

Chapter one

Introduction

1.1 Introduction:

X-ray discovered in 1895 by Roentgen, x-rays have played an important role in medicine. Evolution of modern medicine would have never been possible without the use of x-rays. Today, equipments that utilize x-rays are used in all aspects of diagnostic as well therapeutic radiology. Applications include: general radiography, fluoroscopy, computed tomography and mammography. Despite the discovery of alternative imaging techniques such as ultrasound and magnetic resonance imaging (MRI), that utilize non-ionizing radiation, and hence deliver no risk to patients, x-ray radiation remains the most favorable technique for reasons related to its low cost and easy operation (UNCEAR, 2000).

However, the use of x-ray radiation, even at low doses typically encountered in diagnostic radiology, is associated with the risk of cancer induction and other stochastic effects. In interventional radiology procedures radiation doses often exceed the threshold for deterministic effects. Numerous cases were reported in the literature for skin injuries during such procedures (ICRP, 2001).

In applications of ionizing radiation to problems related to medicine it is important to measure the quantity of radiation dose delivered to the patient. In diagnostic procedures such as x-ray examinations, nuclear medicine, CT scans, PET etc. such measurements are important both for the optimization of image quality and for radiation protection purposes.

The present work constitutes a part of the global effort in radiation protection of patients in diagnostic as well as interventional radiology.

Focal spot is defined as the area on the target of the x-ray tube which the electron stream strikes and from which x-rays are emitted. called also focus. The imaging diameter of star resolution pattern was measured via fixation the star resolution

pattern in out of x-ray beam and record its image on the x-ray film. From the image the focal spot size for x-ray machine under study was calculated using the following equation:

$$f = \frac{\pi \theta D}{180 (M-1)}$$

M = Imaging diameter /Original diameter

D: the internal diameter between starting and ending the resolution

$$\theta = 20$$

F = focal spot size, mm

1.2. Problem of Study:

Fine and broad focal spot sizes are available on general x-ray tubes. excessive used of fine focus can impact on the tube life and whilst it is established that fine focal spot size reduces geometric unsharpness, the extent of this benefit on clinical image quality is unclear.

1.3. Objectives of the study:

1.3.1. General Objective:

Effect of focal spot size on image quality

1.3.2. Specific Objectives:

- To determine the effect of focal spot size in image quality.
- To achieve standard image quality.
- Assessment of the sharpness of the focal spot on multi images,
- Evaluate the quality of the X-ray tomography (resolution, sharpness, brightness.

1.4. Overview of the Study:

This study falls into five chapters, Chapter one, which is an introduction, deals with theoretical frame work of the study and (Literature review). It presents the statement of the study problems, objectives of the study, chapter two deals with radiological physics and back ground. Chapter three deal with material and method, Chapter fours deals with results and discussions. Chapter five conclusions, recommendations and references.

Chapter two

Literature Review

2.1. X-ray Production:

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 generator is 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. (Bushberg JT et al , 2002).

The differential absorption of x-rays in tissues and organs, owing to their atomic composition, is the basis for the various imaging methods used in diagnostic radiology. The principles in the production of x-rays have remained the same since their discovery. However, much refinement has gone into the design of x-ray tubes to achieve the performance required for today's radiological examinations Bushberg JT et al , 2002).

2.2. Fundamentals of x-ray Production:

The production of x-rays involves the bombardment of a thick target with energetic electrons. These electrons undergo a complex sequence of collisions and scattering processes during the slowing down process, which results in the production of bremsstrahlung and characteristic radiation.

2.2.1. Bremsstrahlung:

Energetic electrons are mostly slowed down in matter by collisions and excitation interactions. If an electron comes close to an atomic nucleus, the attractive Coulomb forces cause a change of the electron's trajectory. An accelerated electron, or an electron changing its direction, emits electromagnetic radiation, given the name bremsstrahlung (braking radiation), and this energy of the emitted photon is subtracted from the kinetic energy of the electron. The energy of the bremsstrahlung photon depends on the attractive Coulomb forces and hence on the distance of the electron from the nucleus.

Using classical theory to consider the electron bombardment of a thin target yields a constant energy fluence from zero up to the initial electron kinetic energy.

A thick target can be thought of as a sandwich of many thin target layers, each producing a rectangular distribution of energy fluence. As the electron is slowed down in each layer, the maximum energy in the distribution becomes less, until the electron comes to rest. The superposition of all these rectangular distributions forms a triangular energy fluence distribution for a thick target, the 'ideal' spectrum. Indeed, this model is a simplification, as quantum mechanical theory shows that the distribution for a thin layer is not rectangular and a stepwise reduction of the electron energy from layer to layer does not conform to the slowing down characteristics of electrons.

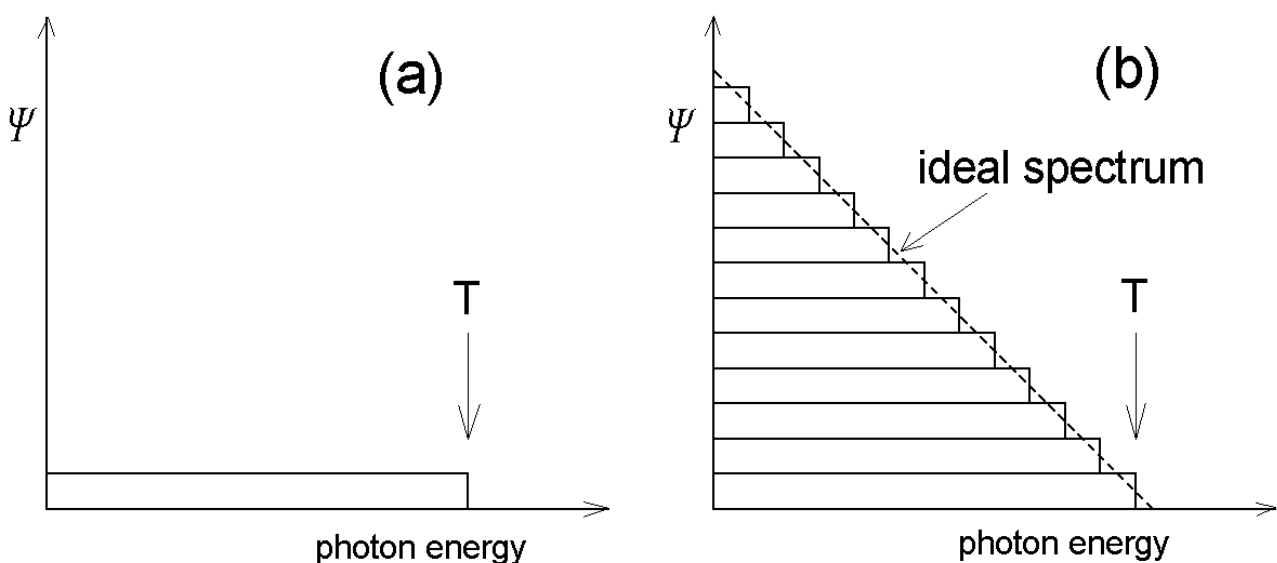


FIG 2.1 (a) Rectangular distribution of the X ray energy fluence Ψ for a thin target bombarded with electrons of kinetic energy T . (b) For a stack of thin targets, each target layer contributes a rectangular fluence distribution, assuming a uniform electron energy decrease from layer to layer. The superposition forms a triangular ideal spectrum (Bushberg JT et al , 2002).

2.2.2. Characteristic radiation:

A fast electron colliding with an electron of an atomic shell could knock out the electron, provided its kinetic energy exceeds the binding energy of the electron in that shell. The binding energy is highest in the most inner K shell and decreasing for the outer shells (L, M, etc.). The scattered primary electron carries away the difference of kinetic energy and binding energy. The vacancy in the shell is then filled with an electron from an outer shell, accompanied by the emission of an x-ray photon with an energy equivalent to the difference in binding energies of the shells involved. For each element, binding energies, and the mono energetic radiation resulting from such interactions, are unique and characteristic for that element.

K radiation denotes characteristic radiation for electron transitions to the K shell, and likewise, L radiation for transitions to the L shell. The origin of the electron filling the vacancy is indicated by suffixes (a, b, g, etc.), where a stands for a transition from the adjacent outer shell, b from the next outer shell.

Ka radiation results from L to K shell transitions; Kb radiation from M to K Shell transitions, etc. Energies are further split owing to the energy levels in a shell, indicated with a numerical suffix. Further, each vacancy in an outer shell following from such a transition gives rise to the emission of corresponding characteristic radiation causing a cascade of photons.

Table 5.1 gives the binding energies and the K radiation energies for the common anode materials used in diagnostic radiology.

Instead of characteristic radiation, the energy available could be transferred to an electron that is ejected from the shell (Auger electron). The probability of Auger electron production decreases with atomic number. (Bushberg JT et al , 2002).

Table (2.1) Binding Energies and K Radiation Energies of Common Anode Materials

| Element ^a | Binding energy (keV) | | Energies of characteristic X rays (keV) | | | |
|----------------------|----------------------|---------|---|-----------------|-----------------|-----------------|
| | L shell | K shell | K _{α1} | K _{α2} | K _{β1} | K _{β2} |
| Mo | 2.87/2.63/2.52 | 20.00 | 17.48 | 17.37 | 19.61 | 19.97 |
| Rh | 3.41/3.15/3.00 | 23.22 | 20.22 | 20.07 | 22.72 | 23.17 |
| W | 12.10/11.54/10.21 | 69.53 | 59.32 | 57.98 | 67.24 | 69.07 |

^a Mo: molybdenum; Rh: rhodium; W: tungsten.

2.3. X ray spectrum:

The electrons are slowed down and stopped in the target, within a range of a few tens of micrometres, depending on the tube voltage. As a result, X rays are not generated at the surface but within the target, resulting in an attenuation of the X ray beam. This self-filtration appears most prominent at the low energy end of the spectrum. Additionally, characteristic radiation shows up if the kinetic electron energy exceeds the binding energies. L radiation is totally absorbed by a typical filtration of 2.5 mm Al. The K edge in the photon attenuation of tungsten can be noticed as a drop of the continuum at the binding energy of 69.5 keV. For tungsten targets, the fraction of K radiation contributing to the total energy fluence is less than 10% for a 150 kV tube voltage. (Bushberg JT et al , 2002).

2.4. X-ray Tubes:

2.4.1. Components of the X-ray tube:

The production of both bremsstrahlung and characteristic radiation requires energetic electrons hitting a target. Accordingly, the principal components of an X-ray tube are an electron source from a heated tungsten filament, with a focusing cup serving as the tube cathode, an anode or target, and a tube envelope to maintain an interior vacuum. The filament is heated by a current that controls the thermionic emission of electrons, which, in turn, determines the electronic current flowing from the cathode to the anode (tube or anode current).

The accelerating potential difference applied between the cathode and anode controls both X ray energy and yield. Thus, two main circuits operate within the X-ray tubes the filament circuit and the tube voltage circuit. (Bushberg JT et al , 2002).

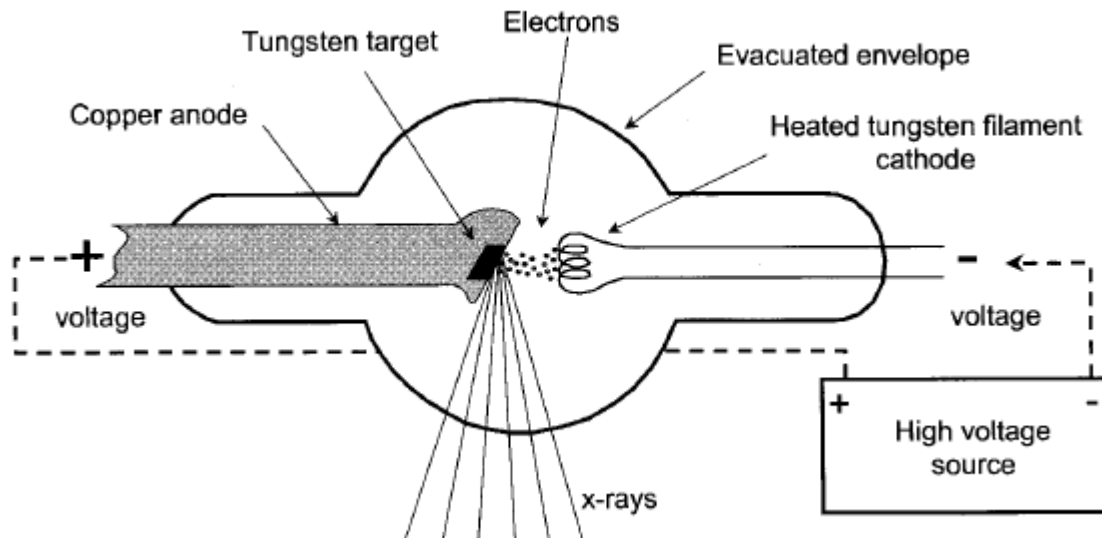


FIG. 2.2. Principal components of an X-ray tube . (Bushberg JT et al , 2002).

2.4.1.1. Cathode:

The arrangement of the filament, the focusing cup, the anode surface and the tube voltage generates an electric field accelerating the electrons towards the focal spot at the anode. X-ray tubes with two focal spots usually employ two separate filament/cup assemblies (cathode blocks).

The degree of focusing depends on the potential difference or bias voltage between the filament and focusing electrode. The focal spot will be largest if both are at the same potential. With an increasing negative bias voltage at the focusing cup, the focus size will decrease and finally the electron current will be pinched off.

This effect is sometimes used to control the focus size electronically, or for a fast switching of the anode current (grid controlled tubes) when short radiation pulses are required, as in pulsed fluoroscopy. Some bias can simply be achieved by connecting the filament and cup with a high resistance grid leak resistor. Some electrons emitted

from the filament surface will hit and charge the cup. The cup is discharged via the grid leak resistor, maintaining the cup at a negative potential difference.

2.4.1.2. Anode:

2.4.1.2.1. Choice of material:

For common radiographic applications, a high bremsstrahlung yield is mandatory, requiring materials with high atomic numbers (Z). Additionally, because of the low effectiveness of X ray production, it is also essential that the thermal properties are such that the maximum useful temperature determined by melting point, vapour pressure, heat conduction, specific heat and density is also considered. Tungsten ($Z = 74$) is the optimum choice here.

For mammography, other anode materials such as molybdenum ($Z = 42$) and rhodium ($Z = 45$) are frequently used. For such anodes, X ray spectra show less contribution by bremsstrahlung but rather dominant characteristic X rays of the anode materials. This allows a more satisfactory optimization of image quality and patient dose. In digital mammography, these advantages are less significant and some manufacturers prefer tungsten anodes.

2.4.1.2.2. Line focus principle (anode angle):

For measurement purposes, the focal spot size is defined along the central beam projection. For high anode currents, the area of the anode hit by the electrons should be as large as possible, to keep the power density within acceptable limits. To balance the need for substantial heat dissipation with that of a small focal spot size, the line focus principle is used.

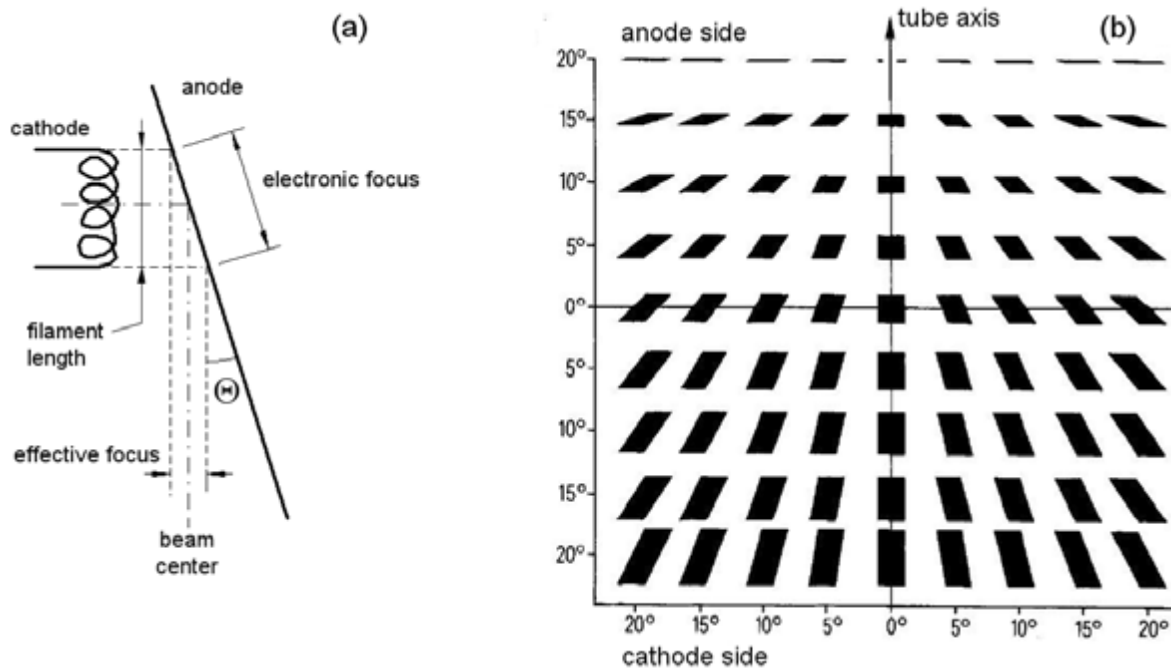


Fig 2.3. (a) Line focus principle: the length of the filament appears shortened in the beam direction; (b) graphic representation of the focal spot shape at different locations in the radiation field (anode angle 20°).

2.4.1.2.3. Anode Angle and Focal Spot Size:

The anode angle is defined as the angle of the target surface with respect to the central ray in the x-ray field. Anode angles in diagnostic x-ray tubes, other than some mammography tubes, range from 7 to 20 degrees, with 12- to 15-degree angles being most common. Focal spot size is defined in two ways. The actual focal spot size is the area on the anode that is struck by electrons, and it is primarily determined by the length of the cathode filament and the width of the focusing cup slot.

The effective focal spot size is the length and width of the focal spot as projected down the central ray in the x-ray field. The effective focal spot width is equal to the actual focal spot width and therefore is not affected by the anode angle. However, the anode angle causes the effective focal spot length to be smaller than the actual focal spot length. The effective and actual focal spot lengths are related as follows:

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

where θ is the anode angle. This foreshortening of the focal spot length,

There are three major tradeoffs to consider for the choice of anode angle. A smaller anode angle provides a smaller *effective* focal spot for the same *actual* focal area.

However, a small anode angle limits the size of the usable x-ray field owing to cutoff of the beam. Field *coverage* is less for short focus-to-detector distances.

The optimal anode angle depends on the clinical imaging application. A small anode angle (approximately 7 to 9 degrees) is desirable for small field-of-view image receptors, such as cine angiographic and neuro angiographic equipment, where field coverage is limited by the image intensifier diameter. Larger anode angles (approximately 12 to 15 degrees) are necessary for general radiographic work to achieve large field area coverage at short focal spot-to-image distances.

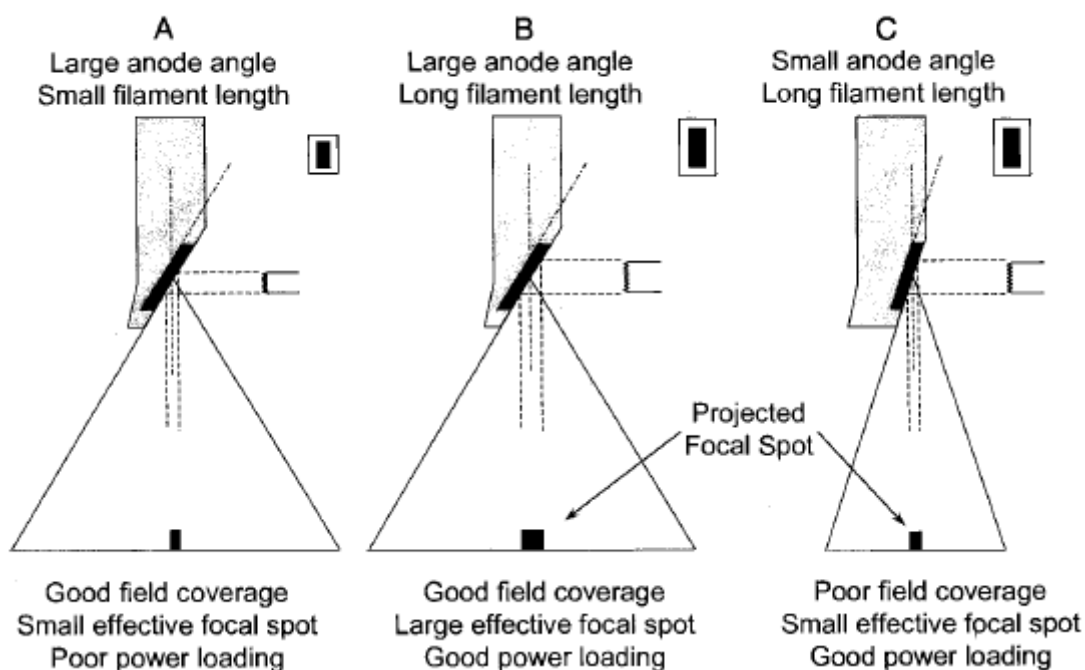


Fig 2.4 Field coverage and effective focal spot length vary with the anode angle.

A: A large anode angle provides good field coverage at a given distance; however, to achieve a small effective focal spot, a small actual focal area limits power loading.

B: A large anode angle provides good field coverage, and achievement of high power loading requires a large focal area; however, geometric blurring and image degradation occur.

C: A small anode angle limits field coverage at a given distance; however, a small effective focal spot is achieved with a large focal area for high power loading.

2.4.1.2.4. Stationary and rotating anodes:

For x ray examinations that require only a low anode current or infrequent low power exposures, an X ray tube with a stationary anode is applicable. Here, a small tungsten block serving as the target is brazed to a copper block to dissipate the heat efficiently to the surrounding cooling medium. As the focal spot is stationary, the maximum loading is determined by the anode temperature and temperature gradients.

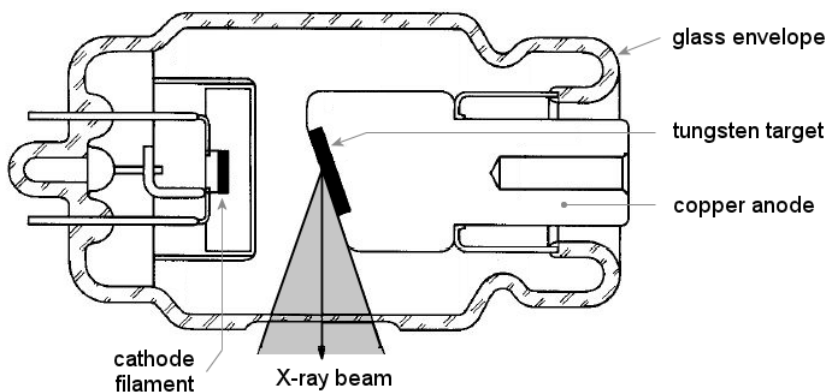


Fig 2.5. Dental X ray tube with a stationary anode (Carl A Carlsson, et al, 1996)

2.4.1.3. Energizing and Controlling the X-ray Tube:

The X ray generator provides all the electrical power sources and signals required for the operation of the X ray tube, and controls the operational conditions of X ray production and the operating sequence of exposure during an examination. The essential components are a filament heating circuit to determine anode current, a high voltage supply, a motor drive circuit for the stator windings required for a rotating anode tube, an exposure control providing the image receptor dose required, and an operational control. The operational control is often accomplished by a microprocessor system but electromechanical devices are still in use. Modern generators provide control of the anode temperature by monitoring the power applied to the tube and calculating the cooling times required according to the tube rating charts.

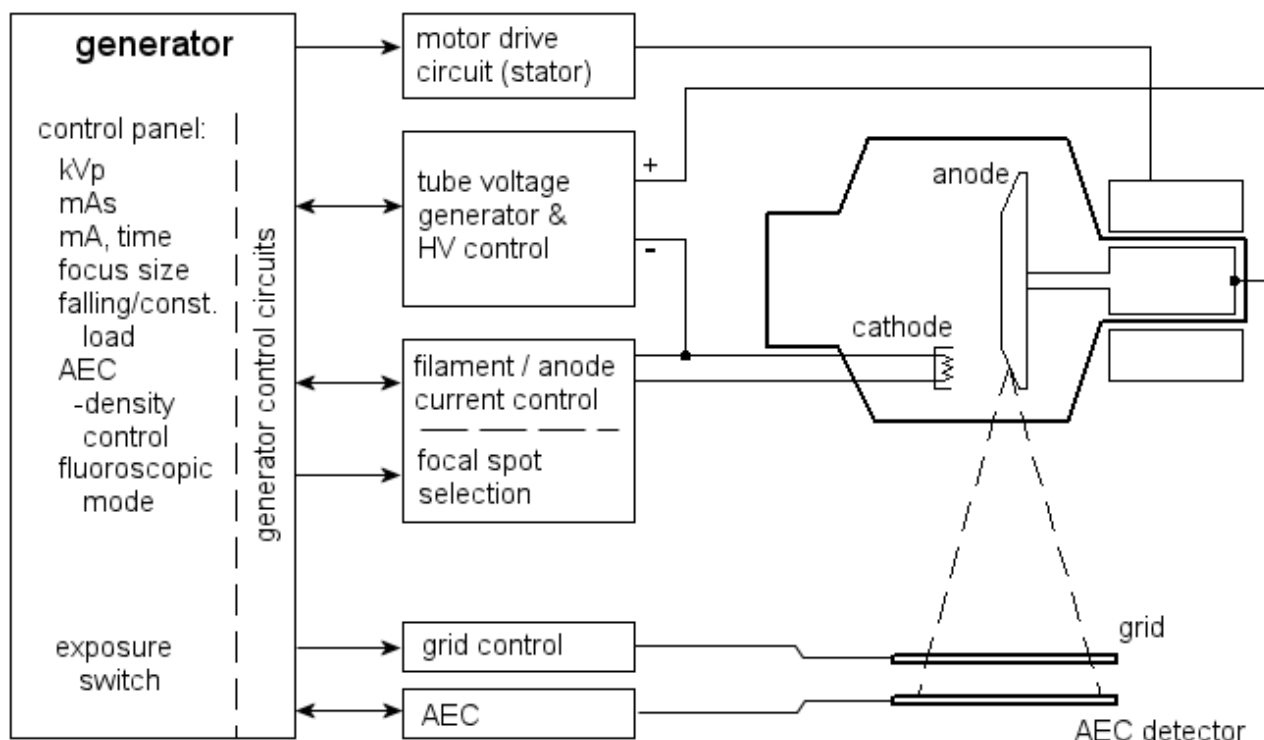


Fig 2.6. Schematic diagram of a basic X ray generator. AEC: automatic exposure control; kVp: peak voltage.

2.4.1.4. Collimation and Filtration:

2.4.1.4.1. Collimator and light field:

The limitation of the x ray field to the size required for an examination is accomplished with collimators. The benefits of collimating the beam are twofold reduction in patient dose and improvement of image contrast due to a reduction in scattered radiation. A collimator assembly is typically attached to the tube port, defining the field size with adjustable parallel opposed lead diaphragms or blades. To improve the effectiveness of collimation, another set of blades might be installed at some distance from the first blades in the collimator housing. Visualization of the x ray field is achieved by a mirror reflecting the light from a bulb. The bulb position is adjusted so that the reflected light appears to have the same origin as the focal spot of the tube. The light field then ‘mimics’ the actual X ray field. The congruency of light and X ray field is subject to quality control. One must be aware that some of the penumbra at the edges of the radiation field is due to extra focal radiation.

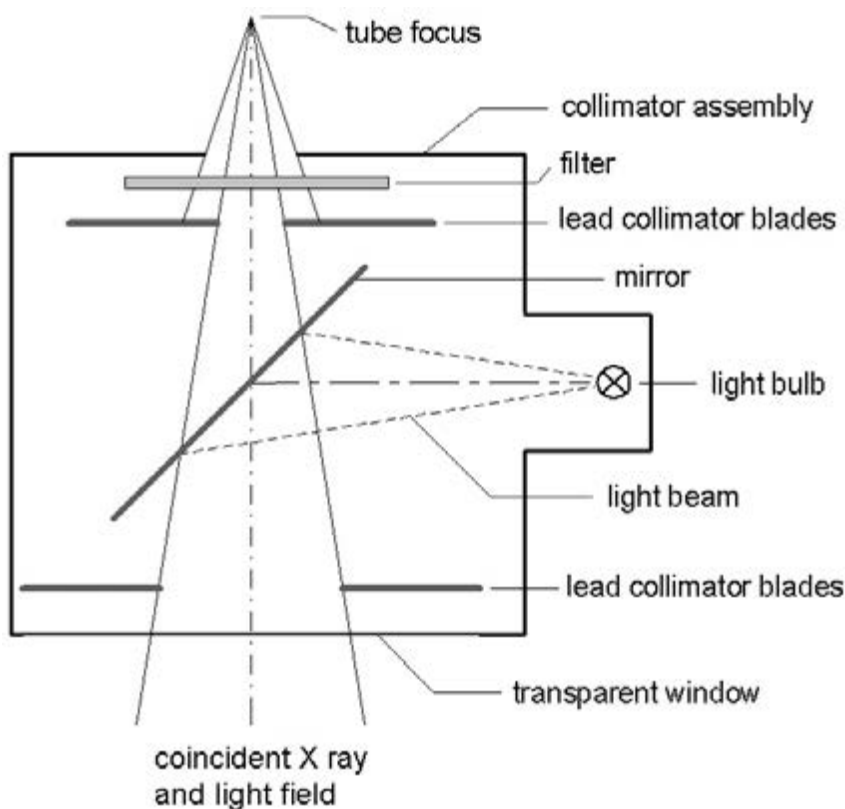


Fig 2.7. Typical X ray field collimator assembly.

Adjustment of the field size is done manually by the operator, but with a positive beam limitation system, the size of the imaging detector is automatically registered and the field size is adjusted accordingly.

For fluoroscopy, other collimator types are in use, with variable circular and slit diaphragms. In some applications (dental and head examinations), beam restrictors with a fixed field size are typically used.

2.4.1.4.2. Inherent filtration:

X-rays generated in the anode pass various attenuating materials before leaving the tube housing. These materials include the anode, tube envelope exit port (glass or metal), insulating oil and the window of the tube housing.

This inherent filtration is measured in a luminiun equivalent (unit: mm Al).

Aluminium does not perfectly mimic the atomic composition of the attenuating

materials present, thus, measurement of the Al equivalent is usually made at 80 kVp (or otherwise the kVp settings should be stated). Typically, the inherent filtration ranges from 0.5 to 1 mm Al. The mirror and the window in the collimator housing also contribute to inherent filtration with an Al equivalent of about 1 mm.

2.4.1.4.3. Added filtration:

Since filtration effectively reduces the low energy component in the X ray spectrum, a minimum total filtration of at least 2.5 mm Al is required to reduce unnecessary patient dose. Additional filter material is positioned between the tube window and collimation assembly as required. Typical filter materials include aluminium and copper, and in some cases, rare earth filters such as erbiums that utilize K edge attenuation effects. Individual filters may be manually selected on some units. In modern fluoroscopy units, filters are inserted automatically, depending on the examination programme chosen.

The effect of added filtration on the X ray output is an increase in the mean photon energy and half value layer (HVL) of the beam. As the X rays become more penetrating, less incident dose at the patient entrance is required to obtain the same dose at the image receptor, giving a patient dose reduction. Since image contrast is higher for low energy X rays, the addition of filters reduces image contrast and optimum conditions must be established, depending on the type of examination. Added filtration also increases tube loading, as the tube output is reduced and must be compensated for by an increase in mAs to obtain the image receptor dose required. In mammography, special provisions concerning filtration are required to obtain the optimum radiation qualities. (Carl A Carlsson, et al, 1996)

2.4.1.4.4. Compensation filters:

In some examinations, the range of X ray intensities incident upon the image receptor exceeds the capabilities of the detector. Compensation or equalization filters can be used to reduce the high intensities resulting from thinner body parts or regions of low attenuation. Such filters are usually inserted in the collimator assembly or close to the

tube port. Examples of compensation filters include wedge filters for lateral projections of the cervical spine, or bowtie filters in CT. (EUR 16260 EN, 1996).

2.5. Factors Influencing X ray spectra and output:

2.5.1. Quantities describing X-ray output:

Total photon fluence is not a satisfactory quantity to describe X ray output; rather, it is the spectral distribution of the photon fluence as a function of photon energy that is useful for research in X-ray imaging. Spectral data are rarely available for individual X-ray units; although computer programs exist that give useful simulations.

X-ray tube output can be expressed in terms of the air kerma and measured free in air. A measure of the penetration and the quality of the X-ray spectrum is the HVL. The HVL is the thickness of absorber needed to attenuate the X ray beam incident air kerma by a factor of two. In diagnostic radiology, aluminium is commonly chosen as the absorber, giving the HVL (unit: mm Al). (EUR 16260 EN, 1996).

2.5.2. Tube voltage and current:

The effect of tube voltage on spectral distribution. Both maximum and mean photon energy depend on the voltage (kV). The shape of the low energy end of the spectrum is determined by the anode angle and the total filtration. Note the appearance of characteristic radiation in the 100 kV beam and the increase in photon yield with increasing tube voltage. Tube current has no influence on the photon distribution; however, photon intensities are proportional to mAs. (Harjit Singh , et al 2012).

2.5.3. Filtration:

As low energy photons do not contribute to the formation of an image, filters are used to reduce the low energy component. Figure 5.21 illustrates the effect of added filters on an X ray spectrum (90 kV, 3.4% ripple) Again. (EUR 16260 EN, 1996).

2.6. Image Quality:

Image quality is a generic concept that applies to all types of images. It applies to medical images, photography, television images, and satellite reconnaissance images.

"Quality" is a subjective notion and is dependent on the function of the image.

In radiology, the outcome measure of the quality of a radiologic image is its usefulness in determining an accurate diagnosis. It is important to establish at the outset that the concepts of image quality discussed below are fundamentally and intrinsically related to the diagnostic utility of an image. Large masses can be seen on poor-quality images, and no amount of image fidelity will demonstrate pathology that is too small or faint to be detected. The true test of an imaging system, and of the radiologist that uses it, is the reliable detection and accurate depiction of subtle abnormalities. With diagnostic excellence as the goal, maintaining the highest image fidelity possible is crucial to the practicing radiologist and to his or her imaging facility. While technologists take a quick glance at the images they produce, it is the radiologist who sits down and truly analyzes them. Consequently, understanding the image characteristics that comprise image quality is important so that the radiologist can recognize problems, and articulate their cause, when they do occur.

As visual creatures, humans are quite adroit at visual cognition and description.

Most people can look at an image and determine if it is "grainy" or "out of focus."
(Bushberg JT et al, 2002).

2.6.1. Contrast:

Contrast is the second major feature of an image. This characteristic describes how well the image distinguishes subtle features in the object (patient). Image contrast is depicted in Figure 16-5. In diagnostic imaging, the contrast of an image is a product of complex interactions among the anatomic and physiologic attributes of the region of interest, the properties of the imaging method and receptor employed, and conscious efforts to influence both the intrinsic properties of the region and its presentation as an image. These interactions are characterized as four influences on image contrast. (Harjit Singh , et al 2012).

2.6.2. Intrinsic Contrast:

The underlying philosophy of diagnostic imaging is that structures in the patient can be distinguished in an image because they differ in physical composition and physiologic behavior. These differences are referred to as intrinsic (sometimes called subject, object, or patient) contrast. Physical properties of the patient that contribute to intrinsic contrast.

In radiography, intrinsic contrast is a reflection primarily of atomic number and physical density differences among different tissues. Some structures (e.g., breast) exhibit very subtle differences in composition and are said to have low intrinsic contrast. Other structures (e.g., chest) provide large differences in physical density and atomic composition and yield high intrinsic contrast. (Harjit Singh , et al 2012).

2.6.3. Imaging Technique

Every imaging application reflects a choice of a specific imaging technique and receptor among many alternatives to yield images of greatest potential to provide the desired information. For a specific imaging application, image contrast can be influenced by careful selection of the technique factors used to produce the image.

In radiographic imaging of the breast, for example, subtle differences among tissues can be accentuated in the image by use of low peak kilovoltage and small amounts of beam filtration. These choices enhance the differential transmission of x rays through tissues that vary slightly in atomic composition and physical density. In chest radiography, relatively high peak kilovoltage and large amounts of beam filtration are chosen to suppress differential absorption of x rays in bone so that lesions of increased physical density can be detected in the lung parenchyma under the ribs. The choice of pulse sequence and other variables in MRI strongly influences the resulting contrast among structures in the image. (Murat et al, 2010).

2.6.4. Contrast Agents:

In some radiologic examinations a substance can be introduced into the body to enhance tissue contrast. This substance, termed a contrast agent (or contrast medium or dye), is selected to provide a signal different from that of the surrounding tissues.

In angiography, a water-soluble agent containing iodine is injected into the circulatory system to displace blood and thereby enhance the contrast of blood vessels by increasing the attenuation of x rays impinging on the vessels. This technique can be further distinguished as arteriography or venography, depending on which side of the circulatory system the injection is made. The iodine attenuates x rays strongly because its K-absorption edge of 33 keV is ideally positioned with respect to the energy distribution of x rays typically employed for diagnosis.

Several other types of contrast agents are used in diagnostic imaging. For example, a contrast agent can be used to displace the cerebrospinal fluid in studies of the central nervous system. For studies of the spinal cord, the contrast agent may be “ionic” or “nonionic,” and the procedure is termed “myelography.” Contrast agents are useful in computed tomography as well as in radiography. Iodine-containing compounds are also helpful in studies of the kidneys and urinary tract, where the applications are known as urography. Compounds containing barium are frequently used for radiographic examination of the gastrointestinal tract, with the barium compound administered orally (barium swallows) or rectally (barium enemas).

2.6.5. Detector Contrast:

An imaging detector receives the incident energy striking it, and produces a signal in response to that input energy. Because small changes in input energy striking the detector (i.e., subject contrast) can result in concomitant changes in signal (image contrast), the detector's characteristics play an important role in producing contrast in the final image. Detector contrast is determined principally by how the detector “maps” detected energy into the output signal. Different detectors can convert input energy to output signal with different efficiency, and in the process amplify contrast. (Murat et al, 2010).

2.6.6. Contrast and dose in radiography:

The screen-film system governs the overall detector contrast. However, on a case by case basis, the contrast of a specific radiographic application is adjusted according to the requirements of each study. The total x-ray exposure time (which governs the potential for motion artifacts) and the radiation dose to the patient are related considerations. The size of the patient is also a factor. The technologist adjusts the subject contrast during a specific radiographic procedure by adjusting the kVp.

Lower kVp settings yield higher subject contrast, especially for bone imaging. Adjusting the kVp is how the technologist adjusts the beam *quality*. (Dowsett DJ, et al, 1998).

2.6.7. Beam Quality:

Quality refers to an x-ray beam's penetration power and hence its spectral properties. However, at lower kVps, the x-ray beam is less penetrating; the mAs must be increased at low kVp and decreased at high kVp to achieve proper film density. The mAs is the product of the mA (the x-ray tube current) and the exposure time (in seconds); it is used to adjust the number of x-ray photons (beam *quantity*) produced by the x-ray system at a given kVp.

How does the technologist determine which kVp to use for a specific examination?

The appropriate kVp for each specific type of examination is dogmatic, and these values have been optimized over a century of experience. The kVp for each type of radiographic examination (e.g., forearm; kidney, ureter, and bladder; cervical spine; foot; lateral hip) should be listed on a *technique chart*, and this chart should be posted in each radiographic suite. The technologist uses the recommended kVp for the radiographic examination to be taken, but usually adjusts this value depending on the size of the patient. Thicker patients generally require higher kVps. Most modern radiographic systems use automatic exposure control (i.e., photo timing), so that the time of the examination is adjusted automatically during the exposure.

Photo timing circuits have a backup time that cannot be exceeded. If the kVp is set too low for a given patient thickness, the time of the exposure may be quite long, and if it exceeds the time that is set on the backup timer, the exposure will terminate.

In this situation, the film will generally be underexposed and another radiograph will be required. The technologist should then increase the kVp to avoid the backup timer problem. Use of a proper technique chart should prevent excessively long exposure times. To illustrate the change in beam penetrability versus kVp. (Murat et al, 2010).

2.7. Unsharpness:

Every image introduces an element of “fuzziness” or blurring to well-defined (sharp) boundaries in the object (patient). This blurring is described as *unsharpness* and is depicted in. Unsharpness (termed “blur” in some texts) is a measurable characteristic of images. (Thayalan, 2001).

Unsharpness in an image is a consequence of four factors that contribute to image formation. These factors, considered collectively as the components of unsharpness, are geometric unsharpness U_g , subject unsharpness U_s , motion unsharpness U_m , and receptor unsharpness U_r . Under conditions where each of these factors contributes independently to the overall unsharpness of the image, the total unsharpness is:

$$U_t = [(U_g)^2 + (U_s)^2 + (U_m)^2 + (U_r)^2]^{1/2}$$

2.7.1. Geometric Unsharpness:

Geometric unsharpness is a direct consequence of the geometry of the image-forming process. In general, geometric unsharpness is influenced by the size of the radiation source and the distances between source and object (patient) and between object and image receptor. These influences are depicted in (Fig 2.8) for the case of x-ray imaging. With minor modifications, they are applicable to all cases of radiologic imaging.

In the upper left diagram of Figure 2.8, the radiation source is very small.

In this case, the sharp borders in the object are blurred over a finite region of the image. The image presents the borders with a degree of fuzziness that increases with the size of the focal spot. The blurring also increases with the distance of the image

receptor from the object, as shown in the lower left diagram of (Fig 2.8). By similar reasoning, the geometric unsharpness is reduced by moving the radiation source away from the object, thereby increasing the source-to-object distance. (Thayalan, 2001).

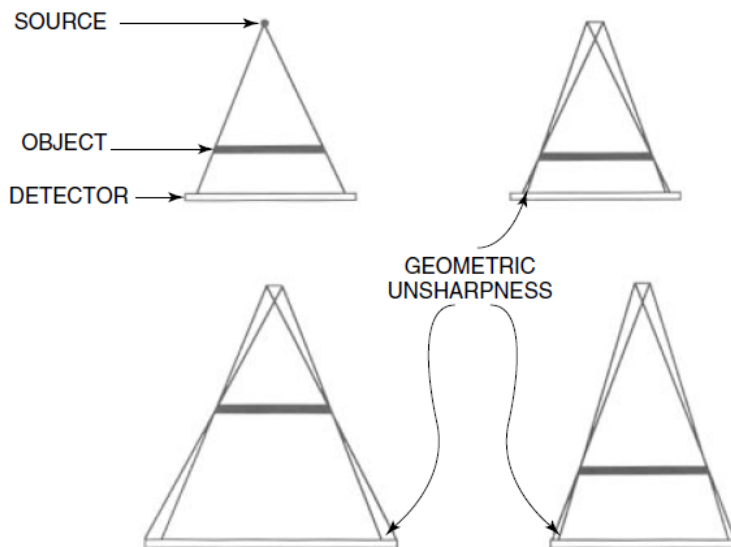


Fig 2.8. Geometric unsharpness. **Upper left:** Minimal unsharpness with a focal spot of infinitesimal size. **Upper right:** Unsharpness contributed by a focal spot of finite size. **Lower left:** Increased unsharpness caused by moving the receptor farther from the object. **Lower right:** Reduction in unsharpness caused by moving the object farther from the radiation source and closer to the detector.

2.7.2. Subject Unsharpness:

Not all structures within the patient present well-defined boundaries that can be displayed as sharp borders in the image. Often a structure is distinguishable anatomically from its surroundings only by characteristics that vary gradually over distance.

Also, the shape of an object may prevent the projection of sharp boundaries onto the image receptor. This latter situation is depicted in (Fig 2.9), where the object on the left presents sharp borders in the image and the object on the right is depicted with blurred edges. Image unsharpness that is contributed by the object is known as subject (object) unsharpness. It can be a result of the composition of the object, its shape, or a combination of both. (Dowsett DJ, et al, 1998).

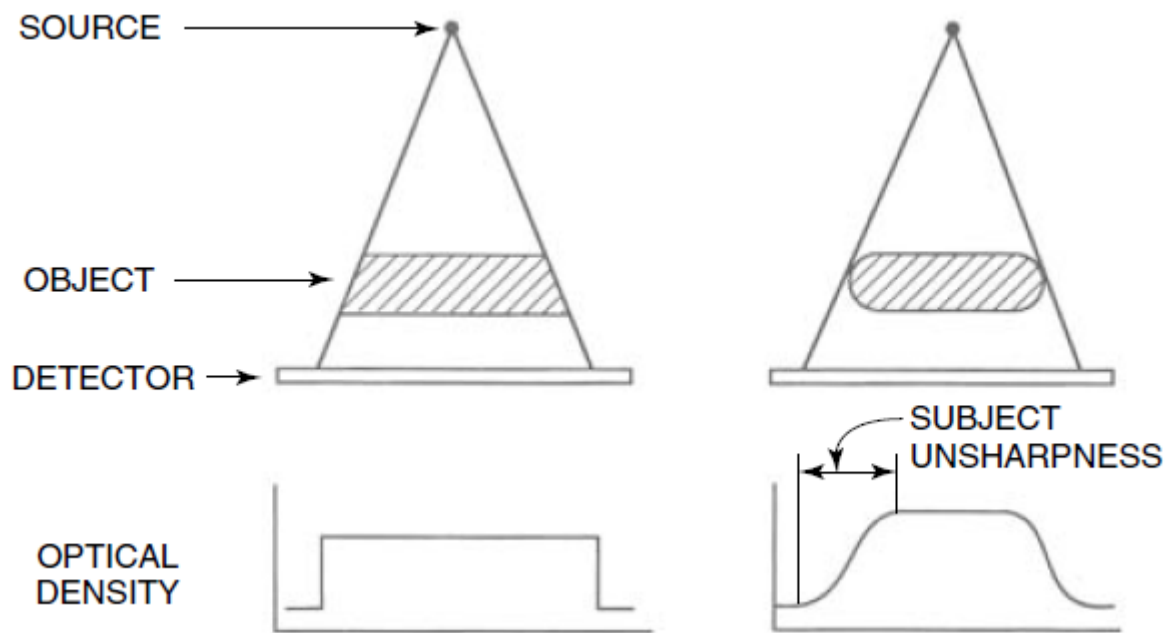


Fig 2.9. Subject unsharpness. Edges of the trapezoid on the left are parallel to the path of x-rays, and the resulting image reveals no subject unsharpness. The ellipsoid on the right yields an image of varying density from the edges to the center and presents the observer with substantial subject unsharpness.

2.7.3. Motion Unsharpness:

Motion is often a major contributor to unsharpness in a radiologic image. Usually the motion occurs in the anatomic region of interest as a result of involuntary physiologic processes or voluntary actions of the patient. Motion causes boundaries in the patient to be projected onto different regions of the image receptor while the image is being formed. As a result, the boundaries are spread over a finite distance, and the resulting borders are blurred in the image.

Voluntary motion often can be controlled by keeping examination times short and asking the patient to remain still during the examination. However, these exhortations sometimes are ineffective, especially when the patient is an infant, demented, or in pain. In some cases, a family member can hold or soothe the patient and reduce voluntary motion during the examination. If the family member is exposed to radiation, protective garments should be provided to reduce the exposure. Occasionally, physical restraints and anesthetics may be required.

Involuntary motion is a more frequent problem and is especially troublesome

when images of fast-moving structures such as the heart and great vessels are required.

For many regions, motion can be “stopped” in the image by the use of very short examination times. In chest images, for example, examination times of a few milliseconds are used to gain a reasonable picture of the cardiac silhouette without the perturbing influence of heart motion. Short examination times are also desired in studies of the gastrointestinal tract to reduce motion unsharpness caused by peristalsis.

In fact, short examination times are a general rule in radiography because voluntary and involuntary motion is invariably present to some degree when any anatomic region is being imaged. (Thayalan, 2001).

2.7.4. Receptor Unsharpness:

The image receptor collects data generated during the imaging process and displays it as a gray-scale or color image. In some techniques such as conventional radiography, the receptor converts x-ray data to an image directly by the use of relatively simple devices such as intensifying screens and film. In other imaging methods the conversion process is more complex and employs a computer to form an image on a video display.

In every display technique, no matter how simple, the image receptor inevitably adds unsharpness to the image. This contribution to image unsharpness is termed *receptor unsharpness*.

In radiography, receptor unsharpness is determined principally by the thickness and composition of the light-sensitive emulsion of the intensifying screens. These characteristics influence not only receptor unsharpness but also the sensitivity of the screens to x rays. With increasing thickness, for example, the sensitivity improves and the unsharpness increases. The choice of screens is, consequently, a trade-off between unsharpness introduced by the receptor and that resulting from motion caused by the finite time to record the imaging data.

In many imaging techniques, the technique of image display can influence the overall unsharpness of the image. In digital radiography, for example, the display device can provide different levels of unsharpness. (Dowsett DJ, et al, 1998).

2.8. Image noise:

Every radiologic image contains information that is not useful for diagnosis and characterization of the patient's condition. This information not only is of little interest to the observer, but often also interferes with visualization of image features crucial to the diagnosis. Irrelevant information in the image is defined as image noise. Image noise has four components: structure noise, radiation noise, receptor noise, and quantum mottle. (Dowsett DJ, et al, 1998).

2.8.1. Structure Noise:

Information about the structure of the patient that is unimportant to diagnosis and characterization of the patient's condition is known as structure noise. For example, shadows of the ribs not only are irrelevant in chest radiographs taken to examine the lung parenchyma, but also can hide small lesions in the parenchyma that could be important to characterization of the patient's condition. However, rib shadows are important to the diagnosis of a cracked rib, and in this case the parenchyma and mediastinum could be characterized as structure noise. Structure noise is defined not only by the information present in the image but also by the reasons why the image was obtained in the first place.

Structure noise is one of the more disturbing features of radiologic images and is often responsible for missed lesions and undetected abnormalities. In radiography, patients frequently are positioned for examination so that structure noise is reduced in the region of interest in the image. The most obvious method of reducing structure noise is tomography, which blurs structures above and below the plane of interest within the patient. Tomographic images are obtained automatically in computed tomography and MRI, and the superior rendition of low-contrast structures by these technologies is attributable in part to their reduced level of structure noise.

2.8.2. Radiation Noise:

Radiation noise is information present in a radiation beam that does not contribute to the usefulness of the image. For example, many x-ray beams exhibit a non uniform intensity from one side to the other. This non uniformity, termed the *heel effect*, is a result of the production and filtration of the beam in the anode of the x-ray tube.

Although the non uniformity provides information about these mechanisms, that information is irrelevant to diagnosis of the patient's condition. Similarly, the radiation beam exiting from the patient contains considerable scattered radiation. This radiation contains information about the patient, although in such a complicated form that it cannot be easily decoded. As it impinges on the image receptor, scattered radiation interferes with the visualization of patient anatomy. Hence, scattered radiation is radiation noise. In projection radiography, the contribution of scattered radiation to image noise is so severe that it must be removed by a grid to permit the structures of interest to be seen clearly. (Dowsett DJ, et al, 1998).

2.8.3. Receptor Noise:

Most image receptors are not uniformly sensitive to radiation over their active surfaces.

These receptors impose a pattern of receptor noise onto the image. In radiography, (Dowsett DJ, et al, 1998).

2.9. Previous Studies:

In Klaus BAVENDIEK et al, 2012. Measurement Methods of Focal Spot Size and Shape of X-ray Tubes in Digital Radiological Applications in Comparison to Current Standards, Unsharpness in the image may reduce the visibility of details. Magnification is often required to achieve a spatial resolution similar to film technique, when using digital systems. A part of the unsharpness results from the focal spot size. To determine the effective focal spot size, thousands of images with different tubes, energies, magnifications and IQIs from different EN and ASTM standards were taken to evaluate the influence of the unsharpness in the digital image due to the focal spot. This leads to reference values for the effective focal spot size of

the different tubes. These reference values were compared with the values of the pin hole camera images according to EN 12543-2 and they deviated significantly. Several modifications of the evaluation method were proposed and a method with Integrated Line Profiles which produces results like an edge profile method was developed and tested providing similar results as the reference values. A second more simple method for end-users was introduced based on the evaluation of Penetrameter hole edges. A proposed update of the standards for focal spot measurements in ASTM and ISO is introduced. New classes will help the users to identify more simple and reliable the X-ray tube for their application. A study was performed on the basis of more than 1000 images from nine different focal spots to measure the size of focal spots in digital images. The results showed small variations ($<<10\%$) for the four different IQIs applied, energies from 90keV to 225keV and magnifications from 2.0 to 15. It could be shown that the image capture procedures with pin hole cameras as described in EN 12543-2 and ASTM E 1165 provide accurate results, and only a different evaluation method is required to meet the values of RVSS. The integrated line profile (ILP) method provides the best results. These are in the range of the RVSS. Since the ILP method converts the pin hole images to physically equivalent edge profiles, the new pin hole method with the ILP evaluation provides values for the spot sizes similar to the standard procedure of EN 12543-5. This resolves the inconsistency of EN 12543-2 and -5 and will lead to compatible standards for measurement of spot sizes of Mini to μ -Focus tubes. A classification with new class limits from $0.4\mu\text{m}$ to $>4\text{mm}$ is proposed for a new ISO standard. A more simple procedure for the users of X-ray tubes, using plaque hole IQIs (e.g. ASTM E 1025), is proposed. The standard deviations of the ILP method and the user method are smaller than 2%.

In Anis Suhana Ahmad Sabri, et al , 2008. Relationship of Focal Spot Size and Entrance Dose with kVp ,mAs ,and Magnification Factor, Focal spot size was measured using focal spot test tool, Model 112B with the determined kVp, mAs, and magnification factor. Then, using the same parameters as the first experiment, entrance dose was measured using flat ionizationc hamber type77337. The

procedures were repeated by selecting different tube voltage (40to80kVp), tube current and exposure time (2to12.5mAs) and magnification factor values (1.25to2.17) to see their effects on focal spot size and entrance dose. The source-to-image distance is fixed to 100cm. For image optimization, smaller focal spot size is more preferable. This is because, finite focal spot size will produce penumbra that will produce blur image. Generally, the smaller the focal spot size the better in order to improve image sharpness. However, as the electron stream is focused to a smaller area, the power of the tube must be reduced to prevent over heating at the tube anode. From this research, by using high tube voltage and tube current, the ability to resolve line pair per millimeter is limited due to over exposed film. Therefore, for higher tube voltage and higher tube current, focal spot size can not be measured because no meaningful image was produced. Thus, it is important to balance tube voltage and tube current to prevent machine break down and improving image quality. For radiation protection purposes, patient dose must be lower as possible to prevent patient receiving unnecessary exposure. However, to produce low entrance dose, lower tube voltage and tube current is needed. However lower tube voltage and tube current may produce low contrast image and low density film. This is not good for image optimization. Therefore, the value of tube voltage and tube current must be compensate each other in order to maintain image quality and reduction of dose. Major factor that effect entrance dose and focal spot size is magnification factor. Increasing magnification factor also may increase entrance dose. Therefore, in order to reduce patient dose, choice of magnification factor also must be consider. The effect of magnification factor to focal spot size is not too crucial compared to other factor. Focal spot size seen to be almost the same for 1.25,1.43, and 1.67 magnification factor.

Chapter three:

Material and Method

3.1 Materials:

3.1.1. Specification of Phantom:

Skull phantom made from real dry bone impeded with an perplex.

3.1.2. Equipments:

This study was done in Modern Medical Center, The X-ray machine modality is digital radiography. Before collection of data x.ray machines must be calibrated and the specification of X-ray machine was shown in the table below:

Table 3.1 Type and main characteristics of X- ray machine

| Center | Manufacturer | Manufacturing Date | Type | Focal spot (mm) | Total Filtration (mm Al) | Max KVp | Max mA | Max time (s) | Year install |
|--------|-------------------|--------------------|-------|-----------------|--------------------------|---------|--------|--------------|--------------|
| MMC | Toshiba FDR smart | June 2013 | fixed | 1.5-.6 | .9/AL/75 | 125 | 320 | .2 | 2014 |

3.2. Methods:

3.2.1. Imaging technique:

In this study 10 lateral skull phantom images was obtained, X rays examinations consist of three views, the frontal view anterior posterior (AP) and posterior anterior (PA) and the lateral (side) view. For chest X rays it is preferred that the patient stand for this exam, particularly when studying collection of fluid in the lungs and during

the actual time of exposure, the technologist usually asks the patient to hold his or her breath. It is very important in taking a chest x- ray to ensure there is no motion that could detract from the quality and sharpness of the film image.

Table 3.2. Show Exposure factors for the broad and fine focal spot size.

| Focal Spot Size | KVp | mA | mAs | Time (s) |
|-----------------|-----|-----|-----|----------|
| Broad | 70 | 250 | 20 | 0.08 |
| Fine | 70 | 100 | 20 | 0.20 |

3.2.2. Evaluation of Image Quality:

A total of 10 images for skull x-ray were obtained using Phantom at diagnostic department in Modern Medical Center hospital in Khartoum. The radiographic equipment used was Shimadzo imaging system. The target angle for the X-ray tube was 12°, and the measured ripple for tube potential was in the region of 1%. Total filtration for the X-ray system was measure as 2.7 mm of aluminum equivalent. A single exposure control system was available for use in the standing position. the image quality was evaluated by visual inspection by questionnaire 100 professional Radiologist were asked to evaluate the image quality for fine focal image compared to broad focal image, the result was 62% of from the asked group evaluated fine focal spot image better than broad focal spot size image were 28% of from the asked group evaluated fine focal spot size image same quality to broad focal spot size image and 10% from the asked group evaluated broad focal spot size image better than fine focal spot size image.

3.2.3. Study duration:

This study was done in the time of June December 2014 .

3.2.4. Study place:

This study was done in Modern Medical Center .

3.2.5. Method of data analysis:

The data will analyze with SPSS program and excel un to assess the effect of focal spot size on image quality.

3.2.6. Ethical issue:

The study will be conduct after obtain informed consent from study subject and Permission from radiology department.

Chapter Four:

The Results

In the following tables and graph data of study are presented the study data of images quality for fine focal spot compared to broad focal spot.

Table 4-1. Showed the visual assessment of image obtained by fine focal image compared to broad focal image.

| <i>Image Quality</i> | <i>Frequency</i> | <i>Percentage %</i> |
|----------------------|------------------|---------------------|
| Better | 62 | 62% |
| Same | 28 | 28% |
| Poor | 10 | 10% |

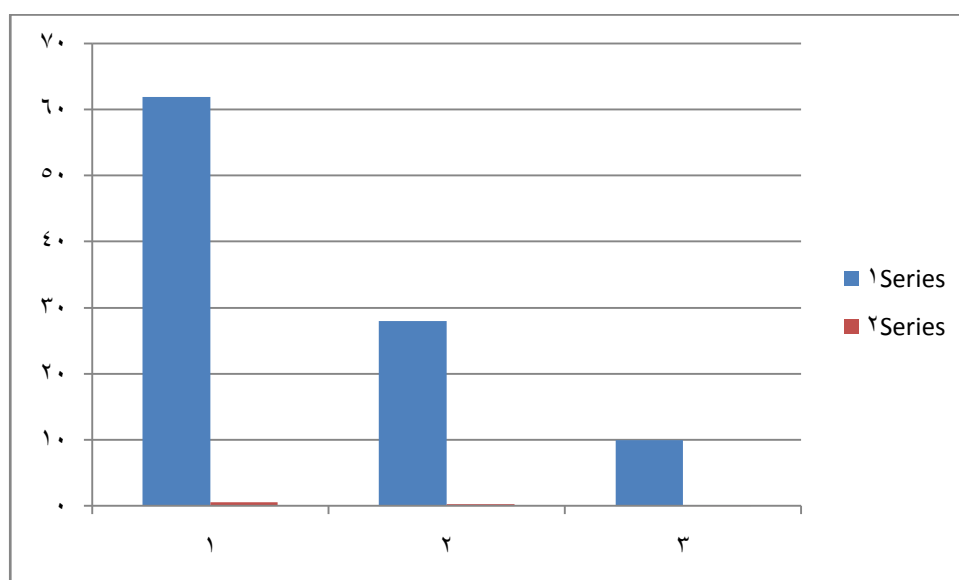


figure 4.1. Showed the visual assessment of image obtained by fine focal image.

Table 4-2. Showed the visual assessment of signal to noise ratio of the image obtained by fine focal image compared to broad focal image.

| | HSNR | ASNR | LSNR |
|------------------|------|------|------|
| Fine Focal Spot | 53 | 35 | 12 |
| Broad Focal Spot | 16 | 33 | 51 |

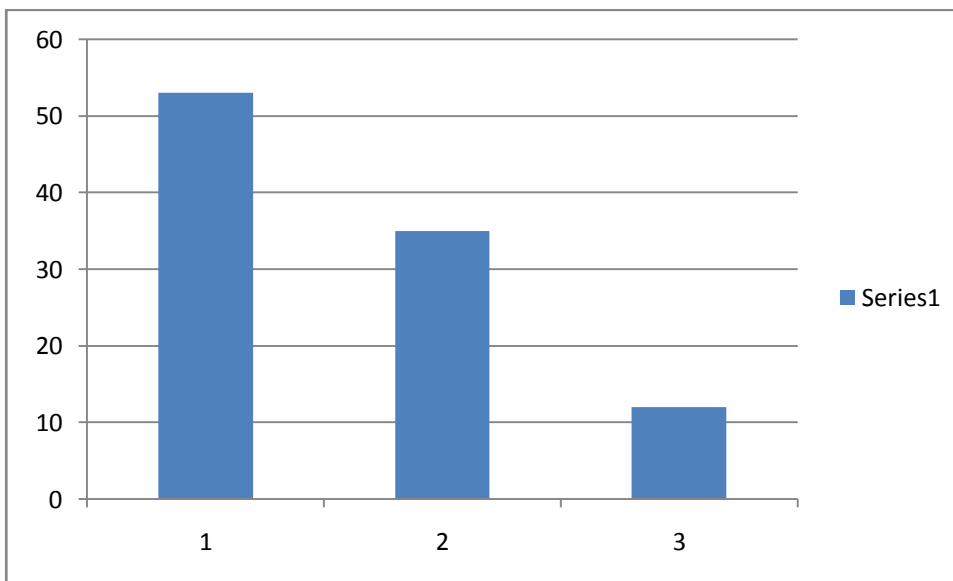


Figure 4.2. Showed the visual Signal to noise ratio assessment of image obtained by fine focal image.

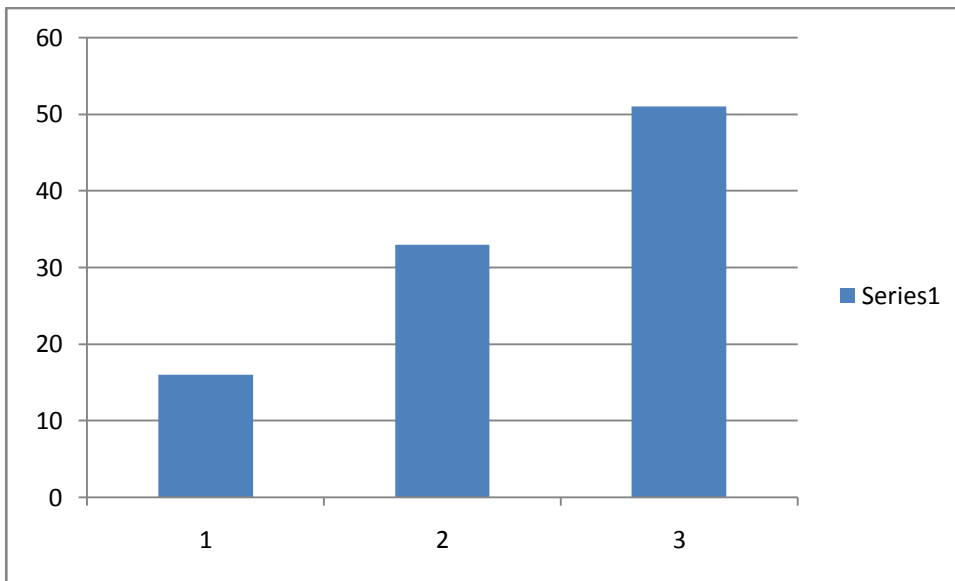


Figure 4.3. Showed the visual Signal to noise ratio assessment of image obtained broad focal image.

Table 4-3. Showed the visual assessment of contrast of the image obtained by fine focal image compared to broad focal image.

| | High Contrast | Acceptable Contrast | Low Contrast |
|------------------|---------------|---------------------|--------------|
| Fine Focal Spot | 61 | 23 | 16 |
| Broad Focal Spot | 13 | 29 | 58 |

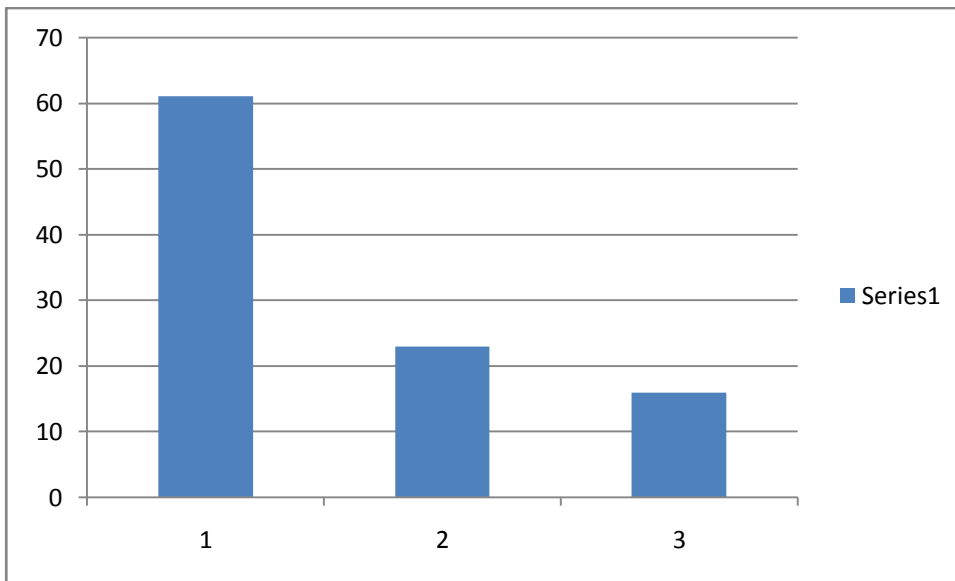


Figure 4.4. Showed the contrast assessment of image obtained fine focal image.

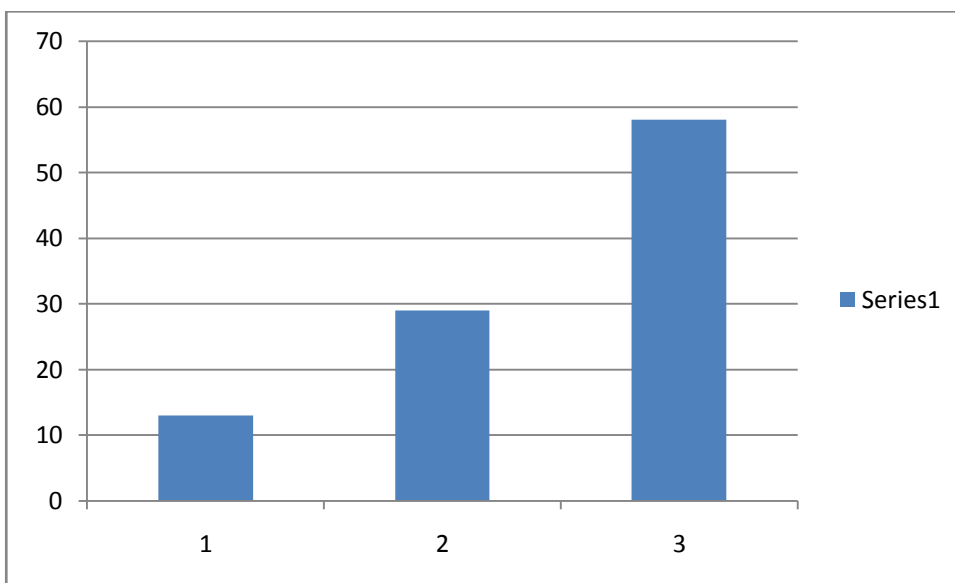


Figure 4.5. Showed the contrast assessment of image obtained broad focal image.

Table 4-4. Showed the visual assessment of resolution of the image obtained by fine focal image compared to broad focal image.

| | High Resolution | Acceptable Resolution | Low Resolution |
|------------------|-----------------|-----------------------|----------------|
| Fine Focal Spot | 59 | 23 | 18 |
| Broad Focal Spot | 16 | 31 | 53 |

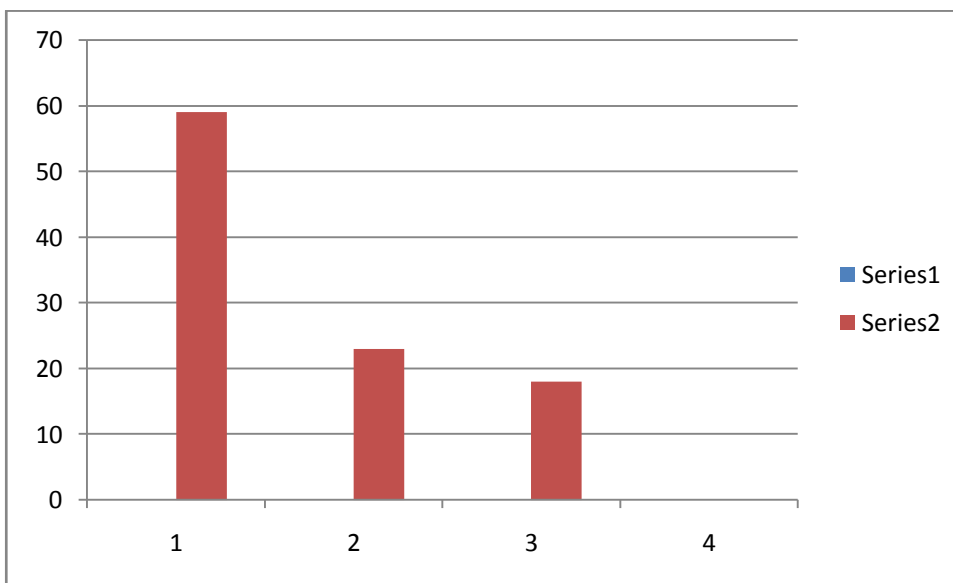


Figure 4.6. Showed the resolution assessment of image obtained fine focal image.

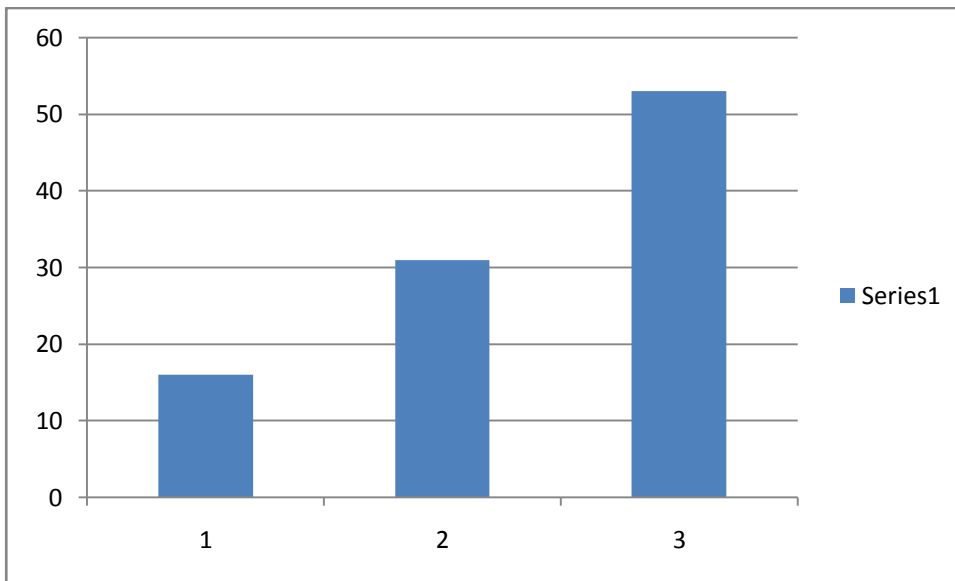


Figure 4.7. Showed the resolution assessment of image obtained broad focal image.

Chapter Five

Discussion, Conclusion and Recommendation

5.1 Discussions:

This study proved that fine focal spot images were sharp in outline compared to broad focal spot images, the image quality was evaluated by visual perception by questionnaire 100 professional Radiologist were asked to evaluate the image quality for fine focal image compared to broad focal image, the result was 62% of from the asked group evaluated fine focal spot image better than broad focal spot size image were 28% of from the asked group evaluated fine focal spot size image same quality to broad focal spot size image and 10% from the asked group evaluated broad focal spot size image better than fine focal spot size image, the result was show that 53% from the asked group evaluated fine focal spot image with in high signal to noise ratio where 35% from the asked group evaluated fine focal spot size image with in the acceptable signal to noise ratio and 12% from the asked group evaluated fine focal spot size image within the low signal to noise ratio . the result was show that 16% from the asked group evaluated broad focal spot image with in high signal to noise ratio where 33% from the asked group evaluated broad focal spot size image with in the acceptable signal to noise ratio and 51% from the asked group evaluated broad focal spot size image within the low signal to noise ratio. The result was show that 61% from the asked group evaluated fine focal spot image with in high contrast where 23% from the asked group evaluated fine focal spot size image with in the acceptable contrast and 16% from the asked group evaluated fine focal spot size image within the low contrast. The result was show that 13% from the asked group evaluated broad focal spot image with in high contrast where 29% from the asked group evaluated broad focal spot size image with in the acceptable contrast and 58% from the asked group evaluated broad focal spot size image within the low contrast. The result was show that 59% from the asked group evaluated fine focal spot image

with in high resolution where 23% from the asked group evaluated fine focal spot size image with in the acceptable resolution and 18% from the asked group evaluated fine focal spot size image within the low resolution. The result was show that 16% from the asked group evaluated broad focal spot image with in high resolution where 31% from the asked group evaluated broad focal spot size image with in the acceptable resolution and 53% from the asked group evaluated broad focal spot size image within the low resolution. The result is same as in previous study; the relative large part of the study (28% + 10%) that presents broad focal images is same quality and better than fine focal spot may be due to the digital image processing.

5.2 Conclusion:

The results demonstrate that the above approach could become e an important part of the image quality in x-ray diagnostic radiography, particularly, to observe changes in focal spot, the result show that brightness distribution is different for images acquired with broad focal spot and for images acquired with fine focal spot, the image obtained with broad focal spot is better than which obtained with fine focal spot in terms of image quality.

5.3Recommendation:

X-ray tube focal spot size is the most important index for QC of x-ray machines, Annual measurement of the x-ray unit focal spot size has been recommended as part of the dental radiographic quality control program.

The size of the radiation source has considerable impact upon the resolution in the image. Focal spot sizes should be measured at acceptance or replacement to ensure proper performance.

Further study should consider another factors which affect image resolution e.g. motions.

In future study the evaluation of images sharpness should be done objectively instead of using visually.

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