to the absorption of one joule of energy per kg of any material including the body of a patient or that of a technologist. The special unit of absorbed dose is the rad (radiation absorbed dose), defined as the absorption of one hundredth of a joule per kg of any material or 1 rad = 0.01 J/kg of material. To measure the absorbed dose by radiation workers, calibrated dosimeters are used. Examples are film badges and thermoluminescent dosimetry (TLD) badges to monitor body dose, and TLD ring dosimeters to monitor hand dose.

IV. Equivalent Dose ($H_{T,R}$)

This concept replaces the former "dose equivalent," and was introduced to account for the differences in the ionizing quality of the various types of radiations. Equivalent dose is defined as the product of the average absorbed dose ($D_{t,r}$) received by tissue($_t$) from radiation (r) times the radiation weighting factor (w_r):

$$H_{t,r} = D_{t,r} \times W_r \qquad (3.4)$$

Average doses, $D_{t,r}$, are expressed in grays (Gy) and equivalent doses, $H_{t,r}$ are expressed in sieverts (Sv) , Radiation weighting factors (w R) recommended by the NCRP for x-ray is 1 .

V. Effective Dose (E)

Due to the fact that organs and tissues have different sensitivities to radiation, the concept of effective dose (E) was introduced and is defined as the

sum of the products of the equivalent doses $D_{t,r}$ received by an organ times the tissue weighting factors W_r

$$E = \sum D_{t,r} \; X \; W_r \qquad (3.5)$$

Effective doses are also expressed in sievert (Sv) Table [3.1] shows the NRCP recommended tissue-weighting factors [45, 46, 22].

Gonads	0.20
Bone marrow	0.12
Bone marrow	0.12
Lung	0.12
Stomach	0.12
Bladder	0.05
Breast	0.05
Liver	0.05
Esophagus	0.05
Thyroid	0.05
Skin	0.01
Bone surface	0.01
Remainder	0.05
Total Body	1.00

Table [3.1] Tissue-weighting factors

3.2.2 Types of Radiation Effects

I. Acute and Chronic Exposures

Radiation effects depend on the type of exposure. Acute exposure means a high dose delivered within a short time: many rems (mSv to Sv) delivered within seconds to minutes. That type of exposure may be seen in partial-body irradiation during cancer therapy. The results of such exposure are deterministic effects. Chronic exposure means that one or more very low doses are delivered during a long time interval, for example, the exposure received by radiologists and NM personnel in the practice of their professions. The results of this type of exposure are stochastic effects.

II. Deterministic Effects

Deterministic effects are the result of a high-level exposure such as the ones that occur in a radiation accident. In that case, the severity of the effect (injuries, symptoms, and signs of illness) increases with the magnitude of the dose received. Typically, the curve has a threshold below which the effect is not observed. Fig [3.2] curve B, shows that, below 20 rems (200 mSv) of whole-body irradiation with hard x-rays or gamma rays, the signs or symptoms of illness are not observed. This is the threshold for radiation illness. At a dose above 20 rems (200 mSv), the illness, also known as the acute radiation syndrome (ARS), becomes more severe, and the severity increases with dose. At a dose of 100 rems (1 Sv), illness approaches 100%, which is the threshold for radiation deaths, another example of deterministic effects. To prevent deterministic effects, such as radiation accidents, experts must set dose limits

well below thresholds, and medical personnel must practice radiation safety at all times [50].

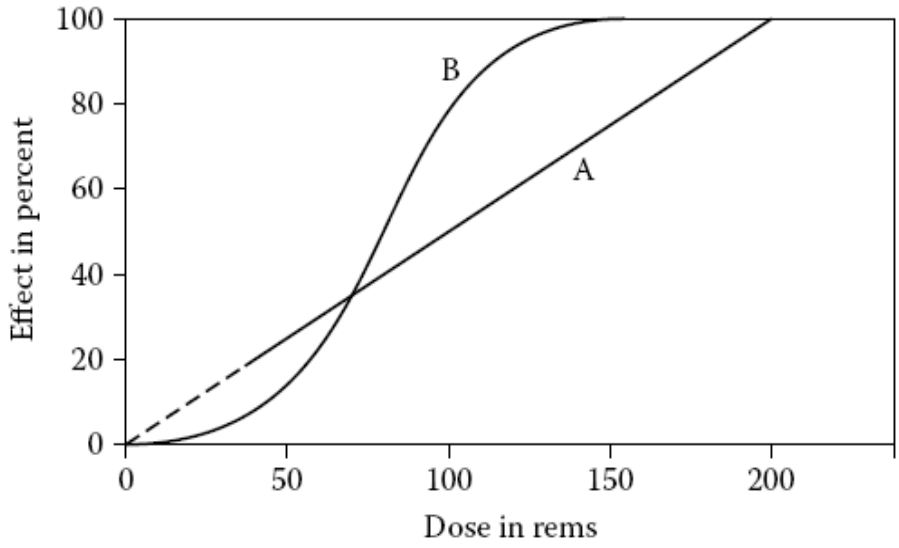


Figure [3.2] (A) Stochastic effects radiation dose. (B) Deterministic.

III. Stochastic Effects

In stochastic effects, the probability of observing the effect increases with the dose received. The accepted point of view is that the curve has no threshold and shows that the effect increases linearly with dose. Figure 3.2, curve A, shows the straight line characteristic of stochastic effects. In this case, there is an increase in the number of genetic mutations with dose, according to estimates by experts. Notice that the low side of the curve is represented by an interrupted line to indicate that the data are either incomplete or inconclusive. Other examples of stochastic effects are the risk of cancer or leukemia and life shortening. To prevent the risk of stochastic effects, ALARA must be practiced at all times [51].

3.3 **The Basic Principles of Radiation Protection**

In general the basic means of reducing your exposure to radiation and keeping your exposure ALARA regardless of the specific source of radiation [52] are as follows:

- Keep the time of exposure to a minimum
- Maintain distance from source
- place shielding between yourself and the source
- Protect yourself against radioactive contamination

The radiation worker can control and limit his/her exposure to penetrating radiation by taking advantage of time, distance, and shielding [53, 54].

By reducing the time of exposure to a radiation source, the dose to the worker is reduced in direct proportion with that time. Time directly influences the dose received: if you minimize the time spent near the source, the dose received is minimized.

The exposure rate from a radiation source drops off by the inverse of the distance squared. If a problem arises during a procedure, don't stand next to the source and discuss your options with others present. Move away from the source or return it to storage, if possible.

The third exposure control is based on radiation shields, automatic interlock devices, and in-place radiation monitoring instruments. Except temporary or portable shields, this type of control is usually built into the particular facility [55- 58].

3.4 Factors controlling the quality of radiography.

There are various factors controlling the quality of X-ray pictures. We provide a brief note enlightening the important points to be observed to produce a good radiograph of acceptable quality. X-rays that come out from the window of the X-ray tube are called primary beam. That pencil of radiation at the geometric centre of this primary beam is said to be the central ray. The wave length of X-ray is one billionth of an inch. They are penetrating in nature. They penetrate through all materials. The penetrating power of X-rays depends upon atomic number, thickness of the part of the body to be traversed and the density of the material. For instance, bone absorbs more X-rays than flesh. Flesh absorbs more than air. That is the reason in a chest radiograph we see different densities depending upon the rate of absorption by the body tissues. This helps us to distinguish and differentiate the different parts in the image. This is otherwise called contrast. Hence, the contrast of the photographic effect on the film can be defined as the degree of differences in density between, around and adjacent areas. The density can be defined as the photographic effect producing different degree of blackness on the film measured by light absorption through it.

The factor that controls contrast is said to be KV and the factor that controls density is termed as mAs i.e. the product of milliampere and the duration of exposure.

Formation of Radiographic Image

The radiographic image is produced by radiation transmitted in varying amounts through the subject by activating silver halide grains in an emulsion present in the film and by producing black deposits of silver. The action of X-

rays in combination with intensifying screens is amplified many times. Thus, a latent image is recorded on the film. By subjecting this film to chemical processing, a real permanent image can be produced which can be viewed through transmitted light. For proper recording of the radiographic image, the following are essential:

(1) Radiant energy, (2) subject (patient) to record, (3) photo-sensitive material and (4) chemicals for processing.

Characteristics of a good radiograph

The image produced should have (1) sufficient sharpness and (2) sufficient radiographic contrast.

Sharpness of the radiographic image

The radiographic image is a shadow image. It is made by the body intercepting radiation. In practice, these shadows always possess some diffusion of outline and the total degree of diffusion is a composite of several factors. These can be divided into three main groups:

(a) Factors connected with the geometry of shadow formation known as geometric factors.
(b) Factors connected with the subject and its movement known as motional factors.

(c) Factors connected with the recording of the image known as photographic factors.

<u>Geometric factors of unsharpness depends upon</u>:

(i) The size of the X-ray source (target or the focal spot), smaller the focal spot lesser the image unsharpness.

(ii) The distance between the X-ray source and the recording surface which is the film (larger the distance lesser the unsharpness).

(iii) The distance between the film and the subject being radiographed. (Closer the distance between the film and the subject lesser the unsharpness).

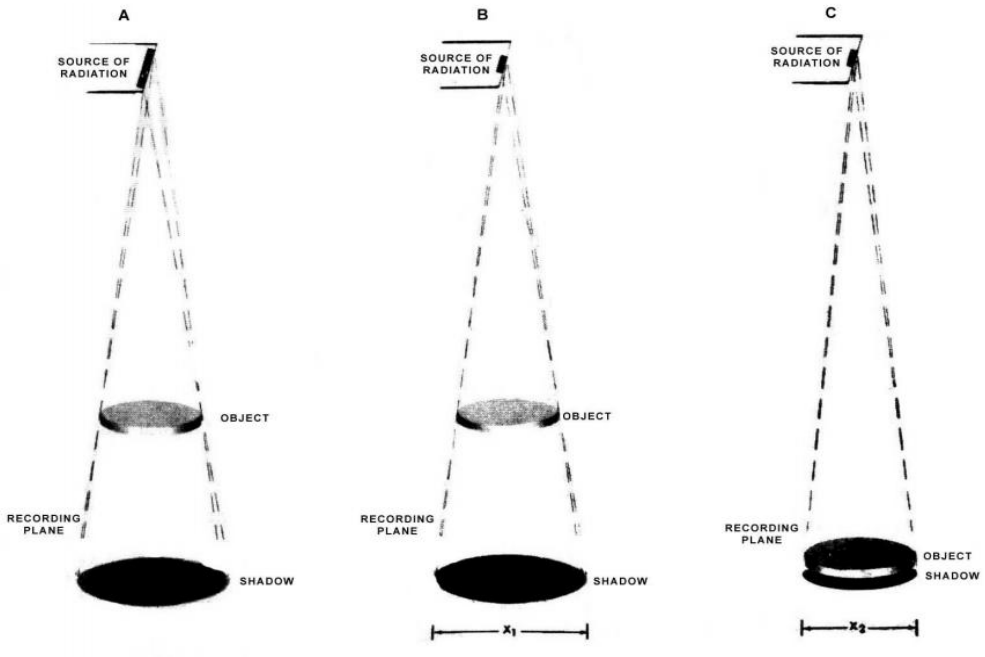


Figure [3.3] Diagrams showing geometrical effects on image sharpness. A: Unsharpness caused by a large focal spot and long object-film distance. B: Improvement in sharpness produced by a small focal spot. C: Superior result of a small focal spot and minimum object-film distance. Notice the enlargement (x1) caused by a long object-film distance as compared to the more accurate size (x2) produced when the object is close to the film.

Motional factors affecting unsharpness

(i) Involuntary: Peristaltic action in the abdominal viscera and the heart beat.

(ii) Voluntary: Movement of the body sufficiently blurs the shadows outline.

Photographic effect of unsharpness

(i) Intensifying screen: It is said that the capacity of an emulsion to resolve detail are influenced by grain size, contrast and the extent to which it may absorb or scatter light.

The salt screens are commonly made of calcium tungstate crystals. The dimensions of these are expressed in microns (a micron being a thousandth part of a millimeter). Even the smallest lesion which could be absorbed on a radiograph is many times larger and its size is normally expressed in terms of one millimeter or more.

These crystals fluoresce as a separate entity under action of X-rays. The image is therefore produced by many such separate sources of light. These separate light sources result in images with a certain diffusion of outline. The extent of this diffusion depends on the size of the light emitting crystals and also upon the distance of the crystal from the film since there is divergence of screen light.

Hence, it can be concluded that screens made of larger crystals and deeper layers of crystals are faster, reducing the exposure time which results in greater screen unsharpness. Screens made of very fine grains and thin layer of crystals usually produce a high definition which results in minimum amount of screen unsharpness requiring longer exposure time. Film screen contact should be good to avoid unsharpness.

(ii) Emulsion factors: Film base is coated with light sensitive material called silver bromide crystals suspended in a suitable binding agent to make photographic emulsion. This sensitive emulsion is spread on base called polyester base (polyethylene terephthalate resin). When X-rays impinge on a photographic emulsion the latter gets converted into black deposits of silver and a latent image is formed.

These silver bromide crystals on being struck by the X-rays produce light, whose sideways dispersion leads to unsharpness of the image. This is minimized by giving anti-halation backing to the film which reduces the light getting reflected back to the emulsion from the base.

Besides, the above, it should be know that contrast of the radiographic image is controlled by three factors viz., (1) film contrast, (2) processing chemicals and (3) radiation factors of objective contrast.

Film contrast: Film contrast which is inherent in the film depends upon the size and sensitivity of the silver bromide crystals.

Processing chemicals - Factors involved are:

(a) Type and constitution of developing solutions

(b) Temperature of the developer

(c) Freshness of the developer

(d) Agitation of the film in the developer

Radiation factors of objective contrast – These are:

(a) Quality of primary radiation.

(b) The scattered radiation reaching the film and its effects on the image.

Primary radiation: A change in penetration of X-rays which is otherwise controlled by KV alters radiation contrast within the subject.

Scattered radiation: The X-rays that did not penetrate through the body are not entirely absorbed by the body tissues.

Exposure factors in diagnostic radiography

The exposure factors are:

1. Tube voltage expressed in KV.
2. Tube current expressed in milliampere.
3. Duration of exposure expressed in seconds.

Tube voltage

This alters the quality of X-ray beam, its penetrating power and its intensity, as the colour and the brightness are separate characteristics of a beam of visible light.

Tube current

Quantity of X-rays emerging from the X-ray tube which is proportional to the intensity of radiation and the same is recorded on the milliammeter.

Duration of exposure

Duration of radiation emerging out of the X-ray tube is measured in seconds. Duration of exposure of tube current control above controls quantity of X radiations which is measured in mAs.

Tube film distance

X-ray beam is divergent from its source and it covers an area which will increase in size with the distance from the source. The energy of the beam is therefore spread over a large area and the intensity is thus diminished as per the inverse square law which states that intensity of light or X-rays reaching the film

is inversely proportional to the square of the distance. From the above it can be inferred that (1) Exposure time can be decreased when KV is increased. (2) mA is raised when the tube film distance is shortened. (3) Exposure time must be increased when KV is lowered. (4) mA is lowered when tube film distance is increased[59,60].

3.5 The Factors Determining Image Quality

All medical anatomical images are taken with the aim of detecting abnormalities of anatomy resulting from injury or disease. The radiologist, whose job it is to examine and report on medical images, has to be able to spot very subtle changes in intensity and make a judgment about whether these changes are real reflections of pathological anatomy or simply artifacts of the imaging process. Given the range of normal anatomical variation, this task would be impossible, except in the most trivial of cases, without a rigid code of radiographic practice and very strict quality controls on the apparatus and materials used to produce the image. Good quality control of both clinical practice and materials requires precise definitions and quantitative measurements of the factors that affect the visibility of small features in the presence of larger scale background variations and noise. In all imaging uses of X-rays there is the additional constraint of radiation dose to both patient and staff that must be known precisely and controlled. All medical images, including X-ray images, are assessed by three quantities: the contrast, the degree of unsharpness (spatial resolution) and the patient dose that is incurred. Apart from the question of radiation dose, which does not apply either to MRI or ultrasound, the definitions and concepts described in this section are common to all medical imaging. [7]

3.5.1 Contrast

At least qualitatively, the idea of contrast is a familiar one. In a high contrast photograph, obtained in good light conditions, objects stand out and their edges are clearly defined. In poor light conditions, or worse in rain or fog, individual objects lose some of their clarity and their edges become softened or blurred. A simple model provides a quantitative description of this important concept. We define contrast in terms of the relative intensity change produced by an object. Referring to Figure 3.4 we take the intensity at the middle of our target object to be I_1, and a reference point in the image, just outside the object in the background to be I2 Then the contrast, C, is defined to be:

$$C= (I_1-I_2)/I_2 \qquad (3.6)$$

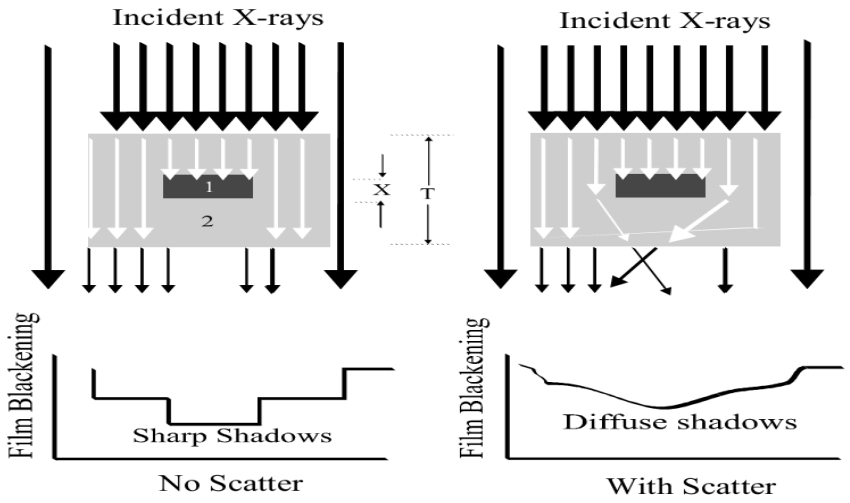


Figure [3.4] simple model defining contrast in X-ray radiography..

We can use this definition to estimate what differences in attenuation coefficient will lead to visible objects in a radiographic image and what this might mean in terms of the dose to the patient. Each of the intensities in our expression for the contrast consists of two terms arising from the primary or line

of sight X-rays, P, and the scattered X-rays, S, that reach the detector. Thus we write:

$$I_1 = P_1 + S_1$$
$$I_2 = P_2 + S_2 \qquad (3.7)$$

We can express the primary contributions in terms of the appropriate attenuation coefficients and thickness of tissue in the beam.

$$P_1 = Ne^{-\mu 1x} \text{ X } e^{-\mu(T-X)}$$
$$P_2 = Ne^{-\mu 2T} \qquad (3.8)$$

Where N is the number of photons per unit area, incident on the upper surface of our slab of tissue. When we put the expressions for P into our definition of contrast we have:

$$C = (N (e^{-(\mu 1-\mu 2)x} e^{-\mu 2T} - e^{-\mu 2T}) + S_1 - S_2) / (Ne^{-\mu 2T} + S_2)$$

Using $e^{-X} \approx 1-X$ for small X and assuming $S_1 = S_2$

We obtain $C = (\mu 1-\mu 2)\, x / (1+R)$

Where $R \approx S_2 / Ne^{-\mu 2T}$ (3.9)

We have written the average ratio of scattered to primary radiation as R. We have left the scatter contribution, S, as a symbol because it is difficult to calculate accurate scattering contributions. This simple expression shows that contrast is maximized by increasing the difference in attenuation factors between the target and its immediate surroundings, and decreasing the amount of secondary, scattered radiation that ends up on the film or in the detectors. Our equation shows that, other factors remaining constant, all X-ray radiographs benefit from a reduction in scatter.

3.5.2 Contrast Enhancement

In general, there are only small differences in attenuation coefficient between different types of soft tissue. Thus a small soft tissue target imbedded in a larger mass of soft tissue is generally unobservable unless other measures are taken. Such measures are called contrast enhancement .A good example of this is the injection of iodine into the blood stream just before the X-ray image is taken. Iodine increases the attenuation coefficient of the blood with respect to surrounding soft tissue, to the extent that small blood vessels less than 1 mm across can be imaged.

3.5.3 Scatter Reduction

The useful image forming primary radiation from an X-ray tube travels in a straight line from the X-ray tube focus through the patient to the detector. These ‘rays’ form the perfectly sharp shadows that are crucial for optimum contrast. Scattered radiation on the other hand arises from any exposed region of the body and ends up at a position on the detector that bears no simple relationship to the position of the X-ray tube focus. All methods of scatter reduction exploit this geometrical difference between primary and scattered radiation. Fig [3.5] illustrates some ways in which scatter reduction may be accomplished.

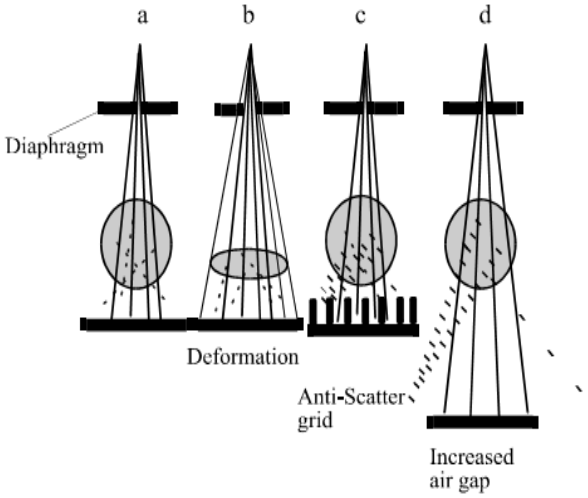


Figure [3.5] methods of scatter reduction.

Methods of scatter reduction as shown in Fig [3.5] (a) Diaphragms are incorporated into the X-ray tube housing to limit the angular extent of the beam and hence the volume of tissue exposed. (b) The tissue can be deformed (flattened) to reduce scatter. (c) Lead anti-scatter screens are put directly over the film that only let through X-rays travelling directly from the tube focus. (d) Increasing the air-gap decreases the flux of scattered radiation relative to the primary beam.

In some cases, the region of the body can be deformed to reduce the path length of the X-rays in the region of interest and hence reduce their scatter. This is a standard technique in mammography. If the distance between patient and detector is increased (the magnification factor), then scattered radiation is preferentially reduced with respect to the primary radiation. The scattered component tends to be distributed isotropically and thus its intensity falls off approximately as 1/ (patient–detector distance)^2 .The primary radiation, on the other hand, is far less divergent. Finally, by using an anti-scatter screen, directly in front of the film, scattered radiation is preferentially stopped. The screen consists of thin lead sheets arranged edge on, which allow the less divergent primary rays to pass on to the film but stop the more divergent scattered rays. Increased patient to detector distance and anti-scatter screens reduce both the primary radiation flux density at the film as well as scattered radiation. X-ray exposure times, and thus patient dose, have to be increased to compensate.

3.5.4 Detector Noise and Patient Dose

Our simple expression for the contrast, C , is dimensionless and gives no hint of how many photons are required to pass through the patient in order to achieve this level of contrast. Any photon detector, be it a film or an electronic

device, only records a small fraction of the photons passing through it. In addition, any detector introduces an inevitable degree of randomness or statistical noise into the recorded image that can only be reduced to an acceptable level by ensuring that a sufficient number of photons are actually recorded at any place in the image. Statistical noise is sometimes called quantum mottle because of the speckled or mottled appearance on film of featureless regions resulting from the random process of X-ray photon detection. In examinations in which small abnormalities are being sought, such false detail in an image is, of course, unacceptable. In principle, increasing the amount of radiation that is used in any examination could always eliminate the statistical noise problem. This is not always possible because of the biological hazard associated with ionizing radiation.

The radiation dose delivered to the patient is a critical factor in all examinations using ionizing radiation and thus all examinations should comply with the ALARA principle. In an ideal world all X-ray examinations would only utilize just enough radiation to resolve the required image feature at a level limited by the statistical noise created by the detection process.

3.5.5 Spatial Resolution and the Modulation Transfer Function

The question of how big an object can be seen or resolved is a critical issue for all imaging systems. Most of us either have had or will have direct experience of this with our own eyesight. With age, our ability to focus on near objects diminishes, so that reading a newspaper without spectacles becomes increasingly difficult. In order to focus light from the printed page on to our retina, without the aid of spectacles, we have to hold the newspaper further and further away but then the details of the printed letters become smaller and

smaller in comparison with the intrinsic spatial resolving properties of the retina. Thus the letters become increasingly blurred and eventually are illegible.

In general, spatial resolution (the size of the smallest clearly visible object) in any X-ray system depends in a complicated way on a number of factors. It depends on the tube focus, degree of scatter, attenuation coefficient and the noise from whatever source. Photographic film, used on its own, does not decrease the resolution, since the size and packing of the light sensitive AgBr grains is so fine that other factors dominate. This, however, is not the case when intensifier screens are used or electronic detectors replace film methods.

The intensifier screen illustrated in Figure 3.6 increases the sensitivity by factors of 100–1000 and hence dramatically reduces the radiation dose to the patient. This comes at the cost of a decrease in spatial resolution, as a result of the spread of light photons within the screen before they activate the film. Electronic detectors used in digital radiography or CT are composed of many discrete elements, each with a finite size. Each element averages together all photon events recorded, over its finite field of view, to produce a single number for that area. Its spatial resolution is then limited, approximately, to the physical size of the detector element. Clearly the microscopic particles of AgBr in a photographic film are very much smaller than even the most modern electronic detector.

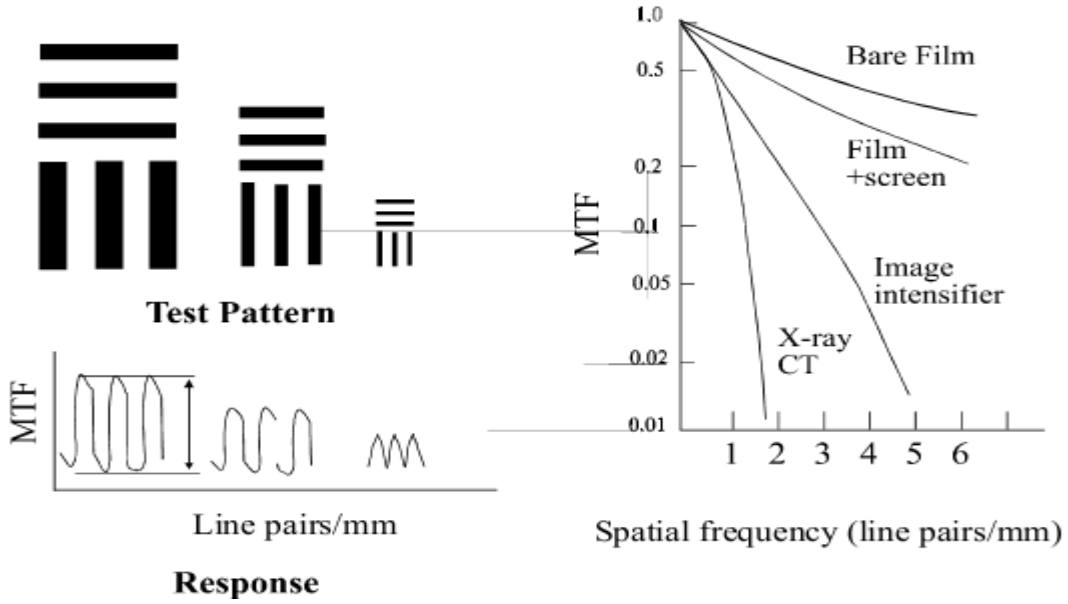


Figure [3.6] typical test pattern used to quantify the spatial resolution of an X-ray imaging system. The pattern would be laid out using lead strips. (b) Some typical values of versus line pairs per millimeter for some X-ray methods. It is generally accepted that an MFT value of 4% corresponds to a threshold of resolution. For CT, spatial resolution is about 1 lp/mm at a contrast level of 1% of water.

Spatial resolution, both in photography and throughout medical imaging, is described quantitatively by the response of the imaging system to a test pattern. This is just like the standard eyesight test card by opticians. The test pattern in X-ray radiography consists of pairs of lines of highly absorbing lead of varying widths and corresponding spacing's. The measure of resolution is the difference in contrast between black and white recorded from the test pattern. This is called the Modulation Transfer Function (MTF), and it is plotted or quoted with respect to line pairs per millimeter as shown in Fig [3.5].The MTF plot is directly analogous to a Bode plot, used in audio/electrical engineering, to describe the quality of a sound recording and reproduction system as a function

of temporal frequency. Both are in fact the Fourier transforms of the ratio of output to input amplitude for different values of test frequency [61-66].

Materials / Methods

&

Results / Discussion

Materials and Methods

4.1 Compared system

The X-ray system used in this study was special system used only in Posteroanterior chest radiographs, it is a photofluorographic system " conventional screen-film system " and the same system upgraded to digital with CCD technology. All imaging techniques were those used at each hospital in routine clinical imaging of the human chest wall.

4.1.1 Conventional X-ray system (Photofluorography)

The photofluorographic system consists of Korean Generator with role film Camera were used at **National Center of Diseases Control (Libya)** as shown in Fig [4.1] for all examinations, a fixed antiscatter grid (10:1, with 34 lines per centimeter) was used. The photofluorography and MIFA film (Fuji, Japan) processed in a COMPACT 2 processor (Protec, Germany) with a cycle time of 90 seconds developed; Film size was 100×100 mm.

Figure [4.1] Photofluorographic system

The system has the following specification

Generator

- Electric Power: Single phase 220V, 40KVA
- Rating: 125KV / 500mA
- Short time rating: Single phase: 125KV at 320mA, 80KV at 500mA
- KV range: 40-125KV
- mA range: 10-500mA
- mAs range 0.1-640mAs
- Time range: 0.001-10 sec
- Display system: Digital LED Display
- Anatomical programming: 216 programs
- Protections: Overload/filament stabilization protection circuits should be available Self-diagnostic message indication

X-ray Tube

- Maximum KVp rating: 125 KVP
- Focal spot size: 1.0 x 2.0mm
- Anode heat storage capacity: 140kHU

- Target angle: 16°
- Weight: 18 kg

Collimator

- Type: Manual, Rectangular
- Electric Power: 24VAC, 150W
- Max. KV range: 150KVP
- Max field X-ray beam: SID at 65cm less than 35cm x 35cm
- Min field X-ray beam: SID at 100cm less than 5cm x 5cm
- Dimensions: 183mm (W) x 183mm (D) x 179mm (H)
- Weight: 5 kg

High Voltage Cable

12m, length with terminal

4.1.2 Conventional X-ray system upgraded to digital using CCD technology

It was the same system, which was discussed before, but roll filme camera was replaced by Korean CCD detector fig [4.2] has the following specifications

- Detector Array: Full Filed, Single CCD.
- Active Image Size: 43cm x 43cm.
- Number of Active Pixel: 3,072 x 3,072 (9Mega).
- Spatial Resolution: 4.3 lp/mm.
- Bit Depth: 12, 14, 16 bit (optional).
- Preview Image: Less than 2 seconds.

- Processed Image Display: Less than 5 seconds.
- Detector size: 990mm.
- Cooling system: Air cooling system.

Figure [4.2] CCD Detector.

4.2 Measuring devices

4.2.1 Fluke Victoreen 4000 M

It is a self-contained, noninvasive X-ray Test Device in a single exposure; it simultaneously measures KVP, Exposure or Air Kerma, Exposure Rate or Air Kerma Rate, Time.

Figure [4.3] Fluke Victoreen

<u>Air Kerma and Air Kerma Rate</u>

Kerma is an acronym for Kinetic Energy Released in the Medium. Air Kerma, then, is a measurement of the amount of photon energy transferred to atomic electrons in a unit mass of air. The displayed units are joules per kilogram of air, also referred to as Grays. Grays computed by multiplying the exposure measurement (in Roentgens) by 0.00873 Gy/R.

4.2.1 PCXMC

It is a PC-based Monte Carlo program for calculating patients' organ doses and the effective dose in medical X-ray examinations. It uses mathematical phantom models, and can use to compute the doses in 25 organs of patients of different ages and sizes in freely adjustable X-ray projections and other examination conditions of projection radiography and fluoroscopy. The organ doses calculated with PCXMC Fig [4.3] agree well with the doses calculated by the National Radiological Protection Board (NRPB) for common X-ray examinations.

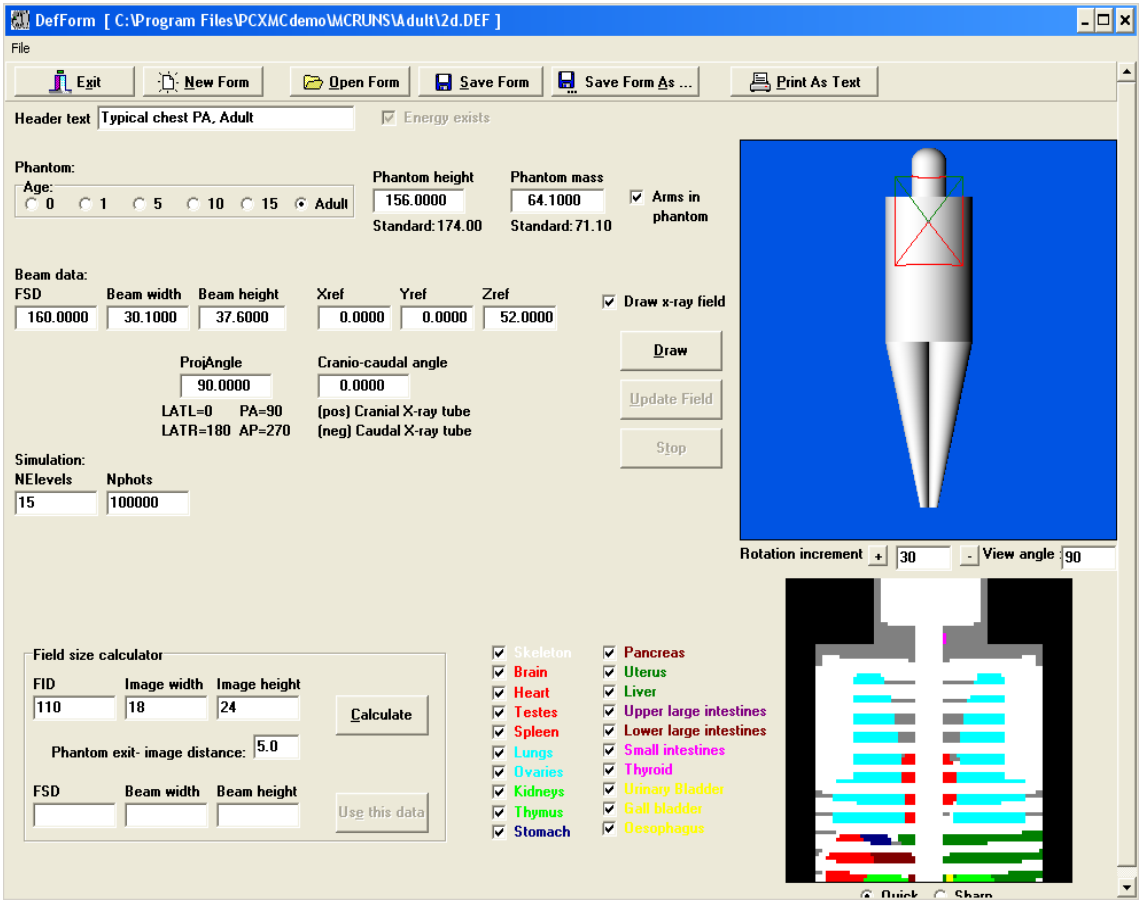


Figure (4.4) PCXMC

To determine effective dose, we will go through the following steps:

- Record all parameter affect on X-ray image and patient radiation dose.
- Fluke Victoreen 4000 M used to calculate air kerma.
- PCXMC used to calculate effective dose and result as this table was summarized it in another table listed in appendix C.

Organs	Dose (mGy)	Error (%)	Organs	Dose (mGy)	Error (%)
Lungs	1.192085	0.5	Pancreas	0.652822	1.7
Skull	0.024763	2.3	Small intestine	0.106532	1.5
UpperSpine	0.125663	3.2	Gall bladder	0.408826	2.8
MiddleSpine	2.801791	0.5	Upper large intestine	0.142084	2.4
LowerSpine	1.881384	1.1	Lower large intestine	0.023132	5.2
Ribs	3.523995	0.3	Urinary bladder	0.004693	23.6
Scapulae	5.447086	0.6	Uterus	0.016794	10.7
Clavicles	0.460678	3.2	Adrenals	1.468269	3.1
Upper arm bones	0.507335	1.4	Thymus	0.221871	5.3
Middle arm bones	0.580425	1.0	Oesophagus	0.489940	2.3
Lower arm bones	0.178282	1.8	Thyroid	0.075921	8.4
Upper leg bones	0.000334	23.6	Brain	0.003568	7.2
Middle leg bones	0.000000	NA	Testes	0.000518	99.4
Lower leg bones	0.000000	NA	Ovaries	0.017766	25.6
Pelvis	0.048679	2.7	Skin	0.305452	0.4
Heart	0.433913	1.4	Remainder (muscle)	0.306307	0.1
Breasts	0.249817	2.1	Total Body	0.406790	0.1
Liver	0.682435	0.6	Active bone marrow	0.415767	0.2
Stomach	0.413280	1.4	Skeleton	0.882250	0.2
Spleen	1.380136	1.3	Effective dose	0.396046	0.3
Kidneys	1.851349	0.8	Abs.fraction (%)	68.089527	

Table (4.1) PCXMC patient effective dose result

To determine Image evaluation, we will go through the following.

Scoring of patient images.—three experienced chest radiologists assessed all chest radiographs and recorded the score of the image quality on a questionnaire. In addition, ranked each image according to the following five-point 1: Identification marking of the patient 2: Patient positioning, 3: Sharpness, 4: Artifacts, 5: Ability for diagnosis assessed all images independently. Each factor scored using the following scoring system: 1 =“Good/None”, 2=“Fair” and 3=“Poor/Present”. The overall assessment of chest radiograph can be decided by the total score for the five factors, which ranges from 5 to 15 points. If the total score is 5or 6, it should be graduate as “Excellent”. If the total score is 8-11 without any “Poor/Present”, you should grade it as “Good”. If the total score is 8-13 with 2 “Poor/Present” or less, it should be grade as “Fair”. In addition, if the total score is 14 or more, or 3 “Poor/Present” or more, you should grade “Poor”. There was no limit imposed on reading time. Readers had no prior knowledge of exposure setting or imaging system. Photofluorographic radiography image viewed exclusively using a light box, while other Digital image viewed on Computer screen asked to evaluate the presence of abnormal features. [Appendix A]

Data analysis for Image and X-ray parameters:

Subjective assessment of image quality

As noted in the ICRU report (1996) “subjective” refers to individual human judgement, so all methods employing human observers are subjective. Subjective evaluation of image quality is of course useful – and often it is the only practical alternative for assessing clinical image quality.

Three radiologists (residents in their third, fourth, and fifth years with training in thoracic radiology from the beginning of their residencies) evaluated the hard copies independently and the digital image by subjective assessment of image quality.The method for assessment of the quality of chest radiography described here is one, which modified for resource-constrained settings by simplifying the JATA model. The simplified assessment method does not need special electrical devices. We begin on how to use the assessment sheet in appendix [A] followed by explaining more details about the procedures and factors necessary for assessment. As the 'five factors': Identification marking of the patient, Patient positioning, Contrast, Sharpness, and Artifacts explained in appendix [A].

The three factors (contrast, sharpness, and artifacts) are mainly dependent upon the specification of X-ray equipment, the combination of intensifying screen and film, X-ray exposure factors, and X-ray film processing procedure. They seriously affect the quality of chest radiographic imaging. [44]

Statistical Analysis

Three quantitative measurements calculated from our statistical analysis for the patient radiation dose (X-ray parameter and patient mass index). Calculations performed by means of statistics software (Excel, MedCalc).

1- The mean value of KV, BMI and effective dose for the results data of all Patients in our study.
2- Standard deviation for KV, BMI and Effective dose.
3- Correlation of KV, BMI and BMI, Effective dose for two systems. Finally, figuers show the relation between values.

4.3 Methods

4.3.1 Conventional X-ray system (film/screen)

Fifty patients referred to the radiology department for routine posteroanterior chest radiography in **National Center of Diseases Control (Libya).** At random, images obtained with the film/screen system. The posteroanterior chest radiographs obtained with the patients in an up-right position. The exposure values KV and BMI for every patient after each acquisition recorded.

Statistical analysis - Relation beteween KV and BMI**.**

	BMI(kg/m2)	KV
Mean value	22.08	86.745
STDEV	3.66	6.78
Correlation	0.709177	

Table (4.2) Mean value, STDEV, Corellation of KV and BMI Conventional X-ray

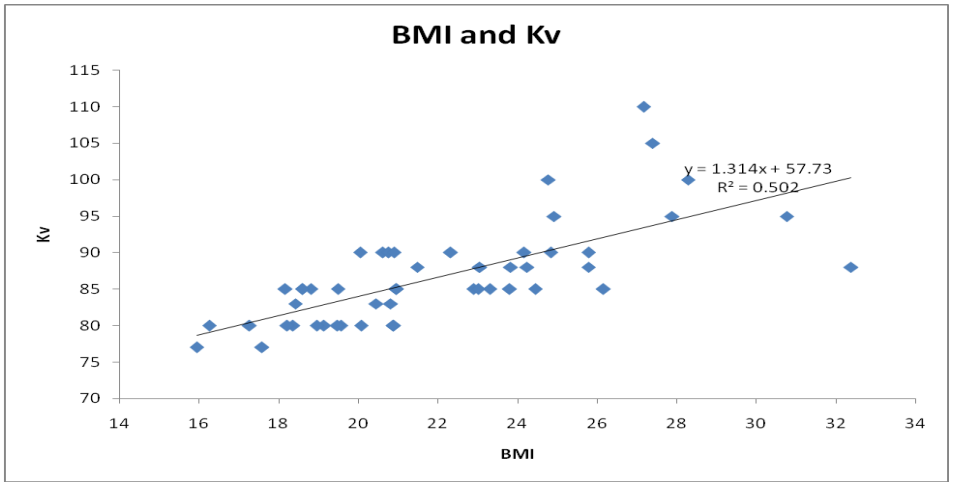


Figure (4.5) BMI (kg/m^2) and KV relation in Film/screen X-ray system.

Statistical analysis - Relation beteween BMI and Effective dose.

	BMI(kg/m^2)	Effective dose(mSv)
Mean value	22.08	.384
STDEV	3.66	.057
Correlation		0.553016688

Table (4.3) Mean value, STDEV, Corellation of KV and effective dose film/screen

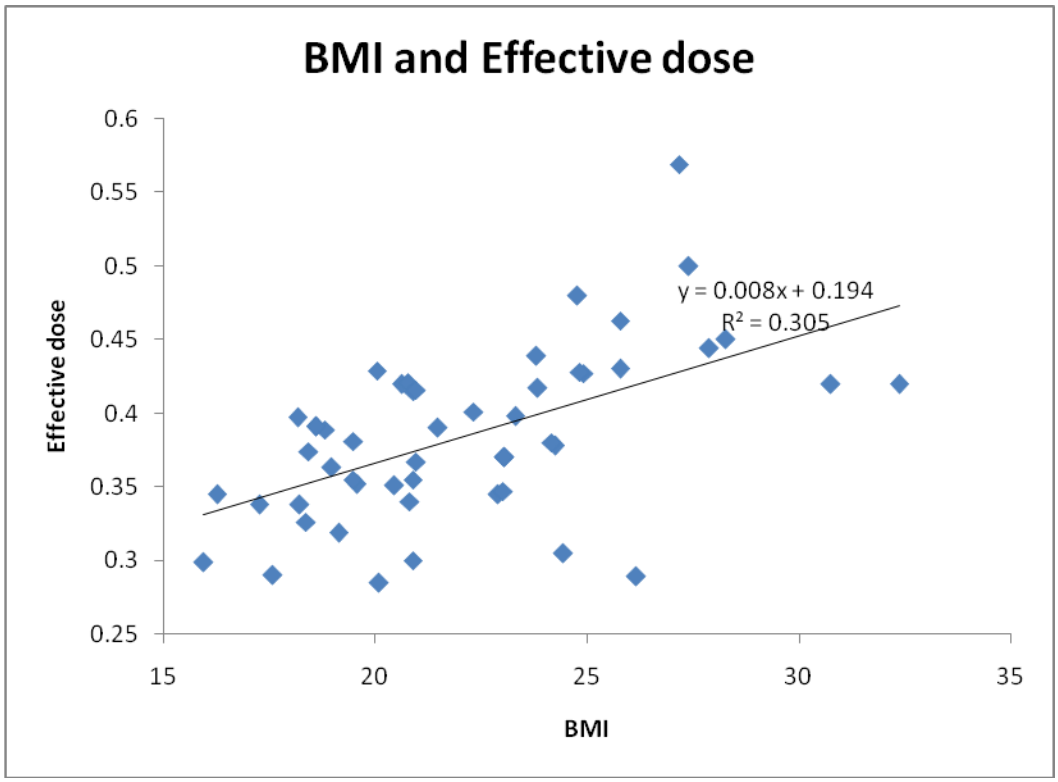


Figure (4.6) BMI (kg/m2) and effective dose (mSv) relation in conventional X-ray system

Image evaluation score for **conventional X-ray system** by doctors recorded in tables listed in appendix B.

Image evaluation score

First doctor image evaluation

Mean value of image evaluation score = 6.36, Standard deviation value for image evaluation score = 1.44

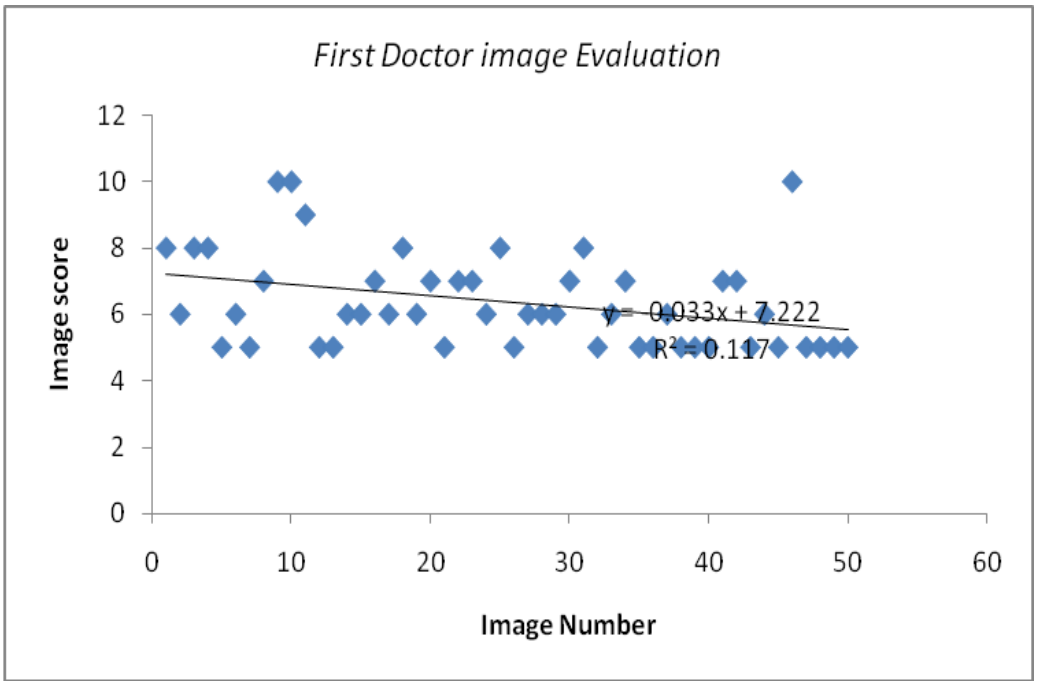


Figure (4.7) Image evaluation (first doctor) conventional X-ray

Second doctor image evaluation

Mean value of image evaluation score = 6.42, Standard deviation value for image evaluation score = 1.39

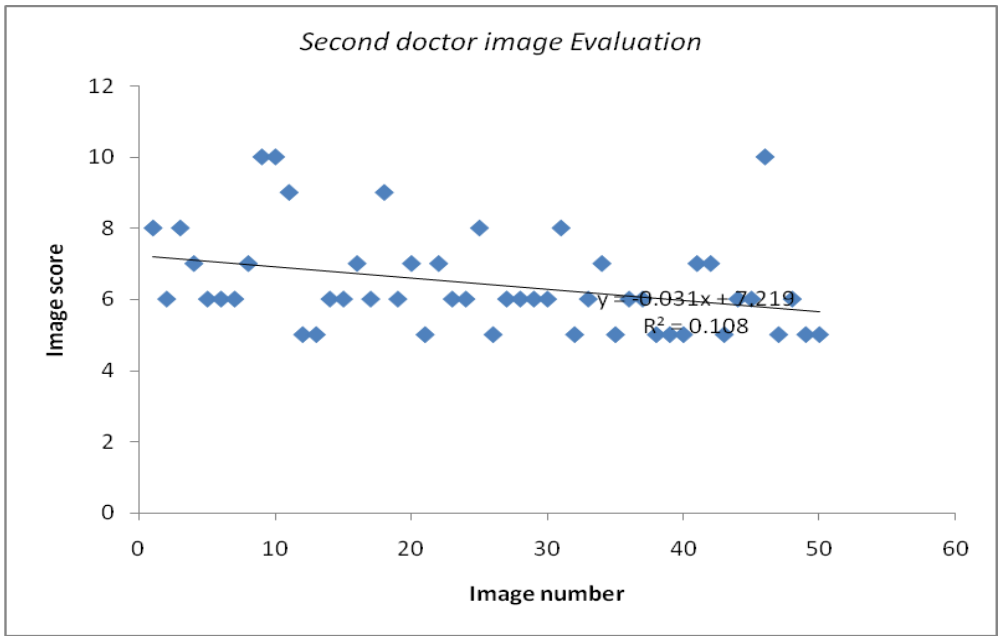


Figure (4.8) Image evaluation (second doctor) conventional X-ray

Third doctor image evaluation

Mean value of image evaluation score = 6.46, Standard deviation value for image evaluation score = 1.33

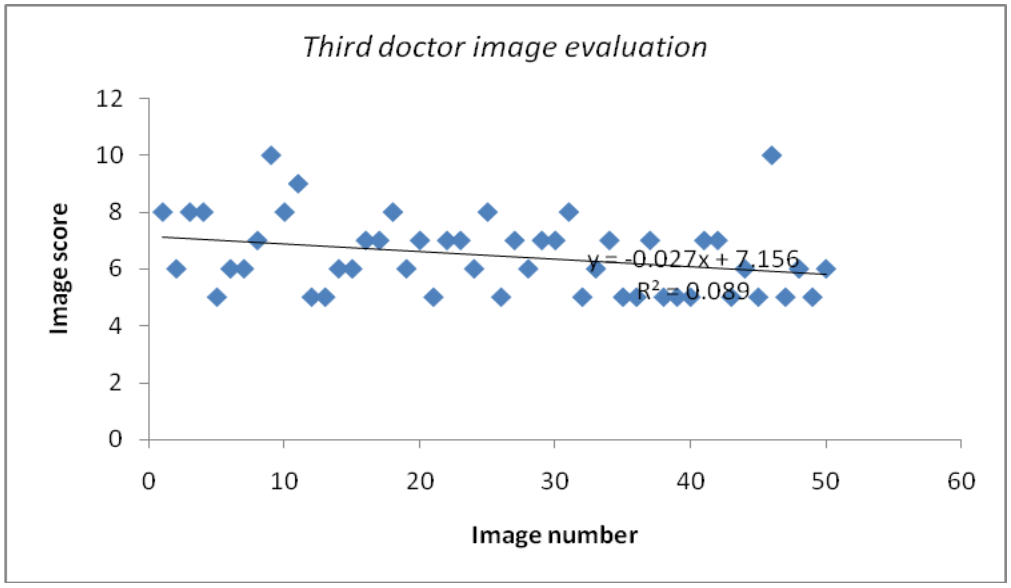


Figure (4.9) Image evaluation (third doctor) conventional X-ray

4.3.2 Digital X-ray system (based on CCD Technology)

Fifty patients referred to the radiology department for routine posteroanterior chest radiography in **National Center of diseases control (Libya).** At random, images obtained with the digital system (CCD camera). The posteroanterior chest radiographs obtained with the patients in an up-right position. The exposure values KVand BMI for every patient after each acquisition recorded.

Statistical analysis -Relation beteween KVand BodyMass Index.

	BMI(kg/m2)	KV
Mean value	26.75	87.02
STDEV	6.4	9.46
Correlation	0.850322	

Table (4.4) Mean value, STDEV, Correlation of KVand BMI digital X-ray

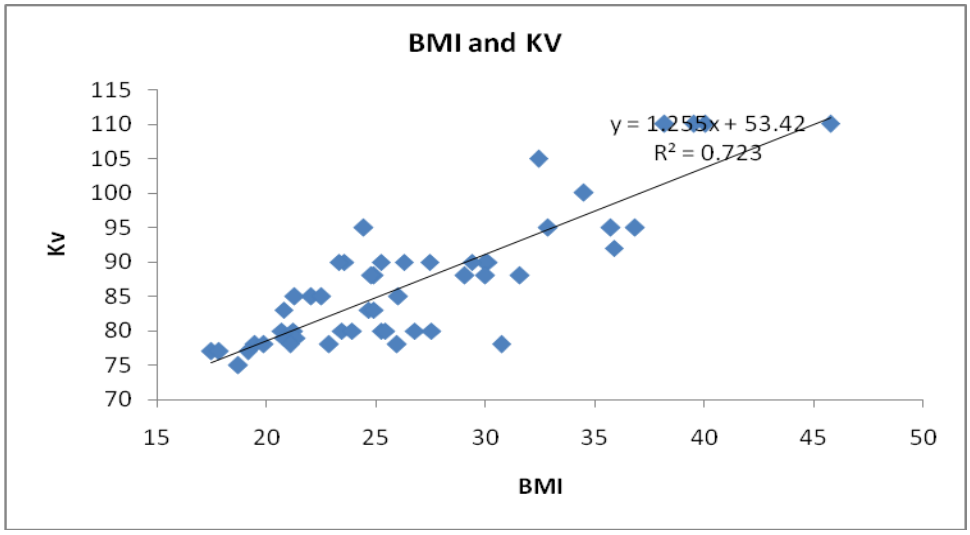


Figure (4.10) BMI (kg/m^2) and KV relation in Digital X-ray

Statistical analysis - Relation between BMI and Effective dose.

	BMI	Effective dose
Mean value	26.75	0.321
STDEV	6.4	0.056607053
Correlation		0.432861464

Table (4.5) Mean value, STDEV, Correlation of KV and effective dose Digital X-ray

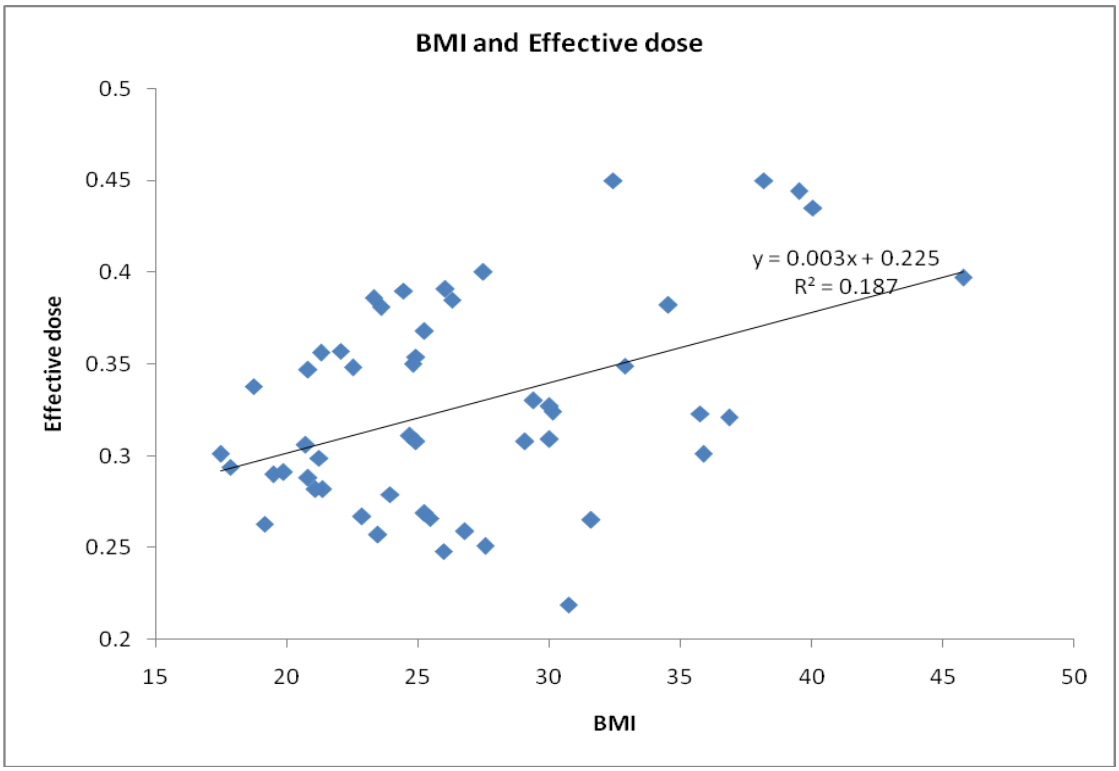


Figure (4.11) BMI (kg/m^2) and effective dose (mSv) relation in Digital X-ray system

Image evaluation score for **Digital X-ray system** by doctors recorded in tables listed in appendix B.

Image evaluation score.

First doctor image evaluation

Mean value of image evaluation score = 5.62

Standard deviation value for image evaluation score = .945

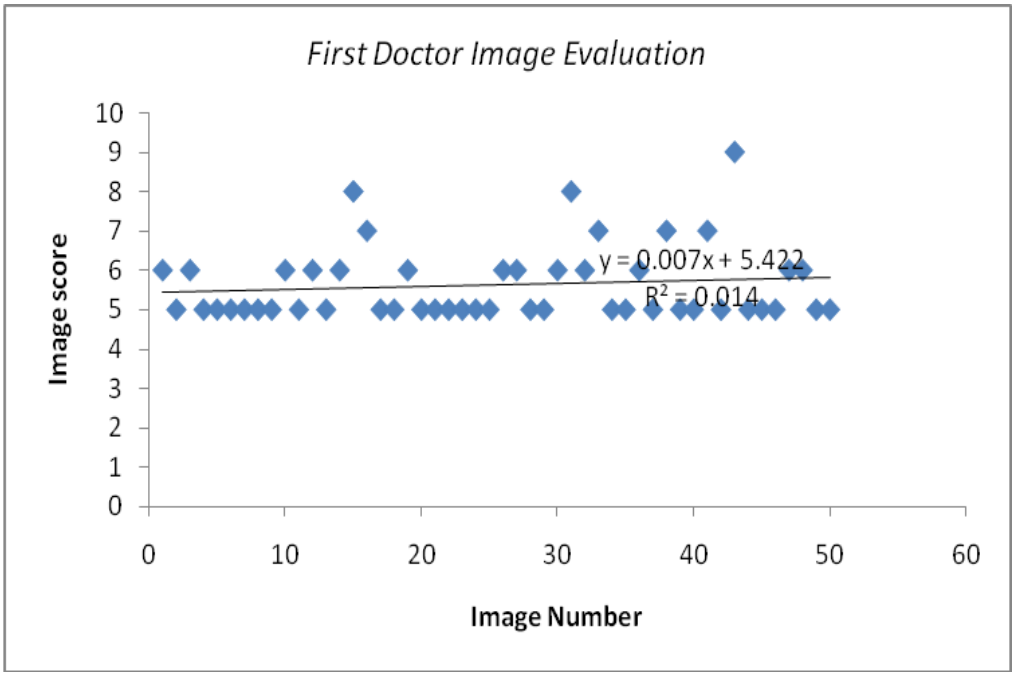


Figure (4.12) Image evaluation (First doctor) digital X-ray

Second doctor image evaluation

Mean value of image evaluation score = 5.52, Standard deviation value for image Evaluation score = .838

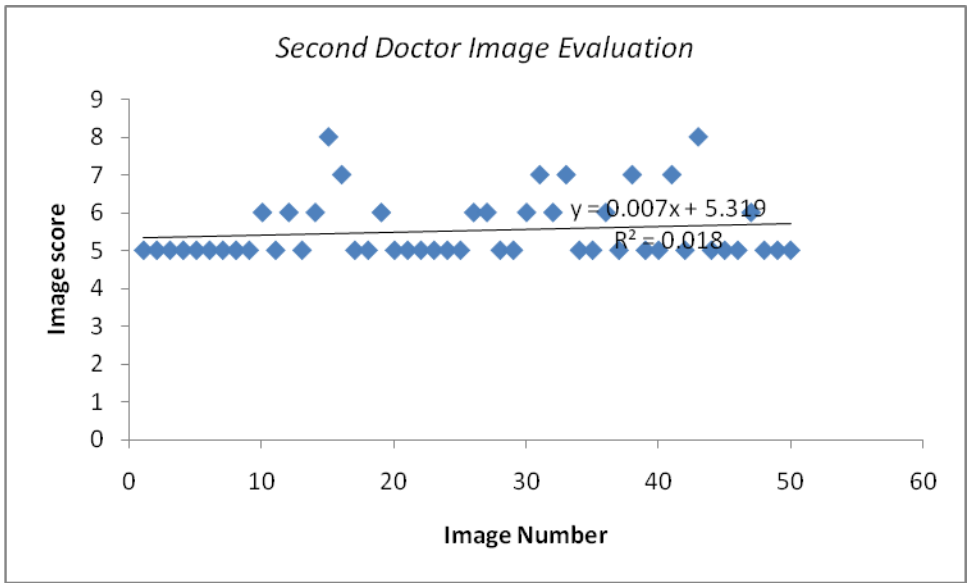


Figure (4.13) Image evaluation (second doctor) digital X-ray

Third doctor image evaluation

Mean value of image evaluation score =5.54, Standard deviation value for image evaluation score = .8621

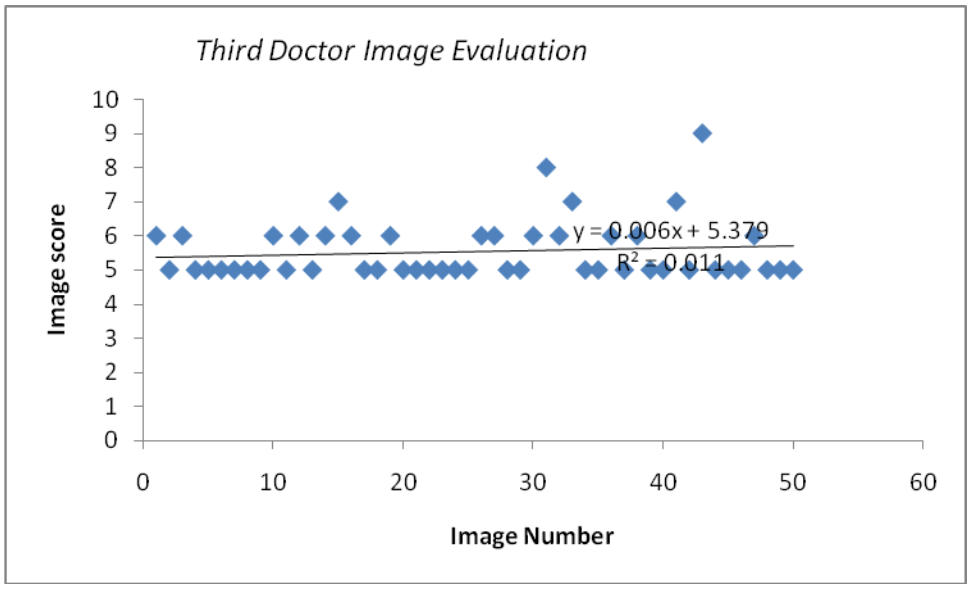


Figure (4.14) Image evaluation (third doctor) digital X-ray

4.4 Results

4.4.1 Results of Patient Radiation Dose Measurements

Form the data obtained in the previous tables and figuers the image quality can be evaluted and patient radiation dose can be comapred acccording to statistical analysis for the two systems. For both the film/screen and the CCD camera digital systems, 100 PA chest radiographs acquired. The patient population who underwent imaging on the film/screen system .In digital sytem mean values for effective dose was 0.321 mSv and in film/screen sytem mean values for effective dose was .384 mSv.

Statistical Analysis

Measurements and statistical analysis for relation between KV and BMI for both systems (digital and conventional).

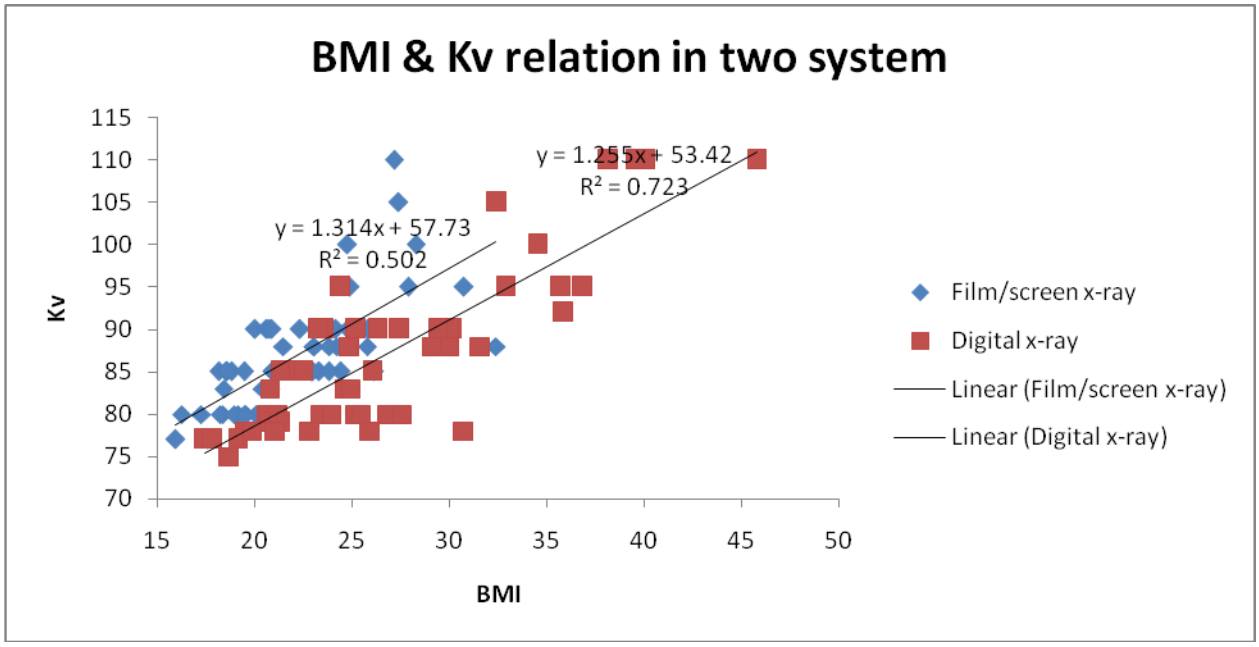


Figure (4.15) BMI (kg/m^2) and KV relation comparison in two system

Analysis for Relation between BMI and Effective dose for both systems.

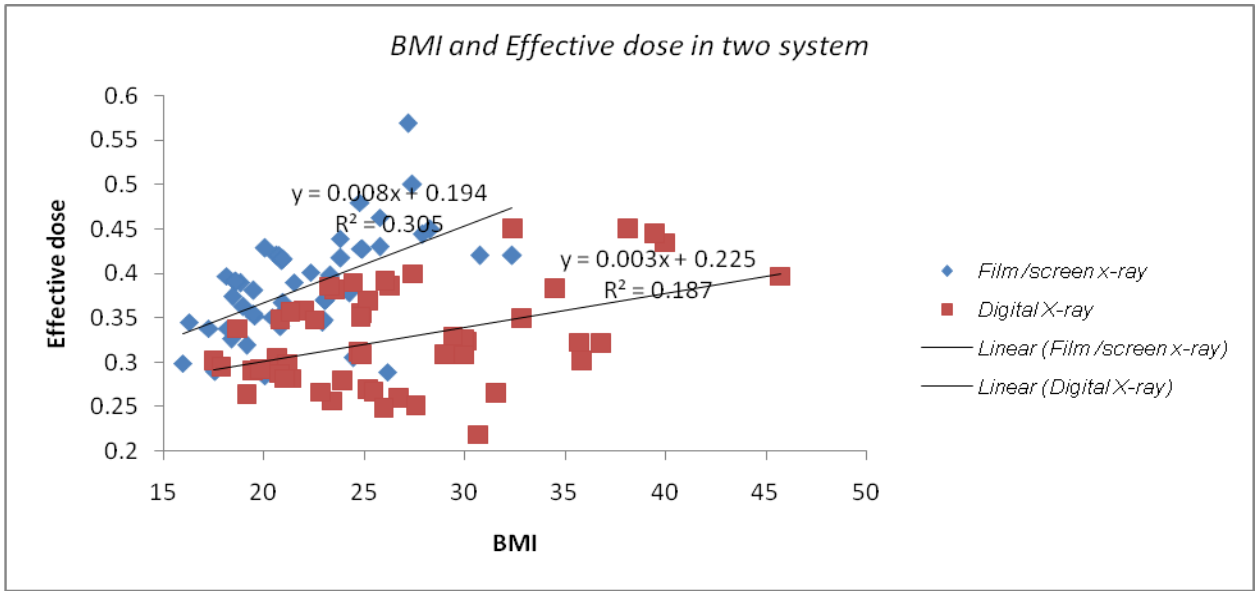


Figure (4.16) BMI (kg/m^2) and effective dose (mSv) relation comparison in two system

Discussion

Conventional projection radiography performed most frequently in diagnostic radiology Therefore; digital radiography systems are playing an important role in the evolution toward a digital medical imaging environment.In particular, chest radiographs represent about 25% of all diagnostic radiography examinations [67].

the results obtained in the present data and in Figures (4.15,4.16) , clearly demonstrated that there is no significant reduction in Patient radiation dose in digital X-ray system, a slight decline in the value of radiation dose was found but it's better than film / screen . This reduction leads to reduction in patient radiation

dose compared with conventional X-ray system .our results have been confirm the results reported by previous workers in this field . However, the digital X-ray system (conventional upgraded using CCD) considered as being more suitable for clinical studies than conventional X-ray system because:

1-For patient, techniation and doctors

- Reduction in KV leads to reduce in patient radiation dose.
- Reduction in KV leads to reduce in scatterd X-ray so it is good for technichian and doctors work close to X-ray machine.
- With patient radiation dose reductiaon which will lead to reduce the Biological Effects of Radiation.

2- For machine

- Reduction in KV leads to reduction of power used for X-ray machine then increase life of machine parts (X-ray tube, High voltage tank)
- Reduction in KV leads to reduce cost of electricity used for X-ray machine work

3-For Envirnoment

- Reduction in KV leads to reduce in radiation dose then reduces the total radiation by medical X-ray.

4.4.2 Results of Images Quality Evaluation

A significant positive correlation observed for the image quality and for the visibility of anatomic structures for all observer combinations: Spearman's rank correlation coefficient was in range (0.906-0.95).There were no statistically significant intermodality differences for the overall image criteria ($p < .0001$). The subjective image quality rating was satisfactory. No artifacts were observed.The ratings and their statistical significances for the matched modalities are shown in Fig (4.17, 4.18, 4.19), we found that image quality of CCD digital system images was best in the total calculated mean of ratings (mean scores: CCD digital, 5; film-screen system, 6) Fig. (4.20)

First doctor image evaluation (Digital and conventional system)

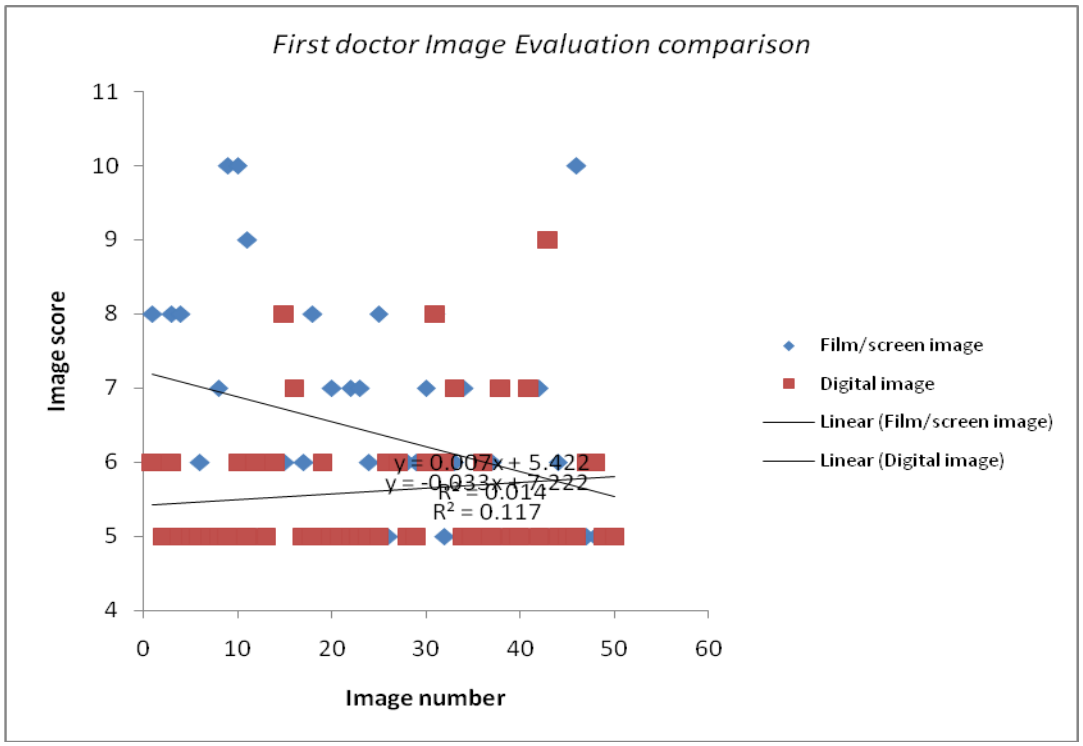


Figure (4.17) First doctor image evaluation between film/screen and digital

Second doctor image evaluation (Digital and conventional system)

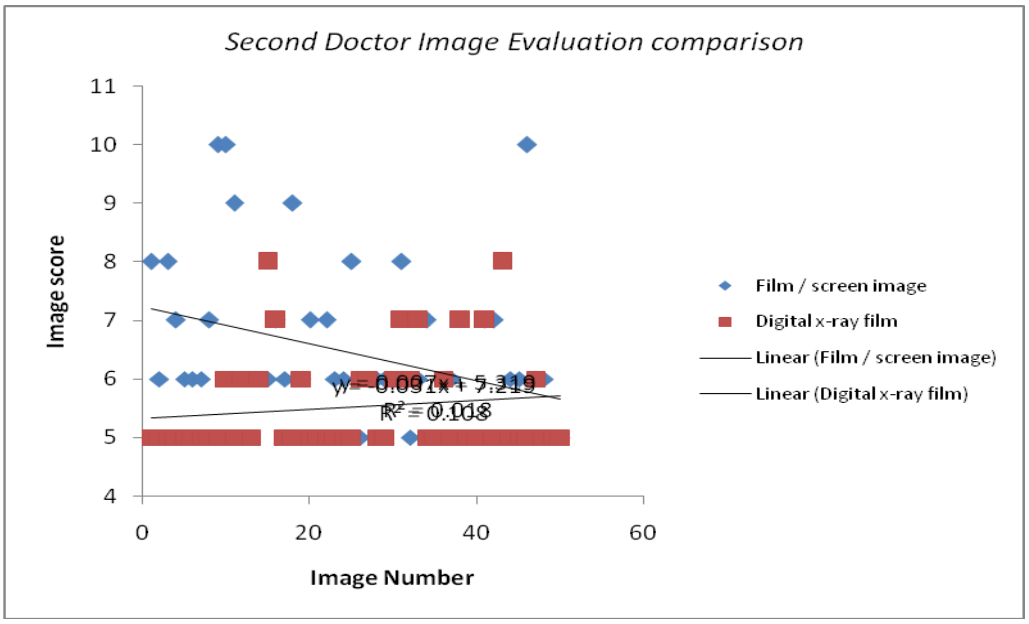


Figure (4.18) Second doctor image evaluation between film/screen and digital

Third doctor image evaluation (Digital and conventional system)

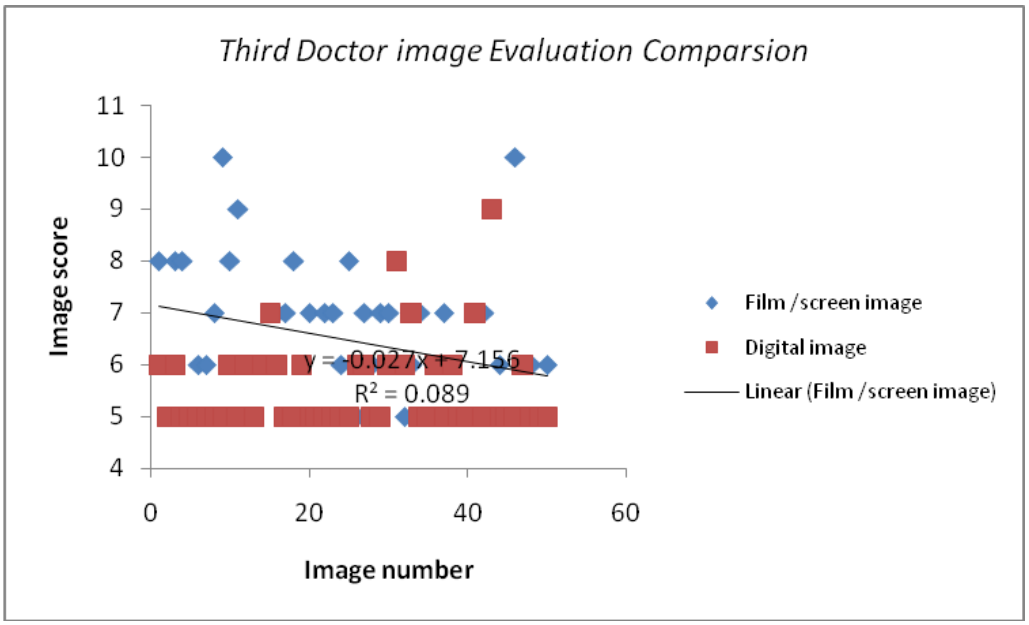


Figure (4.19) Second doctor image evaluation between film/screen and digital

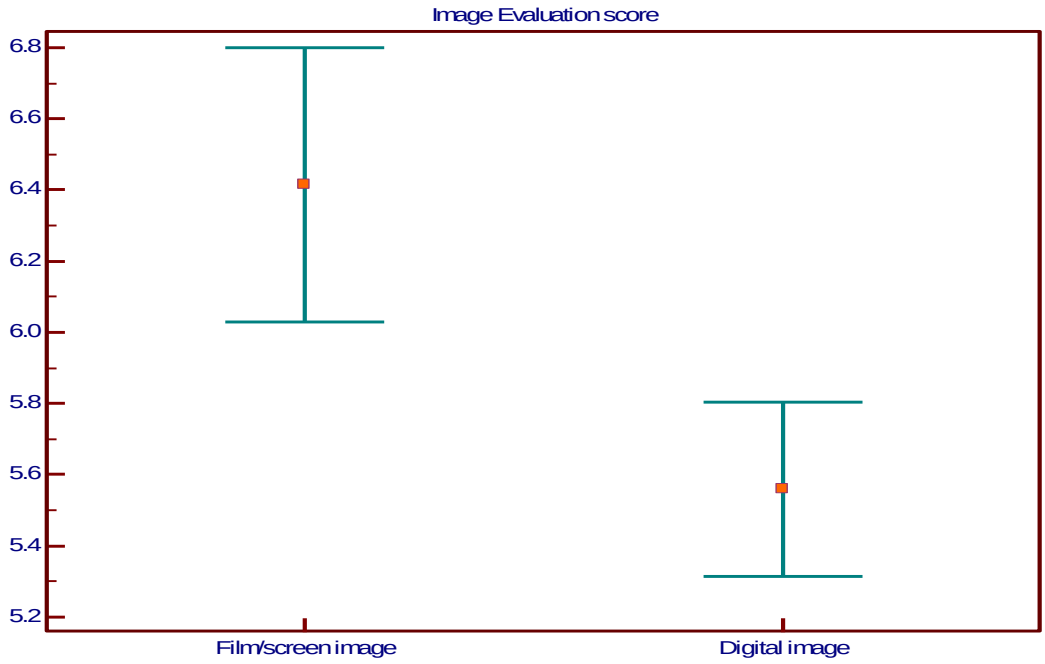


Figure (4.20) comparisons between evaluations image mean score in two systems

Discussion

The results obtained in the present data and Figure (4.17, 4.18, 4.19, 4.20) clearly demonstrated the digital X-ray image is better than film/screen imaged, this confirms the results reported by previous workers in this field. However, the digital X-ray system (upgraded using CCD) considered as being more suitable for clinical studies than film/screen X-ray system because when compared between the film-screen system and CCD detector system, the CCD has definitive practical advantages.The wide dynamic range providing a quick and easy image quality check. The radiologist is able to concentrate on specific aspects of the abnormality shown on the image because of a number of postprocessing features such as interactive window level setting and contrast enhancement. The digital image can easily be inte-grated into an RIS (radiology information system) and PACS (picture archiving and communication system) environment.

Conclusion

In last years, many of conventional film /screen X-ray systems upgraded to digital X-ray systems using different digital X-ray technologies (Computed radiography (CR), Flat panel detector and Charge-Coupled Device (CCD)). Therefore, we need further studies to know the effects on patient radiation dose and image quality when upgrading X-ray system to digital X-ray system. Most previous studies that have been made in this regard focused on the upgrading X-ray to digital X-ray using CR and Flat panel detector. In this study, we have been focusing on evaluation of image quality and radiation doses for patients undergoing chest X-ray examinations using a photofluorographic system "screen / film system and the same system upgraded to digital using (CCD) detector. Effective doses for patients undergoing chest X-ray using CCD detector found to be lower than Film / screen system. Image quality for X-ray image acquired using a digital CCD detector found better than X-ray image acquired using screen / film system. This confirms the results reported by previous workers in this field, however the digital X-ray system (conventional upgraded using CCD) considered as being more suitable for clinical studies than conventional X-ray system.

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Sistan and Baluchestan, Iran and comparing with that of international guidelines levels, Indian Journal of Science and Technology, 2011.

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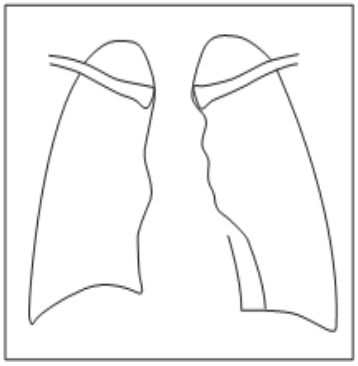
Appendix A

Sheet used for image evaluation

The factors presented here are the minimum required for overall quality assessment of chest Radiography

<table>
<tr><td>1) Identification marking of the patient</td><td>1-Good</td><td>2-Fair</td><td>3-Poor</td></tr>
<tr><td rowspan="2">2) Patient positioning

"Good":
“Fair”: neither “Good” nor “Poor”.
“Poor”: the number of problems you observed is 5 or mo re</td><td colspan="3">Please check the following 7 items:
i. <u>Defective lung fields Yes ☐ NO ☐</u>
ii. <u>Poor inspiration Yes ☐ NO ☐</u>
iii. <u>Oblique positioning Yes ☐ NO ☐</u>
iv. <u>Position of clavicles Yes ☐ NO ☐</u>
v. <u>Position of scapulas Yes ☐ NO ☐</u>
vi. <u>Asymmetric density of lungs Yes ☐ NO ☐</u>
vii. <u>Foreign substances Yes ☐ NO ☐</u></td></tr>
<tr><td>1-Good</td><td>2-Fair</td><td>3-Poor</td></tr>
<tr><td>3) Sharpness</td><td>1-Good</td><td>2-Fair</td><td>3-Poor</td></tr>
<tr><td>4) Artifacts
1-“None” : no artifacts in the lung fields.
2- “Slight” : small and slight artifact(s) in the lung fields
3- “Present”: clear and wide artifact(s) in the lung fields</td><td colspan="3">If 2 or 3 selected set the position of artifact
  </td></tr>
<tr><td>5) Ability for diagnosis</td><td>1-Good</td><td>2-Fair</td><td>3-Poor</td></tr>
<tr><td colspan="4">Assessment result: Total score for the four factors is ……(0-15)</td></tr>
<tr><td>Comments</td><td colspan="3"></td></tr>
</table>

(1) Identification marking of the patient

"Good" example includes all the four identifications (name of patient, age, name of health facility, and date of examination) outside the lung fields.

- "Fair" example is missing 1 or 2 identifications of the four outside the lung fields.
- "Poor" example has no information and/or marking material placed in the lung fields.

2) Patient positioning

Patient positioning is checked in terms of the following 7 items:

i. Defective lung fields

ii. Poor inspiration

iii. Oblique positioning

iv. Position of clavicles

v. Position of scapulas

vi. Asymmetric density of lungs

viii. Foreign substances

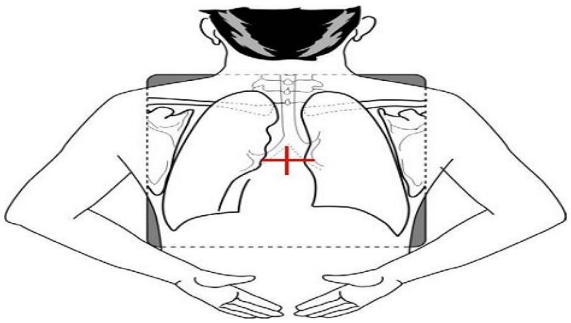
Among the seven, the first 2 items, Defective lung fields and Poor inspiration, are given higher priority due to their significance. If either of the two is observed, you should score it as "Fair" at best. If both of the two are observed, you should sc ore it as "Poor" regardless of the results of the other five items.

"Good": the number of problems excluding defective lung fields or poor inspiration is 0 or 1.

"Fair": neither "Good" nor "Poor".

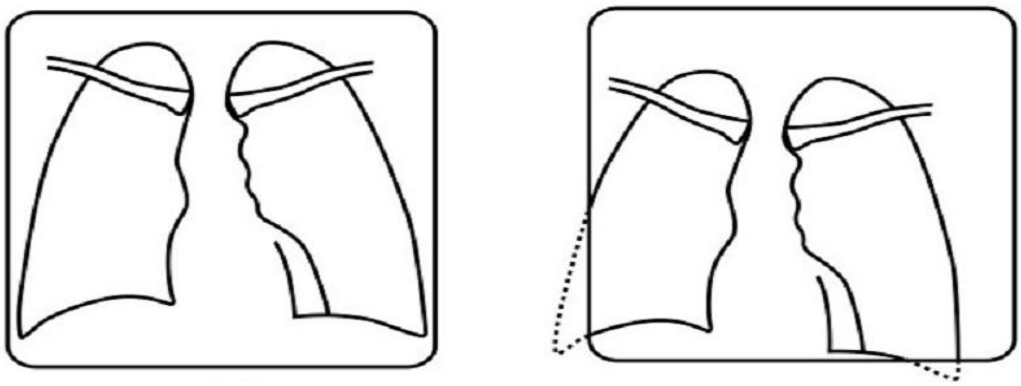
"Poor": the number of problems you observed is 5 or mo re, or with both defective lung fields and poor inspiration.

The gold standard for the positioning of a general chest radiography is 'Erect' meaning the patient is standing, and the X-ray beam directin is 'Posterior-Anterior: PA' meaning back to front. The recommended positioning of chest radiography is always PA even for children and patients who are sitting on a stool because they are unable to stand.



Defective lung fields

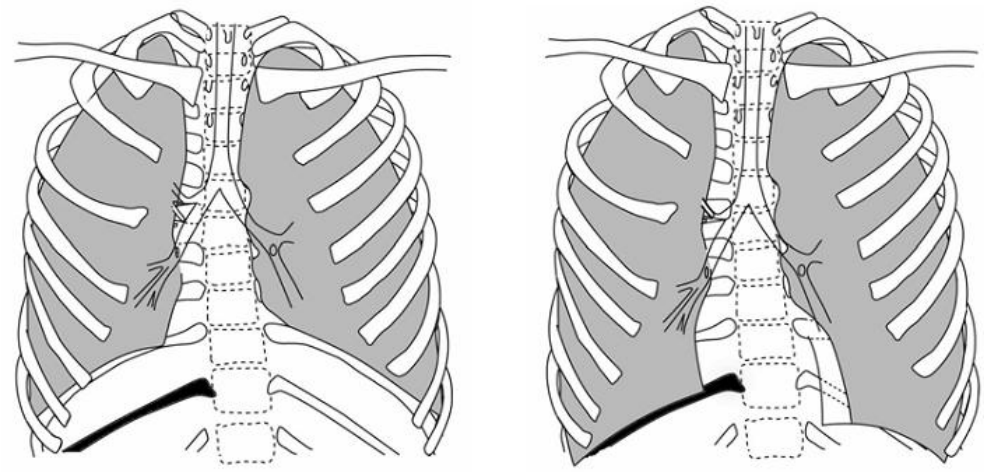
If we observe that any defective part of the lung fields, even a small portion, exist in the X-ray film when you put the chest radiograph on a film viewer, you should tick YES for "Defective lung fields".



ii. Poor inspiration

- If we find all of the lower line of the 10th rib (usually of the right lung) seen below the line of the right diaphragm, you should tick YES for "Poor inspiration"

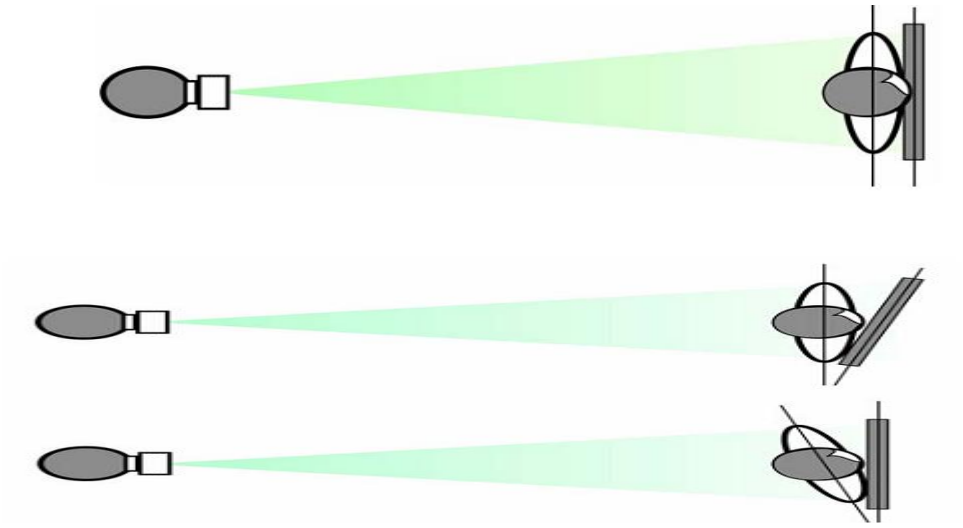
The patients must fully understand the meaning of holding a deep breath during X-ray exposure. Proper understanding of the procedure for chest radiography by the patients achieves the cooperation of patients, which is one of most critical factors for producing good quality chest radiography.



Schematic illustrations of bad (left) and good (right) inspiration

iii. Oblique positioning

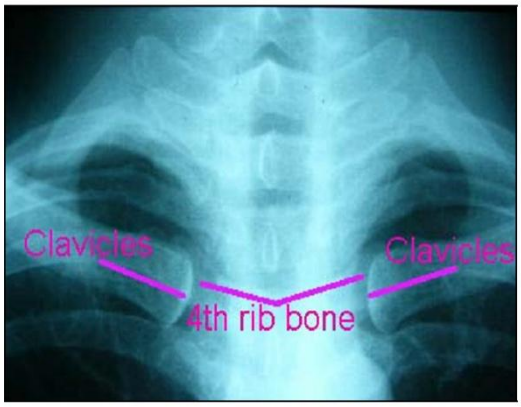
If you observe that the sterno-clavicular joints are asymmetrically placed and the differences of more than 5mm between the right and the left, namely, the difference in the distances from the right or the left clavicles to the spinous process of thoracic vertebrae is more than 5mm, you should tick YES for "Oblique positioning".With regard to the positioning of the patient for PA technique, it is essential that 3 things be well aligned: the patient, the X-ray cassette and the X-ray beam.

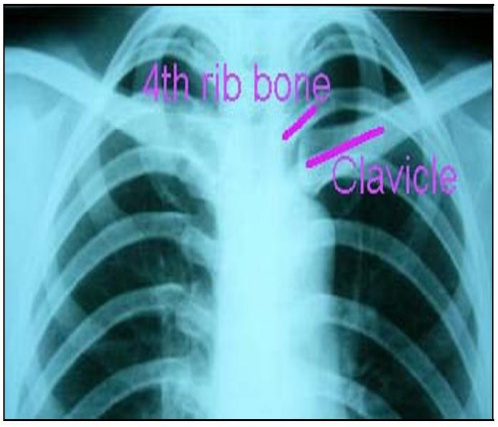


iv. Position of clavicles

If we observe that the clavicles are visualized apart from the 4th rib bones on X-ray film, you should tick YES for "Position of clavicles".

The clavicles of the patient must be placed at any part of 4th posterior rib bone on the X-ray film. To avoid overlapping of the clavicles on the apices of the lungs, the shoulders of the patient must be pressed well toward the X-ray film cassette, as with the positioning of the scapula.





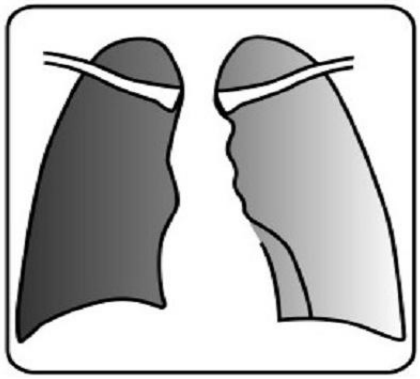
v. Position of scapulas

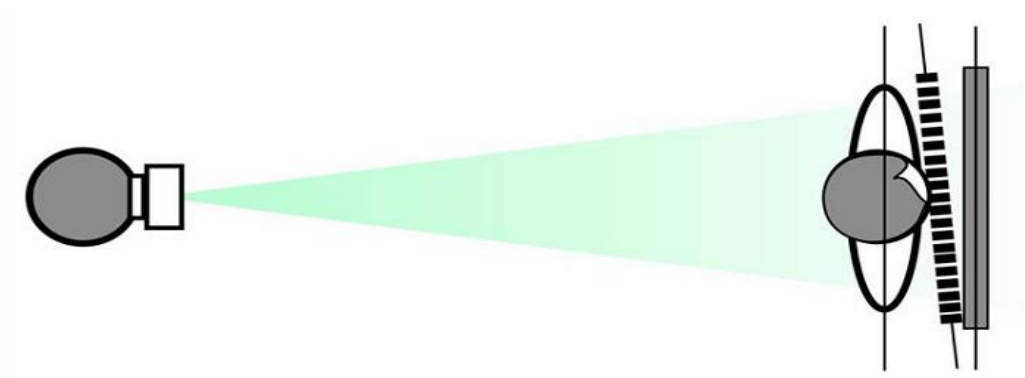
If we observe that the right or left scapula, or both are visualized more than 1 cm in the lung fields, you should tick YE S for "Position of scapulas".

The shoulders of the patient should be pressed well forward to the X-ray film cassette to avoid overlapping the imaging of scapulas in the lung fields.

vi. Asymmetric density of lungs

• If we observe that the density of right and left lungs is not symmetric (except in the case where this caused by physical deformity of the patient), you should tick YES for "Asymmetric density of lungs".





vii. Foreign substances

If we observe that foreign substance(s) are visualized in the lung fields, you should

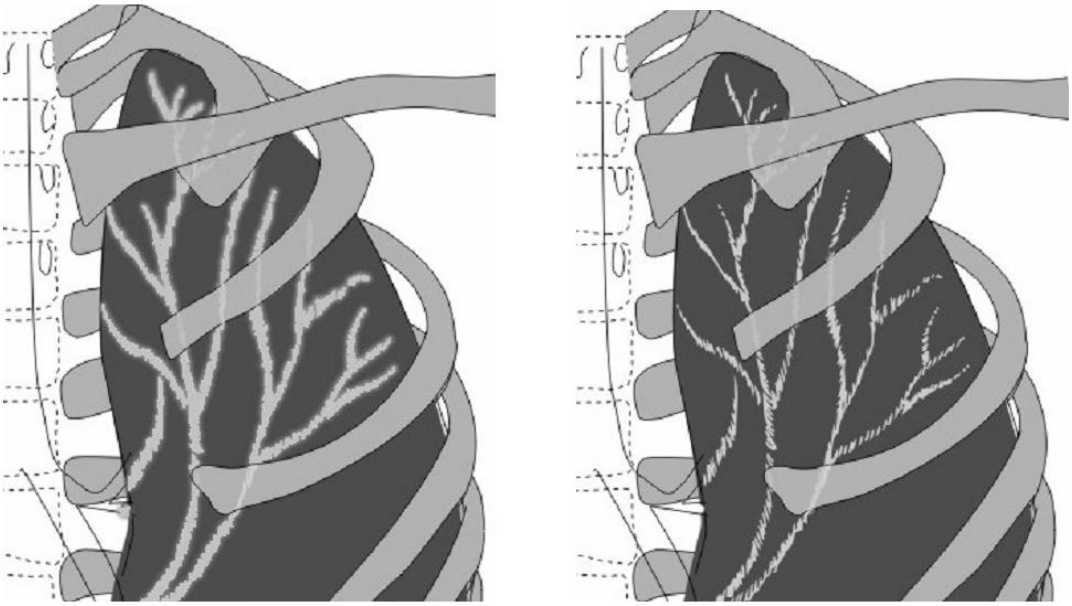
tick YES for "Foreign substances".

Careless preparation may visualize unnecessary substances on the X-ray film

3) Sharpness

- "Good": the pulmonary vessels are clearly visible in the entire left lung fields, especially around the left part of the cardiac shadow.
- "Fair" : the pulmonary vesse ls are clearly visi ble in the upper left lung fields, but obscure around the left part of the cardiac shadow.
- "Poor" : the pulmonary vessels are obscu re in the entire left lung fields

Sharpness is the appearance of the edge between 2 density areas in the imaging subject. The inadequate blurring of edge s in the imaging subject on the X-ray film makes it difficult to differentiate the structure of the subject. Figure 3.1 and Figure 3.2 show the samples visualizing the structure of inad equate and adequate sharpness.



"Poor" sharpness (left) and "Good" sharpness of pulmonary vessel (right)

The quality of sharpness around the left lower part of the card iac shadow should be carefully assessed because the lung surrounding the heart area moves by heart stroke motion (Figure 3.3) . Shorter exposure time (less th an 0.05 seconds) is recommended to produce better sharpness of the chest radiograph.

"Poor" sharpness in the imaging of pulmonary vessels

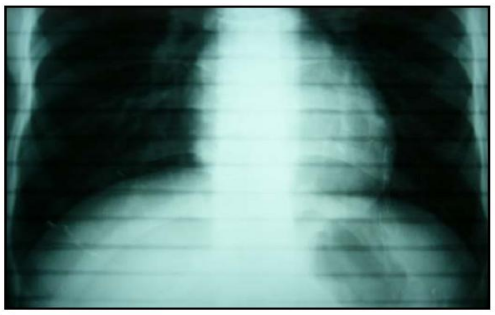
4) Artifacts

- "None" : no artifacts in the lung fields.

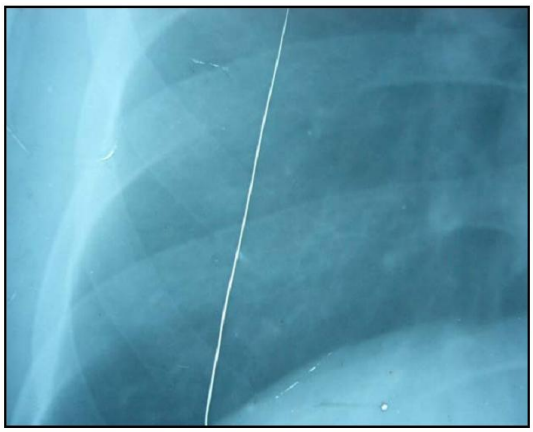
- "Slight" : small and slight artifact(s) in the lung fields, to some degree, but in which your diagnosis is not disturbed.
- "Present": clear and wide artifact(s) in the lung fields to the degree to which your diagnosis is disturbed.

Artifacts are unwanted imaging mainly caused by mishandling of X-ray film, dirty stains on the intensifying screen, improper processing of X -ray film, etc. Figure 3.4 - Figure show some examples of poor imaging quality by unacceptable artifacts on the chest radiograph.

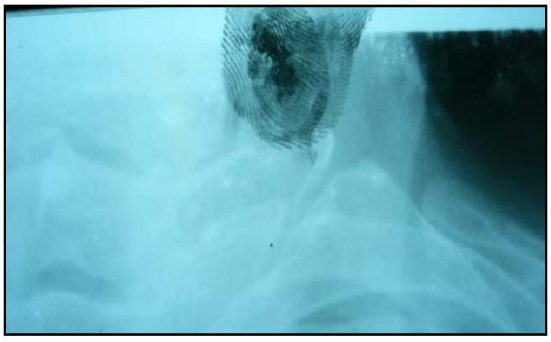
Improper processing of X-ray film by automatic processor



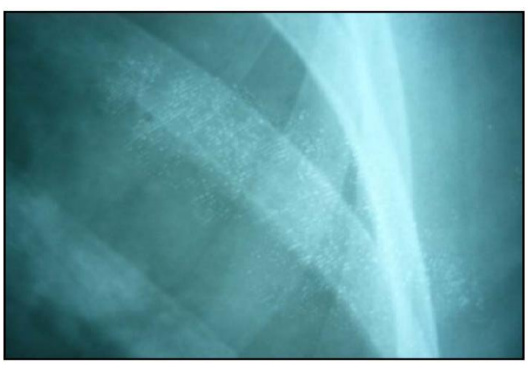
Scratch marks



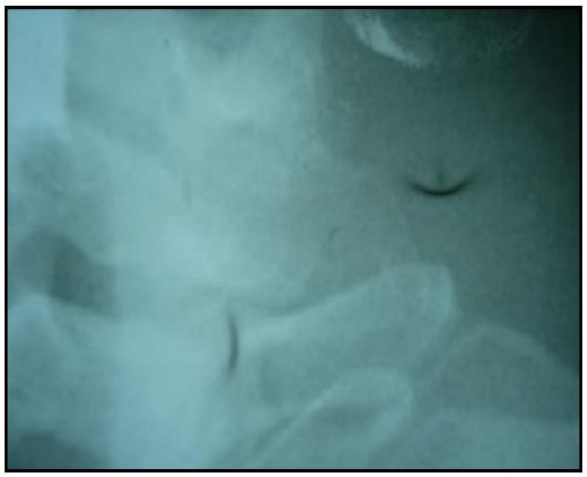
Handling by dirty fingers



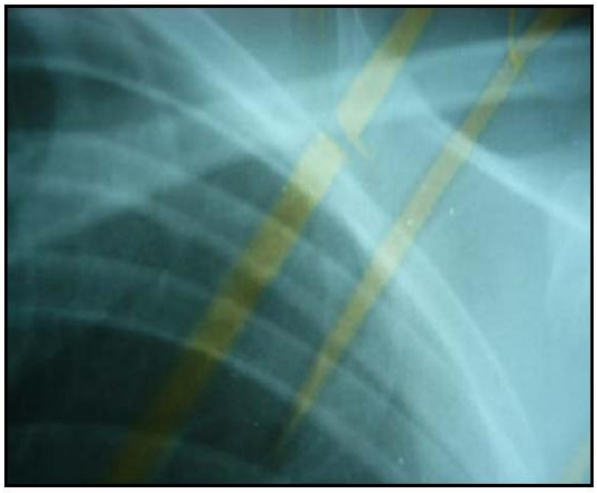
Handling with dirty fingers



Mishandling of X-ray film



Improper processing of X-ray film



Overall assessment result and actions to be taken

Each factor is to be scored using the following scoring system:

1="Good/None", 2="Fair/Slight ", and 3="Poor/Present".

The overall assessment of chest radiograph can be decided by the total score for the

Five factors, which ranges from 5 to 15 points.

- If the total score is 5or 6, you should grad e it as "Excellent".
- If the total score is 8-11 without any "Poor/Present", you should grade it as "Good".
- If the total score is 8-13 with 2 "Poor/Present" or less, you should grade it as "Fair".
- And if the total score is 14 or more, or 3 "Poor/Present" or more, you should grade it as "Poor".

Doctor 1 evaluation image in digital

num	Image num	Identification marking	Patient positioning	Sharpness	Artifacts	Ability for diagnosis	score
1	2011-01-14026	1	1	1	2	1	6
2	2011-01-14029	1	1	1	1	1	5
3	2011-01-14032	1	1	1	2	1	6
4	2011-01-14034	1	1	1	1	1	5
5	2011-01-14039	1	1	1	1	1	5
6	2011-01-14043	1	1	1	1	1	5
7	2011-01-14049	1	1	1	1	1	5
8	2011-01-14053	1	1	1	1	1	5
9	2011-01-14058	1	1	1	1	1	5
10	2011-01-14067	1	1	2	1	1	6
11	2011-01-14071	1	1	1	1	1	5
12	2011-01-14095	1	1	1	2	1	6
13	2011-01-14106	1	1	1	1	1	5
14	2011-01-14113	1	1	1	2	1	6
15	2011-01-14116	1	1	2	2	2	8
16	2011-01-14119	1	1	2	2	1	7
17	2011-01-14139	1	1	1	1	1	5
18	2011-01-14142	1	1	1	1	1	5
19	2011-01-14148	1	1	2	1	1	6
20	2011-01-14153	1	1	1	1	1	5
21	2011-01-14155	1	1	1	1	1	5
22	2011-01-14159	1	1	1	1	1	5
23	2011-01-14160	1	1	1	1	1	5

24	2011-01-14164	1	1	1	1	1	5
25	2011-01-14166	1	1	1	1	1	5
26	2011-01-14168	1	1	2	1	1	6
27	2011-01-14170	1	1	1	2	1	6
28	2011-01-14173	1	1	1	1	1	5
29	2011-01-14174	1	1	1	1	1	5
30	2011-01-14180	1	1	1	2	1	6
31	2011-01-14182	1	1	2	2	2	8
32	2011-01-14189	1	1	1	2	1	6
33	2011-01-14202	1	2	2	1	1	7
34	2011-01-14206	1	1	1	1	1	5
35	2011-01-14209	1	1	1	1	1	5
36	2011-01-14216	1	1	2	1	1	6
37	2011-01-14224	1	1	1	1	1	5
38	2011-01-14261	1	1	2	2	1	7
39	2011-01-14265	1	1	1	1	1	5
40	2011-01-14270	1	1	1	1	1	5
41	2011-01-14274	1	1	2	2	1	7
42	2011-01-14285	1	1	1	1	1	5
43	2011-01-14288	1	1	3	2	2	9
44	2011-01-14304	1	1	1	1	1	5
45	2011-01-14342	1	1	1	1	1	5
46	2011-01-14352	1	1	1	1	1	5
47	2011-01-14356	1	1	1	2	1	6
48	2011-01-14364	1	1	2	1	1	6
49	2011-01-14371	1	1	1	1	1	5
50	2011-01-14395	1	1	1	1	1	5

Doctor 2 evaluation image in digital

num	Image num	Identification marking	Patient positioning	Sharpness	Artifacts	Ability for diagnosis	score
1	2011-01-14026	1	1	1	1	1	5
2	2011-01-14029	1	1	1	1	1	5
3	2011-01-14032	1	1	1	1	1	5
4	2011-01-14034	1	1	1	1	1	5
5	2011-01-14039	1	1	1	1	1	5
6	2011-01-14043	1	1	1	1	1	5
7	2011-01-14049	1	1	1	1	1	5
8	2011-01-14053	1	1	1	1	1	5
9	2011-01-14058	1	1	1	1	1	5
10	2011-01-14067	1	1	2	1	1	6
11	2011-01-14071	1	1	1	1	1	5
12	2011-01-14095	1	1	1	2	1	6
13	2011-01-14106	1	1	1	1	1	5
14	2011-01-14113	1	1	1	2	1	6
15	2011-01-14116	1	1	2	2	2	8
16	2011-01-14119	1	1	2	2	1	7
17	2011-01-14139	1	1	1	1	1	5
18	2011-01-14142	1	1	1	1	1	5
19	2011-01-14148	1	1	2	1	1	6
20	2011-01-14153	1	1	1	1	1	5
21	2011-01-14155	1	1	1	1	1	5
22	2011-01-14159	1	1	1	1	1	5

23	2011-01-14160	1	1	1	1	1	5
24	2011-01-14164	1	1	1	1	1	5
25	2011-01-14166	1	1	1	1	1	5
26	2011-01-14168	1	1	2	1	1	6
27	2011-01-14170	1	1	1	2	1	6
28	2011-01-14173	1	1	1	1	1	5
29	2011-01-14174	1	1	1	1	1	5
30	2011-01-14180	1	1	1	2	1	6
31	2011-01-14182	1	1	1	2	2	7
32	2011-01-14189	1	1	1	2	1	6
33	2011-01-14202	1	2	2	1	1	7
34	2011-01-14206	1	1	1	1	1	5
35	2011-01-14209	1	1	1	1	1	5
36	2011-01-14216	1	1	2	1	1	6
37	2011-01-14224	1	1	1	1	1	5
38	2011-01-14261	1	1	2	2	1	7
39	2011-01-14265	1	1	1	1	1	5
40	2011-01-14270	1	1	1	1	1	5
41	2011-01-14274	1	1	2	2	1	7
42	2011-01-14285	1	1	1	1	1	5
43	2011-01-14288	1	1	2	2	2	8
44	2011-01-14304	1	1	1	1	1	5
45	2011-01-14342	1	1	1	1	1	5
46	2011-01-14352	1	1	1	1	1	5
47	2011-01-14356	1	1	1	2	1	6
48	2011-01-14364	1	1	1	1	1	5
49	2011-01-14371	1	1	1	1	1	5
50	2011-01-14395	1	1	1	1	1	5

Doctor 3 evaluation image in digital

num	Image num	Identification marking	Patient positioning	Sharpness	Artifacts	Ability for diagnosis	score
1	2011-01-14026	1	1	1	2	1	6
2	2011-01-14029	1	1	1	1	1	5
3	2011-01-14032	1	1	1	2	1	6
4	2011-01-14034	1	1	1	1	1	5
5	2011-01-14039	1	1	1	1	1	5
6	2011-01-14043	1	1	1	1	1	5
7	2011-01-14049	1	1	1	1	1	5
8	2011-01-14053	1	1	1	1	1	5
9	2011-01-14058	1	1	1	1	1	5
10	2011-01-14067	1	1	2	1	1	6
11	2011-01-14071	1	1	1	1	1	5
12	2011-01-14095	1	1	1	2	1	6
13	2011-01-14106	1	1	1	1	1	5
14	2011-01-14113	1	1	1	2	1	6
15	2011-01-14116	1	1	1	2	2	7
16	2011-01-14119	1	1	1	2	1	6
17	2011-01-14139	1	1	1	1	1	5
18	2011-01-14142	1	1	1	1	1	5
19	2011-01-14148	1	1	2	1	1	6
20	2011-01-14153	1	1	1	1	1	5
21	2011-01-14155	1	1	1	1	1	5
22	2011-01-14159	1	1	1	1	1	5

23	2011-01-14160	1	1	1	1	1	5
24	2011-01-14164	1	1	1	1	1	5
25	2011-01-14166	1	1	1	1	1	5
26	2011-01-14168	1	1	2	1	1	6
27	2011-01-14170	1	1	1	2	1	6
28	2011-01-14173	1	1	1	1	1	5
29	2011-01-14174	1	1	1	1	1	5
30	2011-01-14180	1	1	1	2	1	6
31	2011-01-14182	1	1	2	2	2	8
32	2011-01-14189	1	1	1	2	1	6
33	2011-01-14202	1	2	2	1	1	7
34	2011-01-14206	1	1	1	1	1	5
35	2011-01-14209	1	1	1	1	1	5
36	2011-01-14216	1	1	2	1	1	6
37	2011-01-14224	1	1	1	1	1	5
38	2011-01-14261	1	1	2	1	1	6
39	2011-01-14265	1	1	1	1	1	5
40	2011-01-14270	1	1	1	1	1	5
41	2011-01-14274	1	1	2	2	1	7
42	2011-01-14285	1	1	1	1	1	5
43	2011-01-14288	1	1	3	2	2	9
44	2011-01-14304	1	1	1	1	1	5
45	2011-01-14342	1	1	1	1	1	5
46	2011-01-14352	1	1	1	1	1	5
47	2011-01-14356	1	1	1	2	1	6
48	2011-01-14364	1	1	1	1	1	5
49	2011-01-14371	1	1	1	1	1	5
50	2011-01-14395	1	1	1	1	1	5

Doctor 1 evaluation image in conventional

num	Image num	Identification marking	Patient positioning	Sharpness	Artifacts	Ability for diagnosis	score
1	46	1	2	2	1	2	8
2	47	1	1	2	1	1	6
3	48	1	2	2	2	1	8
4	49	1	2	2	1	2	8
5	50	1	1	1	1	1	5
6	51	1	1	2	1	1	6
7	52	1	1	1	1	1	5
8	53	1	2	2	1	1	7
9	54	1	2	3	1	3	10
10	55	1	3	3	1	2	10
11	56	1	2	2	2	2	9
12	57	1	1	1	1	1	5
13	58	1	1	1	1	1	5
14	59	1	1	1	2	1	6
15	60	1	2	1	1	1	6
16	61	1	2	1	2	1	7
17	62	1	2	1	1	1	6
18	63	1	3	1	2	1	8
19	64	1	2	1	1	1	6
20	65	1	2	1	2	1	7
21	66	1	1	1	1	1	5
22	67	1	2	1	2	1	7

Doctors image score

Image evaluation score

23	68	1	2	1	2	1	7
24	69	1	2	1	1	1	6
25	70	1	2	2	2	1	8
26	71	1	1	1	1	1	5
27	72	1	1	1	2	1	6
28	73	1	2	1	1	1	6
29	74	1	2	1	1	1	6
30	75	1	2	1	2	1	7
31	76	1	2	2	2	1	8
32	77	1	1	1	1	1	5
33	78	1	2	1	1	1	6
34	79	1	2	1	2	1	7
35	80	1	1	1	1	1	5
36	81	1	1	1	1	1	5
37	82	1	2	1	1	1	6
38	83	1	1	1	1	1	5
39	84	1	1	1	1	1	5
40	85	1	1	1	1	1	5
41	86	1	2	2	1	1	7
42	87	1	2	2	1	1	7
43	88	1	1	1	1	1	5
44	89	1	2	1	1	1	6
45	90	1	1	1	1	1	5
46	91	1	2	2	3	2	10
47	92	1	1	1	1	1	5
48	93	1	1	1	1	1	5
49	94	1	1	1	1	1	5
50	95	1	1	1	1	1	5

Doctor 2 evaluation image in conventional

num	Image num	Identification marking	Patient positioning	Sharpness	Artifacts	Ability for diagnosis	score
1	46	1	2	2	1	2	8
2	47	1	1	2	1	1	6
3	48	1	2	2	2	1	8
4	49	1	1	2	1	2	7
5	50	1	2	1	1	1	6
6	51	1	1	2	1	1	6
7	52	1	2	1	1	1	6
8	53	1	2	2	1	1	7
9	54	1	2	3	1	3	10
10	55	1	3	3	1	2	10
11	56	1	2	2	2	2	9
12	57	1	1	1	1	1	5
13	58	1	1	1	1	1	5
14	59	1	1	1	2	1	6
15	60	1	2	1	1	1	6
16	61	1	2	1	2	1	7
17	62	1	2	1	1	1	6
18	63	1	3	2	2	1	9
19	64	1	2	1	1	1	6
20	65	1	2	1	2	1	7
21	66	1	1	1	1	1	5
22	67	1	2	1	2	1	7

Doctors image score | Image evaluation score

23	68	1	1	1	2	1	6
24	69	1	2	1	1	1	6
25	70	1	2	2	2	1	8
26	71	1	1	1	1	1	5
27	72	1	1	1	2	1	6
28	73	1	2	1	1	1	6
29	74	1	2	1	1	1	6
30	75	1	1	1	2	1	6
31	76	1	2	2	2	1	8
32	77	1	1	1	1	1	5
33	78	1	2	1	1	1	6
34	79	1	2	1	2	1	7
35	80	1	1	1	1	1	5
36	81	1	2	1	1	1	6
37	82	1	2	1	1	1	6
38	83	1	1	1	1	1	5
39	84	1	1	1	1	1	5
40	85	1	1	1	1	1	5
41	86	1	2	2	1	1	7
42	87	1	2	2	1	1	7
43	88	1	1	1	1	1	5
44	89	1	2	1	1	1	6
45	90	1	2	1	1	1	6
46	91	1	2	2	3	2	10
47	92	1	1	1	1	1	5
48	93	1	2	1	1	1	6
49	94	1	1	1	1	1	5
50	95	1	1	1	1	1	5

Doctor 2 evaluation image in conventional

num	Image num	Identification marking	Patient positioning	Sharpness	Artifacts	Ability for diagnosis	score
1	46	1	2	2	1	2	8
2	47	1	1	2	1	1	6
3	48	1	2	2	2	1	8
4	49	1	2	2	1	2	8
5	50	1	1	1	1	1	5
6	51	1	1	2	1	1	6
7	52	1	2	1	1	1	6
8	53	1	2	2	1	1	7
9	54	1	2	3	1	3	10
10	55	1	1	3	1	2	8
11	56	1	2	2	2	2	9
12	57	1	1	1	1	1	5
13	58	1	1	1	1	1	5
14	59	1	1	1	2	1	6
15	60	1	2	1	1	1	6
16	61	1	2	1	2	1	7
17	62	1	3	1	1	1	7
18	63	1	3	1	2	1	8
19	64	1	2	1	1	1	6
20	65	1	2	1	2	1	7
21	66	1	1	1	1	1	5
22	67	1	2	1	2	1	7

Doctors image score

Image evaluation score

23	68	1	2	1	2	1	7
24	69	1	2	1	1	1	6
25	70	1	2	2	2	1	8
26	71	1	1	1	1	1	5
27	72	1	2	1	2	1	7
28	73	1	2	1	1	1	6
29	74	1	2	2	1	1	7
30	75	1	2	1	2	1	7
31	76	1	2	2	2	1	8
32	77	1	1	1	1	1	5
33	78	1	2	1	1	1	6
34	79	1	2	1	2	1	7
35	80	1	1	1	1	1	5
36	81	1	1	1	1	1	5
37	82	1	2	2	1	1	7
38	83	1	1	1	1	1	5
39	84	1	1	1	1	1	5
40	85	1	1	1	1	1	5
41	86	1	2	2	1	1	7
42	87	1	2	2	1	1	7
43	88	1	1	1	1	1	5
44	89	1	2	1	1	1	6
45	90	1	1	1	1	1	5
46	91	1	2	2	3	2	10
47	92	1	1	1	1	1	5
48	93	1	1	2	1	1	6
49	94	1	1	1	1	1	5
50	95	1	1	2	1	1	6

Doctors image score

Image evaluation score

Digital x-ray system data

num	Image number	KV	mAS	Wight Kg	Height cm	BMI	Air Kerma mR	Air Kerma mGy	Effective Dose mSV
1	2011-01-14026	85	16	60	165	22.03857	235.6	2.056788	0.357
2	2011-01-14029	90	16	64	156	26.29849	250.4	2.185992	0.385
3	2011-01-14032	88	16	67	164	24.91077	242.6	2.117898	0.354
4	2011-01-14034	88	16	87	173	29.0688	242.6	2.117898	0.308
5	2011-01-14039	85	16	52	152	22.50693	235.6	2.056788	0.348
6	2011-01-14043	90	16	63	158	25.23634	250.4	2.185992	0.368
7	2011-01-14049	90	16	73	163	27.47563	250.4	2.185992	0.4
8	2011-01-14053	95	16	80	156	32.87311	268.8	2.346624	0.349
9	2011-01-14058	90	16	84	167	30.1194	250.4	2.185992	0.324
10	2011-01-14067	95	16	95	163	35.75596	268.8	2.346624	0.323
11	2011-01-14071	110	16	99	161	38.19297	344	3.00312	0.45
12	2011-01-14095	80	16	66	157	26.77593	208.8	1.822824	0.259
13	2011-01-14106	77	16	47	164	17.47472	195.1	1.703223	0.301
14	2011-01-14113	80	16	55	163	20.70082	208.8	1.822824	0.306
15	2011-01-14116	92	16	93	161	35.87825	256.4	2.238372	0.301
16	2011-01-14119	83	16	64	161	24.69041	227.7	1.987821	0.311
17	2011-01-14139	110	16	109	165	40.03673	344	3.00312	0.435
18	2011-01-14142	78	16	48	157	19.47341	202.1	1.764333	0.29
19	2011-01-14148	80	16	63	158	25.23634	208.8	1.822824	0.269
20	2011-01-14153	80	16	66	161	25.46198	208.8	1.822824	0.266
21	2011-01-14155	78	16	64	157	25.96454	202.1	1.764333	0.248
22	2011-01-14159	77	16	44	157	17.85062	195.1	1.703223	0.294
23	2011-01-14160	110	16	105	163	39.51974	344	3.00312	0.444
24	2011-01-14164	95	16	92	158	36.85307	268.8	2.346624	0.321
25	2011-01-14166	95	16	55	150	24.44444	268.8	2.346624	0.39

num	Image number	KV	mAS	Wight Kg	Height cm	BMI	Air Kerma mR	Air Kerma mGy	Effective Dose mSV
26	2011-01-14168	90	16	65	166	23.58833	250.4	2.185992	0.381
27	2011-01-14170	80	16	51	155	21.22789	208.8	1.822824	0.299
28	2011-01-14173	77	16	46	155	19.14672	195.1	1.703223	0.263
29	2011-01-14174	90	16	65	167	23.30668	250.4	2.185992	0.386
30	2011-01-14180	100	16	94	165	34.52709	291.2	2.542176	0.382
31	2011-01-14182	79	16	50	155	20.81165	202.1	1.764333	0.288
32	2011-01-14189	79	16	52	156	21.36752	202.1	1.764333	0.282
33	2011-01-14202	80	16	50	146	23.45656	208.8	1.822824	0.257
34	2011-01-14206	83	16	58	167	20.79673	227.7	1.987821	0.347
35	2011-01-14209	105	16	81	158	32.44672	308.6	2.694078	0.45
36	2011-01-14216	85	16	58	165	21.30395	235.6	2.056788	0.356
37	2011-01-14224	88	16	70	168	24.80159	242.6	2.117898	0.35
38	2011-01-14261	90	16	73	156	29.99671	250.4	2.185992	0.327
39	2011-01-14265	88	16	73	152	31.59626	242.6	2.117898	0.265
40	2011-01-14270	88	16	73	156	29.99671	242.6	2.117898	0.309
41	2011-01-14274	83	16	63	159	24.9199	227.7	1.987821	0.308
42	2011-01-14285	110	16	103	150	45.77778	344	3.00312	0.397
43	2011-01-14288	80	16	59	157	23.93606	208.8	1.822824	0.279
44	2011-01-14304	80	16	75	165	27.54821	208.8	1.822824	0.251
45	2011-01-14342	78	16	52	157	21.09619	202.1	1.764333	0.282
46	2011-01-14352	85	16	65	158	26.03749	235.6	2.056788	0.391
47	2011-01-14356	75	16	51	165	18.73278	235.6	2.056788	0.338
48	2011-01-14364	90	16	81	166	29.39469	250.4	2.185992	0.33
49	2011-01-14371	78	16	63	166	22.86253	202.1	1.764333	0.267
50	2011-01-14395	78	16	49	157	19.8791	202.1	1.764333	0.291
51	2011-01-14427	78	16	71	152	30.73061	202.1	1.764333	0.219

Film /screen x-ray

num	Image_N	KVCon	MAS	Wight	Height	BMI	Air Kerma mR	Air Kerma mGy	Effective Dose mSV
1	46	110	16	90	182	27.17063157	344	3.00312	0.56902
2	47	90	16	60	173	20.04744562	250.4	2.185992	0.429
3	48	90	16	61	172	20.61925365	250.4	2.185992	0.42
4	49	88	16	55	160	21.484375	242.6	2.117898	0.39
5	50	100	16	77	165	28.28282828	291.2	2.542176	0.45
6	51	95	16	72	170	24.91349481	268.8	2.346624	0.427
7	52	100	16	75	174	24.77209671	291.2	2.542176	0.48
8	53	85	16	55	174	18.16620425	235.6	2.056788	0.397
9	54	105	16	82	173	27.39817568	308.6	2.694078	0.5
10	55	90	16	60	170	20.76124567	250.4	2.185992	0.421
11	56	80	16	47	150	20.88888889	208.8	1.822824	0.3
12	57	77	16	45	160	17.578125	195.1	1.703223	0.29
13	58	83	16	50	155	20.81165453	227.7	1.987821	0.34
14	59	85	16	55	145	26.15933413	235.6	2.056788	0.289
15	60	85	16	55	150	24.44444444	235.6	2.056788	0.305
16	61	88	16	59	135	32.37311385	242.6	2.117898	0.42
17	62	85	16	55	155	22.89281998	235.6	2.056788	0.345
18	63	88	16	59	160	23.046875	242.6	2.117898	0.37
19	64	80	16	47	153	20.07774787	208.8	1.822824	0.285
20	65	80	16	47	165	17.26354454	208.8	1.822824	0.338
21	66	88	16	58	150	25.77777778	242.6	2.117898	0.43
22	67	85	16	56	155	23.30905307	235.6	2.056788	0.398
23	68	80	16	47	150	20.88888889	208.8	1.822824	0.355
24	69	95	16	72	153	30.757401	268.8	2.346624	0.42
25	70	80	16	47	170	16.26297578	208.8	1.822824	0.345

num	Image_N	KVCon	MAS	Wight	Height	BMI	Air Kerma mR	Air Kerma mGy	Effective Dose mSV
26	71	95	16	67	155	27.88761707	268.8	2.346624	0.444
27	72	90	16	63	168	22.32142857	250.4	2.185992	0.401
28	73	85	16	55	171	18.80920625	235.6	2.056788	0.389
29	74	80	16	47	160	18.359375	208.8	1.822824	0.326
30	75	80	16	49	160	19.140625	208.8	1.822824	0.319
31	76	85	16	55	162	20.95717116	235.6	2.056788	0.367
32	77	80	16	49	164	18.21832243	208.8	1.822824	0.338
33	78	88	16	59	160	23.046875	242.6	2.117898	0.37
34	79	85	16	55	172	18.59113034	235.6	2.056788	0.391
35	80	90	16	59	168	20.90419501	250.4	2.185992	0.415
36	81	85	16	55	172	18.59113034	235.6	2.056788	0.391
37	82	83	16	52	168	18.42403628	227.7	1.987821	0.374
38	83	85	16	55	168	19.48696145	235.6	2.056788	0.381
39	84	85	16	56	156	23.01117686	235.6	2.056788	0.347
40	85	88	16	59	156	24.24391847	242.6	2.117898	0.378
41	86	90	16	65	164	24.1671624	250.4	2.185992	0.38
42	87	83	16	53	161	20.44674202	227.7	1.987821	0.351
43	88	77	16	45	168	15.94387755	195.1	1.703223	0.299
44	89	90	16	58	150	25.77777778	250.4	2.185992	0.463
45	90	85	16	55	152	23.80540166	235.6	2.056788	0.439
46	91	80	16	47	155	19.56295525	208.8	1.822824	0.352
47	92	90	16	62	158	24.8357635	250.4	2.185992	0.428
48	93	80	16	51	164	18.96192742	208.8	1.822824	0.363
49	94	85	16	55	162	20.95717116	235.6	2.056788	0.416
50	95	88	16	58	156	23.8330046	242.6	2.117898	0.417
51	96	80	16	48	157	19.47340663	208.8	1.822824	0.355

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Key words:PCXMC, x-ray diagnostics, patient dose, effective dose, organ dose,

Monte Carlo method, risk assessment, computer program

Summary

PCXMC is a Monte Carlo program for calculating patients' organ doses and effective doses in medical x-ray examinations. The organs and tissues considered in the program are: active bone marrow, adrenals, brain, breasts, colon (upper and lower large intestine), extrathoracic airways, gall bladder, heart, kidneys, liver, lungs, lymph nodes, muscle, oesophagus, oral mucosa, ovaries, pancreas, prostate, salivary glands, skeleton, skin, small intestine, spleen, stomach, testicles, thymus, thyroid, urinary bladder and uterus.

The program calculates the effective dose with both the present tissue weighting factors of ICRP Publication 103 (2007) and the old tissue weighting factors of ICRP Publication 60 (1991). The anatomical data are based on the mathematical hermaphrodite phantom models of Cristy and Eckerman (1987), which describe patients of six different ages: new-born, 1, 5, 10, 15-year-old and adult patients. Some changes are made to these phantoms in order to make them more realistic for external irradiation conditions and to enable the calculation of the effective dose according to the new ICRP Publication 103 tissue weighting factors. The phantom sizes are adjustable to mimic patients of an arbitrary weight and height. PCXMC allows a free adjustment of the x-ray beam projection and other examination conditions of projection radiography and fluoroscopy. All organ doses calculated by PCXMC are relative to the incident air kerma, Ka,i This quantity represents the air kerma at the point where the central axis of the x-ray beam enters the patient. It is given in units of milligray (mGy), free-in-air, without backscatter; see ICRU 74 (2005). The user must supply this datum to the program. The amount of radiation can also be input as the entrance exposure (mR, free-in-air, without backscatter), air kerma-area product or dose-area product ($mGy \cdot cm^2$), or exposure-area product ($R \cdot cm^2$). If radiation measurements are not available, the program is able to estimate the incident air kerma also from an input of the x-ray tube current-time product (mAs).

The dose calculation method in PCXMC is the Monte Carlo method. The Monte Carlo calculation of photon transport is based on stochastic mathematical simulation of interactions between photons and matter. Photons are emitted (in a fictitious mathematical sense) from a point source into the solid angle specified by the focal distance and the x-ray field dimensions, and followed while they randomly interact with the phantom according to the probability distributions of the physical processes that they may undergo: photo-electric absorption, coherent (Rayleigh) scattering or incoherent (Compton) scattering. At each interaction point the energy deposition to the organ at that position is calculated and stored for dose calculation. Other interactions are not considered in PCXMC because the maximum photon energy is limited to 150 keV. This chain of interactions forms a so-called history of an individual photon. A large number of independent photon histories is generated, and estimates of the mean values of energy depositions in the various organs of the phantom are used for calculating the doses in these organs.

Calculated organ doses can be used for the assessment of the risk of exposure induced cancer. The risk estimates are based on the combined absolute and relative risk models of BEIR VII committee (BEIR 2006). PCXMC calculates the risk of exposure-induced death for leukaemia, cancers in colon, stomach, lung, urinary bladder, prostate, uterus, ovaries, breast, liver, thyroid and for all other solid cancers combined. The user may use the risk calculation module for estimating the cancer risk resulting from a single exposure or multiple exposures simulated in PCXMC. The user may also edit the organ doses without calculating doses with PCXMC: a radiation risk assessment can be made for arbitrary irradiation cases.

The present version (2.0) of the program runs in a PC under Windows 95/98/NT/2000/XP/Vista. The Monte Carlo simulation time depends on the desired accuracy and on the speed of the PC, but is typically less than a minute in a PC with a 1.8 GHz processor. The same Monte Carlo data can be used for calculating doses for any x-ray spectrum of interest when the other conditions of the examination remain unchanged; in this case the calculation time is very short because no further Monte Carlo simulations are needed.

The data calculated by PCXMC have been earlier compared to the organ dose conversion factors calculated in NRPB by Jones and Wall (1985) and Hart et al. (1994, 1996) and were found to agree well (Tapiovaara et al. 1997). The excellent agreement with the NRPB data

still exists for most irradiation conditions. Small differences are noted in some irradiation conditions, because the composition and density of the phantom tissues have been changed and the phantoms have been modified from the earlier versions of the program. Reasonable agreement of PCXMC results has also been found in many comparisons with dose measurements and calculations with other phantom models, e.g., Schmidt et al. (2000), Schultz et al. (2003), Helmrot et al. (2007). The usability of the phantom size modification feature of PCXMC has been demonstrated in an extreme case by Smans et al. (2008) who calculated doses in two premature babies with weights of 590 g and 1910 g. The differences between the doses calculated with their voxel phantoms and PCXMC were explained by the differences in the phantom models and the difficulty to place an x-ray field in them in an equivalent fashion. Similar differences between computational and voxel phantoms have also been seen in the papers of Staton et al. (2003), Lee et al. (2006) and Pazik et al. (2007); dose differences of the same order are obtained also in dose calculations with different voxel phantoms (Zankl et al. 2002, Schlattl et al. 2007).