Chapter One: Introduction

1.1. Introduction:

Fluoroscopy systems produce projection x-ray images and allow real-time. x-ray viewing of the patient with high temporal resolution. They are commonly used to position the imaging system for the recording of images and to provide imaging guidance for interventional procedures Modern fluoroscopic systems use image intensifiers (IIs) coupled to digital video systems or flat panel digital detectors as image receptors.(Dimov and Vassileva, 2004)

Fluoroscopy has undergone much technological advancement in recent years. medical images information to the interpreting physician such that an accurate diagnosis can be made. A Quality Assurance (QA). program, which includes quality control tests, helps to ensure that high quality diagnostic images are consistently produced while minimizing radiation exposure.(Dimov and Vassileva, 2004)

The QA program covers the entire x-ray system from machine, to processor, to view box. This program will enable the facility to recognize when parameters are out of limits, which will result in poor quality images and can increase the radiation exposure to patients. Simply performing the quality control tests is not sufficient. When quality control test results exceed established operating parameters, appropriate corrective action must be taken immediately and documented. This guide is intended to assist the facility in setting up their QA Program and performing the quality control tests required to maintain high quality images and reduce patient exposure. A fluoroscopy imaging system comprises an X-ray tube and generator with an image intensifier as the image receptor Simply quality control tests is not sufficient. Image quality measurements are needed for several purposes, such as equipment design, performance specification, acceptance and constancy testing in quality assurance and imaging technique

optimization. Most commonly the evaluation of image quality is based on a subjective assessment, either from the images of actual patients or from those of suitable test phantoms. In addition to these methods, there exist several objective measures that can be used to achieve more precise and portable results.

1.2. Problem of study:

Image quality and radiation dose are considered accossiated with each other. and in order to optimize the practice regular and standard QC program should be established in Sudan the regular QC of the fluoroscopy unites is not well perform.

1.3. Objectives

1.3.1 General objective:

to evaluate the QC performance in some fluoroscopy units (image quality).

1.3.2 Specific Objectives:

- to evaluate the frequencies of the image quality tests performance
- to evaluate image quality of some fluoroscopy units in term of resolution and contrast.

1.4 The layout:

This study composed of five chapters, chapter one is introduction, problem of study and objective, chapter two background and pervious study, chapter three material and methods, chapter four results and chapter five discussion, conclusion and recommendation.

Chapter Two

Theoretical Background

2.1. Fluoroscopy:

Fluoroscopy, or real-time projection X-ray imaging, has been in clinical use since shortly after Roentgen's discovery of X-rays. Early fluoroscopes consisted simply of an X-ray source and a fluorescent screen, between which the patient was placed. After passing through the patient, the remnant beam impinged upon the fluorescent screen and produced a visible glow, which was directly observed by the practitioner. In modern systems, the fluorescent screen is coupled to an electronic device that amplifies and transforms the glowing light into a video signal suitable for presentation on an electronic display. One benefit of the modern system compared to the earlier approach is that the fluoroscopist need not stand in close proximity to the fluorescent screen in order to observe the live image. This results in a substantial reduction in radiation dose to the fluoroscopist. Patients receive less radiation dose as well, because of the amplification and overall efficiency of the imaging system. Fluoroscopy differs from most other X-ray imaging in that the images produced appear in real-time, allowing evaluation of dynamic biological processes and guiding interventions. Electronic fluoroscopy systems create this perception by capturing and displaying images at a high frame rate, typically 25 or 30 frames per second. At these frame rates, the human visual system cannot distinguish frame-to-frame variation and motion appears to be continuous, without visible flicker. To achieve high frame rates while keeping cumulative radiation dose to a reasonable level, the radiation dose to the image receptor per image (i.e., per frame) must be kept quite low, about 0.1% of the dose used in radiography. Fluoroscopic images appear with an inverted grayscale (black/white is reversed) compared with standard radiographs. This convention is derivative of the appearance of the early non-intensified fluoroscopic screens, and it has

been retained in the digital age even though the capability now exists to digitally reverse the grayscale. A schematic of an image-intensified fluoroscopy system is shown in Figure 1. The key components include an X-ray tube, spectral shaping filters, a field restriction device (aka collimator), an anti-scatter grid, an image receptor, an image processing computer and a display device. Ancillary but necessary components include a high-voltage generator, a patient-support device (table or couch) and hardware to allow positioning of the X-ray source assembly and the image receptor assembly relative to the patient.

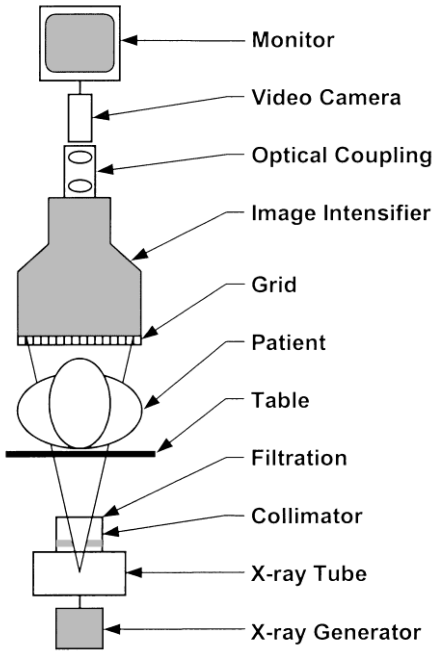


Figure (2-1): Schematic Diagram of a fluoroscopic system using an X-ray image intensifier (XRII) and video camera

(Reprinted from RadioGraphics;20(4), Schueler BA, The AAPM/RSNA physics tutorial for resident's general overview of fluoroscopic imaging – higher-dose radiographic images acquired in rapid succession to visualize opacified vessels. These runs are often interspersed with fluoroscopic imaging in a diagnostic or interventional procedure, and the combination can result in a high demand on the X-ray tube. Special X-ray tubes are generally found in such systems Fig 2, p1117, 2000, with permission from RSNA(Dance et al., 2014)

2.2.X-ray Source:

voltage The generator and X-ray tube used in most fluoroscopy systems is similar in design and construction to tubes used for general radiographic applications. For special purpose rooms such as those used for cardiovascular imaging, extra heat capacity is needed to all angiographic "runs," sequences of high. Focal spot sizes in fluoroscopic tubes can be as small as 0.3 mm (when high spatial resolution is required but low radiation output can be tolerated) and as large as 1.0 or 1.2 mm when higher power is needed. The radiation output can be either continuous or pulsed, with pulsed being more common in modern systems. Automatic exposure rate control maintains the radiation dose per frame at a predetermined level, adapting to the attenuation characteristics of the patient's anatomy and maintaining a consistent level of image quality throughout the examination.(Dance et al., 2014)

2.2.1. Beam Filtration:

It is common for fluoroscopic imaging systems to be equipped with beam hardening filters between the X-ray tube exit port and the collimator. Added aluminum and/or copper filtration can reduce skin dose at the patient's entrance surface, while a low Kvp produces a spectral shape that is well-matched to the barium or iodine k-edge for high contrast in the anatomy of interest. Insertion of this added filtration in the beam path may be user-selectable, providing the operator with the flexibility to switch between low dose and higher dose modes as conditions dictate during a fluoroscopic procedure. In other systems the added filtration is automatic, based on beam attenuation conditions, to achieve a desired level of image quality and dose savings. In addition to beam shaping filters, many fluoroscopy systems have "wedge" filters that are partially transparent to the X-ray beam. These moveable filters attenuate the beam in regions

selected by the operator to reduce entrance dose and excessive image brightness.(Dance et al., 2014)

2.2.2. Collimation:

Shutters that limit the geometric extent of the X-ray field are present in all X-ray equipment. In fluoroscopy, the collimation may be circular or rectangular in shape, matching the shape of the image receptor. Shutters that limit the geometric extent of the X-ray field are present in all X-ray equipment. In fluoroscopy, the collimation may be circular or rectangular in shape, matching the shape of the image receptor. When the operator selects a field of view, the collimator blade positions automatically move under motor control to be just a bit larger than the visible field. When the source-to-image distance (SID) changes, the collimator blades adjust to maintain the field of view and minimize “spillover” radiation outside of the visible area. This automatic collimation exists in both circular and rectangular field of view systems.(Dance et al., 2014)

2.3. Patient Table and Pad:

Patient tables must provide strength to support patients and are rated by the manufacturer for a particular weight limit. It is important that the table not absorb much radiation to avoid shadows, loss of signal and loss of contrast in the image. Carbon fiber technology offers a good combination of high strength and minimal radiation absorption, making it an ideal table material. Foam pads are often placed between the patient and the table for added comfort, yet with minimal radiation absorption.(Dance et al., 2014)

2.3.1. Anti-Scatter Grid:

Anti-scatter grids are standard components in fluoroscopic systems, since a large percentage of fluoroscopic examinations are performed in high-scatter conditions, such as in the abdominal region. Typical grid ratios range from 6:1 to 10:1. Grids may be circular (XRII systems) or rectangular (FPD systems) and are often removable by the operator.(Dance et al., 2014)

2.3.2. Image Receptor —X-ray Image Intensifier:

The X-ray image intensifier (Figure 2) is an electronic device that converts the X-ray beam intensity pattern (aka, the “remnant beam”) into a visible image suitable for capture by a video camera and displayed on a video display monitor. The key components of an XRII are an input phosphor layer, a photocathode, electron optics and an output phosphor. The cesium iodide (CsI) input phosphor coverts the X-ray image into a visible light image, much like the original fluoroscope. The photocathode is placed in close proximity to the input phosphor, and it releases electrons in direct proportion to the visible light from the input phosphor that is incident on its surface. The electrons are steered, accelerated and multiplied in number by the electron optic components, and finally impinge upon a surface coated with a phosphor material that glows visibly when struck by high-energy electrons. This is the output phosphor of the XRII. In principle, one could directly observe the intensified image on the small (1” diameter) output phosphor, but in practice a video camera is optically coupled to this phosphor screen through an adjustable aperture and lens. The video signal is then displayed directly (or digitized), post-processed in a computer and rendered for display.(Dance et al., 2014)

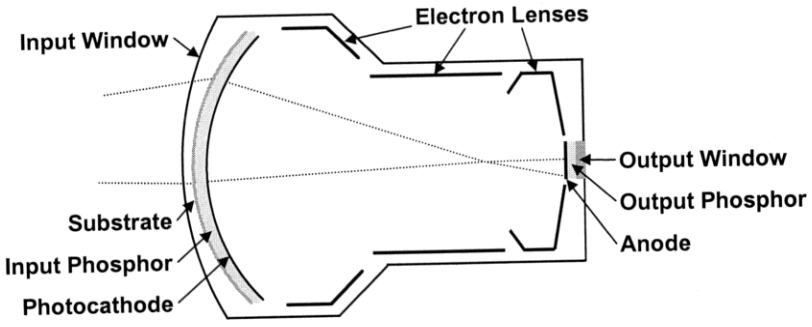


Figure (2- 2): Components of an X-ray image intensifier (Reprinted from RadioGraphics;20(4), Schueler BA, The AAPM/RSNA physics

tutorial for resident's general overview of fluoroscopic imaging – Fig 5, p1120,2000, with permission from RSNA)

The XRII achieves orders of magnitude more light per X-ray photon than a simple fluorescent screen. This occurs through electronic gain (amplification by the electron optics) and minification gain (concentrating the information from a large input surface area to a small output phosphor area) as shown in Figure 2. This allows relatively high image quality (signal-to-noise ratio) at modest dose levels compared with non-intensified fluoroscopy. The use of video technology added an important convenience factor — it allows several people to observe the image simultaneously and offers the ability to record and post-process fluoroscopic image sequences. Image intensifiers are available in a variety of input diameters, ranging from about 10–15 cm up to 40 cm. The input surface is always circular and curved, a design characteristic of the vacuum tube technology from which it is constructed. The video cameras used in XRII systems were originally vidicon or plumbicon analog devices borrowed from the broadcast television industry. In later systems, digital cameras based on charge-coupled device (CCD) image sensors or complementary metal oxide semiconductor (CMOS) technology came into common use.(Dance et al., 2014)

2.4. Image Recepter—Flatpanela Detector (FPD):

In recent years we have seen the introduction of fluoroscopic systems in which the XRII and video camera components are replaced by a "flat panel detector" (FPD) assembly.

When flat panel X-ray detectors first appeared in radiography, they offered the advantages of a "digital camera" compared with existing technologies. In fluoroscopic applications, the challenge for FPDs has been the requirement of low dose per image frame, meaning that the inherent

electronic noise of the detector must be extremely low, and the required dynamic range is high. It has proven to be quite difficult to manufacture FPDs with electronic noise characteristics low enough to achieve good signal-to-noise ratio (SNR) under low exposure conditions, yet such devices do now exist.

Flat panel detectors are more physically compact than XRII/video systems, allowing more flexibility in movement and patient positioning. However, the most important benefit of the FPD is that it does not suffer from the many inherent limitations of the XRII, including geometric "pin-cushion" distortion, "S" distortion, veiling glare (glare extending from very bright areas) and vignetting (loss of brightness at periphery). These phenomena simply do not occur in FPDs. FPDs often have wider dynamic range than some XRII/video systems. Another advantage of FPDs is that the image receptor's spatial resolution is defined primarily by the detector element size, and unlike the XRII/video, is independent of the field of view. In XRII systems, the minification gain requires the entrance dose to vary inversely with field-of-view to maintain a constant brightness at the output phosphor. No such constraint exists for FPDs; the entrance detector dose is independent of the field of view. Flat panel detectors consist of an array of individual detector elements. The elements are square, 140 – 200 microns per side and are fabricated using amorphous silicon thin-film technology onto glass substrates.

Detector arrays used for fluoroscopy range from about 20 x 20 cm up to 40 x 30 cm. A single detector may contain as many as 5 million individual detector elements. A cesium iodide (CsI) scintillation layer is coated onto the amorphous silicon, with thin-film photodiodes and transistors capturing the visible light signal from the scintillator to form the digital image, which is then transferred to a computer at a frame rate selected by the user (Figure 3). Frame rates can be as high as 30 frames per second.(Dance et al., 2014)

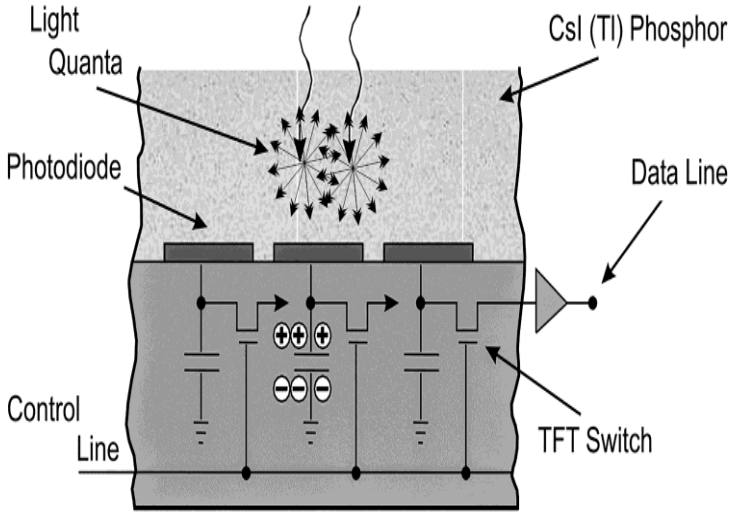


Figure (2- 3): Cross-section of flat panel detector for fluoroscopic imaging (Reprinted from Radiology; 234(2), Pisano ED, Yaffe MJ, State of the Art: Digital Mammography – Fig 1, p355, 2005, with permission from RSNA)

2.4.1. Image Display:

Fluoroscopy requires high-quality video displays that allow users to appreciate fine details and subtle contrast differences in the anatomy of interest. Medical image display technology has been fortunate to "ride on the coattails" of the television industry over the last several years.

Modern systems feature high resolution flat-panel LCDs with high maximum luminance and high- contrast ratios. These displays should be calibrated to a standard luminance response function (such as the DICOM part 14 Grayscale Standard Display Function) to ensure that the widest range of gray levels are visible The newest interventional/angiographic systems feature 60" diagonal high-definition displays supporting up to 24 different video input sources that can be arranged in various ways on the single large display monitor. Display layouts can be uniquely customized and saved for individual physician preference.(Dance et al., 2014)

2.4.2. System Configuration:

receptor above the table, and are most often used for gastrointestinal imaging (upper and lower GI barium enhanced studies). Fluoroscopic systems are manufactured in a variety of configurations to optimize usability for the clinical task(s) for which they are intended. Conventional radiography/fluoroscopy systems consist of a patient table that often tilts all the way to vertical position permitting fluoroscopy while the patient stands upright. These systems have the X-ray tube positioned under the table-top, with the image The tilting capability of the patient table allows the operator to utilize gravity to assist the movement of the barium contrast material through the esophagus, stomach and bowel. Older systems may contain a "spot film" device that allows placement of a radiographic cassette in front of the fluoroscopic image receptor, facilitating the acquisition of radiographs using the fluoroscopic X-ray source. In modern systems, static images are routinely acquired using the same digital image receptor that is used for fluoroscopy, so the spot film is going away. A variation on this conventional R/F configuration is the remote controlled system, in which the X-ray tube and image receptor positions are reversed with the tube above the patient table and the image receptor below. These systems can be fully controlled, including table movements, at an operator's console featuring a joystick-type controller in a shielded control booth. This protects the staff from secondary radiation exposure. Angiographic systems employ a "C-arm" geometry to enable easy patient access as fluoroscopy guides selective arterial and venous catheter placement. These systems include advanced features like digital subtraction and road mapping. The newest systems have 3D imaging capability, achieved by spinning the C-arm around the patient and performing a tomographic reconstruction to produce a volumetric image data set. This is sometimes referred to as cone beam CT (CBCT), while in angiographic mode it is known as 3D rotational angiography. Systems

designed for vascular/interventional radiology and cardiology/electrophysiology have sophisticated fluoroscopic capabilities, including variable frame rate, automatic beam filtration and advanced image post-processing. Finally, the mobile C-arm configuration is popular in the surgical suite and for office-based procedures in musculoskeletal radiology, orthopedics, urology, gastroenterology and pain management among others. Mobile C-arms are often small inexpensive systems, but some are available with higher-power X-ray sources that have the ability to produce substantial radiation output levels.(Dance et al., 2014)

2.4.4. Quality Control:

Following successful installation and acceptance, equipment must be monitored on an ongoing basis to ensure continued, reliable performance. This ongoing, periodic evaluation procedure is quality control (QC). The purpose of QC testing is to detect changes that may result in a clinically significant degradation in image quality or a significant increase in radiation exposure. (Shepard et al, 2002).

2.5. Image Quality:

Unlike snapshots taken on the ubiquitous digital camera, medical images are acquired not for aesthetic purposes but out of medical necessity. The image quality on a medical image is related not to how pretty it looks but rather to how well it con- eyes anatomical or functional information to the interpreting physician such that an accurate diagnosis can be made. Indeed, radiological images acquired with ionizing radiation can almost always be made much prettier simply by turning up the radiation levels used, but the radiation dose to the patient then becomes an important concern. Diagnostic medical images therefore require a number of important trade-offs in which image quality is not necessarily maximized but rather is optimized to perform the specific diagnostic task for which the exam was ordered. The following discussion of image quality is also meant to

familiarize the reader with the terms that describe it, and thus the vernacular introduced here is important as well. This chapter, more than most in this book, includes some mathematical discussion that is relevant to the topic of image science. To physicians in training, the details of the mathematics should not be considered an impediment to understanding but rather as a general illustration of the methods. To image scientists in training, the mathematics in this chapter are a necessary basic look at the essentials of imaging system analysis.(Bushberg et al., 2003)

2.5.1. Spatial Resolution:

Spatial resolution describes the level of detail that can be seen on an image. In simple terms, the spatial resolution relates to how small an object can be seen on a particular imaging system—and this would be the limiting spatial resolution. How- ever, robust methods used to describe the spatial resolution for an imaging system provide a measure of how well the imaging system performs over a continuous range of object dimensions. Spatial resolution measurements are generally performed at high dose levels in x-ray and g-ray imaging systems, so that a precise (low noise) assessment can be made. The vast majority of imaging systems in radiology are digi- tal, and clearly the size of the picture element (pixel) in an image sets a limit on what can theoretically be resolved in that image. While it is true that one cannot resolve an object that is smaller than the pixel size, it is also true that one may be able to detect a high-contrast object that is smaller than the pixel size if its signal amplitude is large enough to significantly affect the gray scale value of that pixel. It is also true that while images with small pixels have the potential to deliver high spatial resolution, many other factors also affect spatial resolution, and in many cases, it is not the pixel size that is the limiting factor in spatial resolution.(Bushberg et al., 2003)

2.5.2. Contrast Resolution:

Contrast resolution refers to the ability to detect very subtle changes in gray scale and distinguish them from the noise in the image. Contrast resolution is character- ized by measurements that pertain to the signal-to-noise ratio (SNR) in an image. Contrast resolution is not a concept that is focused on physically small objects per se (that is the concept of spatial resolution); rather, contrast resolution relates more to anatomical structures that produce small changes in signal intensity (image gray scale), which make it difficult for the radiologist to pick out (detect) that struc- ture from a noisy background.(Bushberg et al., 2003)

2.6. Signal – to - noise ratio:

In all measured or recorded signals there is some contribution from noise. Crackle over the radio or on a mobile phone is perhaps the most familiar phenomenon. Noise refers to any signal that is recorded, but which is not related to the actual signal that one is trying to measure (note that this does not include image artifacts. In the simplest cases, noise can be considered as a random signal which is superimposed on top of the real signal. Since it is random, the mean value is zero which gives no indication of the noise level, and so the quantitative measure of the noise level is conventionally the standard deviation of the noise. It is important in designing medical imaging instrumentation that the recorded signal is as large as possible in order to get the highest signal-to-noise ratio (SNR). An example of the effects of noise on image quality is shown in Figure a. As the noise level increases, the information content and diagnostic utility of the image are reduced significantly. The factors that affect the SNR for each imaging modality are described in detail in the relevant sections of each chapter. However, two general cases are summar- ized here. If the noise is truly random, as in MRI, then the image SNR can be increased by repeating a scan a number of times and then adding the scans together. The true signal

is the same for every scan, and so adds up ‘coherently’: for N co-added scans the total signal is N times that of a single scan. However, the noise at each pixel is random, and basic signal theory determines that the standard deviation of a random variable increases only as the square root of the number of co-added scans. Therefore, the overall SNR increases as the square root of the number of scans. The trade-off in signal averaging is the additional time required for data acquisition which means that signal averaging cannot be used, for example, in dynamic scanning situations. In ultrasonic imaging the situation is more complicated since the major noise contribution from speckle is coherent, and so signal averaging does not increase the SNR. However, if images are acquired with the transducer oriented at different angles with respect to the patient, a technique known as compound imaging, then the speckle in different images is only partially coherent. Averaging of the images, therefore, gives an increase in the SNR, but by a factor less than the square root of the number of images.

In the second general case, as discussed in detail in Chapters 2 and 3, the SNR in X-ray and nuclear medicine is proportional to the square root of the number of X-rays and c-rays, respectively, that are detected. This number depends upon many factors including the output dose of the X-ray tube or the amount of.

The ultimate limit to the SNR is the radiation dose to the patient, with limits which are controlled by various government guidelines throughout the world. X-ray images in general have very high SNR, but for nuclear medicine scans the number of c-rays detected is much lower, and so the scanning time is prolonged compared to X-ray scans, with the total time limited by patient comfort.(Smith and Webb, 2010)

2.7 Automatic Exposure Control:

Radiographic system use automatic exposure control (AEC) device that automatically adjust radiographic technique factor (most often the mAs) to deliver a constant signal intensity at the image receptor in response to differences in patient thickness, X–ray tube energy, focus to detector distance and other technical factor, similarly, in fluoroscopy system the AEC control IAKR to the XRII, to prevent fluctuation in image brightness and SNR that would make diagnosis or navigation of instrument difficult. Fluoroscopy AEC may use the signal a sensor such as a photodiode or photomultiplier tube or more, commonly, the signal from the video camera or directly from flat panel image receptor, to detriment necessary adjustments of fluoroscopic technique factors such as tube voltage and tube current. The selection of fluoroscopic technique factors follow spredeter mined curves that are stored in the generator and which usually allows for some choices, including a standard curve, including a standard curve, low dose curve and high contrast curve .the complexity of fluoroscopic AEC increases with advanced applications where the AEC assumes control over additional equipment parameters such as pulse length, added filtration and variable aperture setting.(Dance et al., 2014)

2.7.1 Sharpness

The sharpness of a fluoroscopic image is influenced by several factors, including the display matrix, FOV, video camera matrix, focal spot size, geometric magnification, image noise and motion. The impacts of both focal spot size and geometric magnification on image sharpness are discussed. XRii fluoroscopic systems differ from a screen film image receptor in that the limiting resolution varies with operating mode, as described. image noise interacts with sharpness, as it can obscure and blur small details in the image that would normally be visible at a higher iakR.The large

number of signal conversions that occur in an XRii also degrade the sharpness of the fluoroscopic image the sharpness of a fluoroscopic image acquired with a flat panel receptor is affected by the size of the image matrix compared with the display matrix and the pixel size of the receptor, which may vary if pixels are binned at certain FOVs.(Dance et al., 2014)

2.7.2 Artefacts:

artefacts in fluoroscopic imaging usually stem from image distortions caused by the image chain components. XRiis suffer from several common image distortions, including veiling glare, vignetting, blooming, pincushion distortion and s distortion, while flat panel image receptors are generally free from image distortions. Veiling glare is a contrast reducing ‘haze’, not unlike the effect of X ray scatter, that results from the scattering of information carriers within the XRii, including electrons within the electron optical system and, most importantly, light photons within the glass output window, to address the latter, a thick XRii output window is used that may incorporate do pants to absorb scattered light and whose sides are coated with a light absorbing material. in some cases, the optical coupling system between the XRii output phosphor and the video camera is replaced by a direct fiber optic linkage, which also reduces veiling glare. Vignetting is an optical distortion that produces a falloff in light intensity or darkening near the edges of an image this may be caused by a number of factors, including deterioration of the video camera, and is also inherent to multi element lenses. Vignetting can be reduced in some cases by restricting the aperture size.(Dance et al., 2014)

2.7.3 Electronic magnification:

Electronic magnification refers to the use of a focusing electrode to deminify the fluoroscopic image by selecting a smaller portion of the input Phosphor to project on to the output phosphor. Electronic magnification

improves the image Mtf but also decreases magnification gain and decreases the sampling pitch of the input phosphor, increasing noise.

in practice, the increased noise in a magnified fluoroscopic image is compensated for by adjusting the technique factors to maintain a constant perceived noise level in the displayed image. in an XRii the iakR usually increases as the ratio of the areas of the FOV as the image is magnified this not only compensates for the decreased photon fluence per image pixel, but also exactly compensates for the decrease in magnification gain, and therefore image brightness, in an XRii system. flat panel based systems also increase the iakR as the image is magnified in response to changes in the image matrix size.(Dance et al., 2014)

2.8. Image quality measurement instruments:

For assessment of image quality, Huttner Type 53-line pair phantom has been used for measurement of the spatial resolution. The contrast detail phantom (TOR FG1) z was used for determination of contrast detail performance.(Tapiovaara, 2005)

2.8.1 Visual evaluation methods:

In medical imaging it is necessary to define image quality with respect to what is needed to be detected in the image, i.e. as a task-based quantity. Therefore, it may be thought that the most useful way of measuring image quality would be using actual patient images and radiologists. In principle, diagnostic performance can be measured using the receiver operating characteristic (ROC) methodology, but in practice this is too laborious for routine evaluation purposes. Clinical image quality criteria that are based on the visibility of normal anatomy which have been suggested for quality assurance use and imaging technique optimization tasks. Of course, both approaches are important and useful for many purposes, but it is difficult to see how either of them could be considered as an actual measurement that can be calibrated, repeated and compared with results obtained elsewhere.

In addition to image quality, the results depend on the patient material and the radiologist interpreting the images. In the case of clinical quality criteria, the actual significance of the criteria for diagnostic performance is not always guaranteed and the subjective nature of the evaluation will cause additional variability in the results. It cannot be expected that other than exceptionally large changes in imaging performance will be reliably noted by this method. The imprecision caused by the variability in patients can be avoided by using test phantoms instead of patients. This would improve the sensitivity and precision of the assessment and allow, at least partly, the test to be repeated by others. Phantoms can be manufactured with a variable amount of anatomical detail, but usually simple homogeneous phantoms that mimic the radiation attenuation and scattering properties of the human body will suffice. Common test details consist of disks of various contrasts and diameters for measuring the low-contrast-detail detectability (contrast resolution) or contrast-detail performance or lead bar patterns for measuring the limiting spatial resolution. The latter are most often used without an attenuating phantom in order to measure the maximum spatial resolution of the imaging system. In these tests, a numerical test result is obtained, which expresses the faintest or smallest detail seen in the image.(Tapiovaara, 2005)

2.8.2 Objective measurement methods:

In addition to the visual measurements, there exist objective measures related to large-area signal transfer (K), image sharpness (MTF) and image noise (NPS). These can be combined to form the

. quantity NEQ:

$$NEQ\left(\int x, \int y\right) = \frac{k^2 * MTF^2\ (\int x, \int y)}{\mathrm{NPS}\left(\int x, \int y\right)} \quad \text{..... (1)}$$

which can be interpreted to express the quantum fluency that the image is worth at various spatial frequencies (fx, fy). NEQ can be compared with the actual fluency at the image receptor (Q). This
results in the DQE:

$$DEQ(\int x, \int y) = \frac{NEQ(\int x, \int y)}{\mathrm{Q}} \ldots\ldots\ldots\ldots (2)$$

which expresses the efficiency with which the imaging system uses the information carried by the quanta impinging on it.

DQE measures the efficiency of the image receptor, it does not refer to the patient's dose and neither NEQ nor DQE take into account all factors that influence the detectability of the actual object detail, such as the energy dependence of the radiation contrast. These image quality descriptors are therefore not sufficient when, e.g. the imaging conditions are being optimized. They are intended for the evaluation of only one component of the imaging system: the image receptor. If the noise in the image is normally distributed and signal independent, and the imaging system is linear and shift – invareant , the best possible observer can detect a deteal object DS (fx, fy) with SNR:

$$SNR = \int k^2 * MTF^2 (\int x, \int y) * \Delta S(\int x, \int y^2)/NPS(\int x, \int y)]d \int x d \int y$$

$$= \frac{\Delta I(\int x, \int y^2)]}{(NPS(\int x, \int y)]d \int x d \int y} \ldots\ldots\ldots\ldots (3)$$

where the expected (noiseless) image of the detail has been denoted by I (fx, fy). This SNR specifies the ideal observer's detection performance of the given detail completely. For example, the fraction of correct answers the ideal observer achieves in multiple alternative forced-choice (MAFC) tests or its whole ROC curve can be calculated from this quantity. The ideal observer's SNR is the proper quantity to use when the task-dependent

image quality is considered; it takes into account all factors of importance, including the subject contrast. If it is required to relate image quality to the patient's dose, one can evaluate the dose efficiency by calculating the quotient SNR^2/D, where D is the patient dose and can be either the entrance dose or the effective dose, whichever is more appropriate for the evaluation. Image quality measurement in fluoroscopy differs only slightly from the above discussion of static images. However, in this case the NPS and MTF are 3-D quantities: in addition to the two spatial frequencies, they also depend on the temporal frequency. The measurement of the spatial-temporal NPS is straightforward but we are aware of no practical methods for the direct measurement of the spatial-temporal MTF. However, in most imaging systems it may be possible to assume it to be of form:

$$\mathrm{MTF} = \left(\int \mathrm{x}, \int \mathrm{y}\right), \left(\int \mathrm{t}\right) = \mathrm{MTF}\left(\int \mathrm{x}, \int \mathrm{y}\right)\mathrm{MTF}\left(\int \mathrm{t}\right) \ldots\ldots\ldots (4)$$

The SNR^2 of static imaging must be replaced by the accumulation rate of SNR^2, which we denote as SNR^2 rate. It describes the accumulation of information with the temporal length of the image sequence. At first thought, it would appear that the measurement of SNR^2 rate in fluoroscopy would be much more complicated than the measurement of SNR in static imaging. It is, in fact, the other way round. Measurement in fluoroscopy is very easy because a large number of image samples can be readily obtained. The measurement can be done either by analytical calculation, using equation (of SNR^2)
applied to temporal averages of image sequences of reasonable lengths, or by constructing a quasi-ideal observer and letting it observe image samples.

2.9. Literature review:

I. I. Suliman et al, presented result of assessed and evaluated digital fluoroscopy quality control measurements performed in beam quality (Half- value layer) peak tube voltage (KV) accuracy and reproducibility tube-current exposure- time product (mAs) linearity and automatic brightness control (ABC) and patient dose in terms of entrance surface air kerma rate plus image intensifier input air kerma rate. Dose measurements were made using Calibrated dose rate meter. Field limitation and source to skin distant measurements in addition to evaluation radiation protection tools for occupation exposure were performed. Image quality was evaluated in terms of spatial resolution and Contrast detail delectability. Patient dose measurements were performed using polymethyl methacrylate (PMMA) patient equivalent phantom whereas image quality was assessed using Huettner Type 53 spatial frequency grating and TO10 contrast detail phantom. The results show that the measured HVL and peak tube voltage were within the recommended limits of 10 % in four fluoroscopy units. Entrance surface air kerma rate measured ranged from 6.1 to 250 mGy/min for fluoroscopy units operated in pulsed, continuous and cine mode of operation. These results were obtained using varying thicknesses of PMMA phantom. Most values are in reasonable agreement with internationally established reference levels with exception to one fluoroscopy unit where doses were remarkably high. Field limitation and minimum source to skin distance were well within the recommended limits of 30 cm for all fluoroscopy units. The limiting resolution was ranged from1.0 to 2.2 LP /mm for image intensifier field diameters between 7 and 23 cm.(Suliman et al., 2007)

Conor U O, et al, reported result of Acceptance testing of fluoroscopy systems used for interventional purposes and Acceptance

testing performed on 18 fluoroscopy installations (interventional II/TV, interventional FPD, and mobile C-arm II/TV) in a number of Irish hospitals from 1999 to 2008. Acceptance testing and routine quality assurance (QA) of X-ray systems are the requirements of the EU Medical Exposures Directive (MED) and these requirements were subsequently implemented into Irish legislation. The MED states that special consideration should be given to the QA and dose assessment of high dose procedures such as interventional fluoroscopy. QC tests performed include: Tube and generator performance, automatic exposure control (AEC) entrance doses, image quality, electrical safety, mechanical safety, RP and equipment design. The authors stated that, all systems were found to have failed one or more acceptance tests. Dose rate, image quality and RP issues were identified on the majority of systems tested. About50% of systems tested were found to have significant issues requiring action by equipment supplier sprior to the system going in to clinical use. A comparison of entrance doses from a new vascular intervention a l FPD system Vs. two conventional vascular II/TV was done. All three systems were from the same manufacturer. Entrance dose measurements were performed in line with IPEM protocols. The results for doses in the fluoroscopy mode showed that patient EDRs made with a 20 cm water phantom were similar for all three systems (full field setting, normal flouro mode). Detector EDR was highest on the FPD system. In the digital acquisition mode, the dose per frame at the FPD entrance was greater than that at the majority of interventional systems and greater than both conventional II/TV systems (on the same default clinical „dose“ setting). Results from subjective image quality tests using standard Leeds test objects were comparable between the FPD and conventional II/TV fixed systems (noise, threshold contrast detail delectability and limiting spatial resolution). Spatial resolution was observed to be slightly greater on the

FPD system (1.4 Lp/mm vs. 1.25 Lp/mm on both II/TVs). The FPD has a full field size diagonal of 48 cm and the II/TV systems have a field diameter of 40 cm.(O'Connor et al., 2008)

Dimov A A, et al. reported result of Assessment of performance of new digital image intensifier fluoroscopy system and result used to assess the physical parameters including image quality and patient dose rates on a recently installed digital fluoroscopy unit. The Digital Fluoroscopy was new for Bulgaria and there was a gap in the experience within the radiologists in exploring the advantages of this modality for imaging. At the same time in Bulgaria, Quality Assurance protocols in Digital Fluoroscopy does not exist and based on the findings obtained some initial recommendations are prepared. The purpose of these efforts was to propose optimization strategies for digital fluoroscopy of maintaining good diagnostic image quality at minimal patient doses. The modern fluoroscopy units are often automated and software controlled. In the work various users defined and automated modes were examined on an Axiom Icons R200 unit (Siemens, Germany) as respected image quality parameters and patient doses were measured, low and high contrast resolution were assessed for different field sizes and fluoroscopy modes using Leeds type test objects. The Incident Dose rates were measured using standard 30x30 cm2 PMMA phantom with thickness varying between16 and30 cm at different available filtrations, automatic brightness control curves, and pulsed fluoroscopy modes. The Incident Dose Rate (without backscatter) measured on 20 cm PMAA and largest field of view were from 2.9 to 4.0 mGy/min for the different dose modes available. The low contrast sensitivity varied from 1.3 to 1.8 %, as the limiting spatial resolution was changing from 1.6 to 2.8 Lp/mm for the available magnification and dose modes. The authors performance on systems showed a big potential for performance optimization in terms of image

quality and dose. It completely satisfies Quality Control requirements applicable for conventional Image Intensifier systems. The results obtained can be used in two main directions-development of better optimized local practice standards and development of a quality control programmer relevant to digital fluoroscopy systems.(Dimov and Vassileva, 2008)

Zoeelife J, et al. presented result of Quality control measurements for fluoroscopy systems in eight countries participating in the SENTINELEU coordinator action and in SENTINEL work package 1 on functional performance and standards it was decided to organize and perform a trial on image quality and physical measurements. A survey on inventory of equipment and equipment standards was organized to collect in form at ionon equipment available for measurements in the trial, equipment available for toolkit to be used during the trial and protocols available for the measurements. Eight participants responded to the questionnaire. Equipment for the toolkit could be made available by three participants. Among the protocols available for quality control of (digital)fluoroscopy systems the protocol developed by the Department of Medical Physics & Bioengineering, Dublin, Ireland appeared to be the most suitable. In addition, monitors could be checked using a software tool made available by the University of Leuven. The SENTINEL toolkit containing equipment and instructions circulated among seven participants in the period August 2006 to October 2006. Due to problems related to customs (Bulgariaisnotyetafull EU member state) the measurements in Bulgaria were made with local equipment.(Zoetelief et al., 2008)

Suliman I I et al presented result of Digital Fluoroscopy Quality control and the Quality control was performed on three digital fluoroscopy systems used for cardiovascular, digital subtraction angiography (DSA) and digital spot imaging (DSI). Measurements were based on the QC protocol developed in the framework of European Commission (EU) DIMONDIII

project. For digital fluorography system used for DSI, measurements included were beam quality (half-value layer, HVL), peak tube voltage (Kvp) accuracy and reproducibility, tube-current exposure-time product (mAs) linearity and automatic brightness control (ABC). Patient dose and image quality were measured for the two digital fluoroscopy systems using a patient equivalent phantom. Spatial resolution was measured using the Huettner Type 53 spatial frequency grating. A contrast detail phantom CDRAD 2.0 was used to measure the threshold contrast detail performance. The results obtained showed that the measured HVL and peak tube voltage were within the recommended limits of 10 %. Maximum patient entrance surface dose (ESD) rate of 27.8 mGy/min, 46.5 mGy/min and 84.1 mGy/min were measured using a 20 cm thick polymethylmethacrylate (PMMA) phantom at the cardiovascular unit operated in low, normal and high dose mode of operation, respectively. ESD rate of 28 mGy/min and 42.2 mGy/ min were measured using a 20 cm thick PMMA phantom at the DSA unit operated in pulsed and continuous mode of operation, respectively (lowest value should be for pulsed mode). The results of this study were in reasonable agreement with internationally established reference levels. The median limiting resolution was 1.6 Lp/mm for image intensifier field diameters between 14 and 38 cm. Mean low contrast resolution expressed as image quality figure (IQF) was 107.(Suliman et al., 2007)

Walsh c, et al reported result of Subjective and objective measurements of image quality in digital fluoroscopy the result used to compare between the physics and clinical assessment of image quality. Physics assessments were based on IPEM protocols using Leeds test objects. Clinical assessment was based on a questionnaire. A total of 15 systems in three European locations were assessed, covering a range of image intensifier-TV digital fluoroscopy units. Analysis of 274 clinical

questionnaires showed that clinical and physics assessments did not place systems in the same order, based on a given image quality parameter. In almost all the comparisons, low level correlation was measured for statistical comparison of rank order (rs<0.3). However, broad agreement was observed between physics and clinical assessments for image quality associated with contrast and noise. The authors emphasized the importance of maintaining links with clinical assessment, when developing quality assurance metrics, and measuring the mutual performance of clinical and physical assessments of image quality. Clinical and physics assessments based on questionnaire and test object measurements, do not place systems in the same order of merit based on a given image quality parameter. There was, however, evidence for broad agreement between physics and clinical assessments for image quality associated with contrast and noise. The results suggest that both groups judged the systems as operating well („average to good"), but disagreed on the ordering within this category. The study reflects the difficulties of image quality assessment and quality control in the field. It emphasizes the importance of maintaining links with clinical assessment when developing quality assurance metrics and of measuring the mutual performance of clinical and physical assessments of image quality.(Walsh et al., 2005)

Evan D S, et al presented result of Threshold contrast detail delectability curve for fluoroscopy and digital acquisition using modern image intensifier system and present updated TCDD curve for fluoroscopy and new curve for digital acquisition. The images of this test were acquired under standard reproducible exposure conditions, 6 which allow the contrast of the details to be known and which facilitate consistency by using TO10 to fluoroscopy and TO12 to acquisition contrast test object. The image reader scores the image by visually assessing the lower contrast detail visible for each group of the same diameter. The image intensifier

input air kerma was measured using calibrated MDH2025 electrometer with 60cm 3circularionization chamber and copper filtration. The authors results showed that the generation of TCDD curves from images of standard test object is simple procedure that allow analysis of image quality for a range of imaging modalities, including image intensifier systems and comparing between the fluoroscopic systems in acceptance test or in rotation quality control of image quality.(Evans et al., 2004)

Mary Beth Peter, et al, presented result of evaluated Soft-Copy Quality Control of Digital Spot Images Obtained by Using X-ray Image Intensifier and Quality control test performed to assess (XRII) digital spots: entrance exposure, patient exposure, soft-copy gray scale, and pixel noise. Two additional tests were performed to assess high contrast limiting resolution and threshold contrast detection. The authors results showed that the Digital spot (XRII) entrance exposures averaged1x10-7C/kg (0.38 mR) for units with large fields of view (FOVs), mean entrance exposure in a medium-size patient was 1.25 x 10-5 C/kg (48 mR). Also luminance measurements of the table-side monitors provided a mean of 473 just-noticeable differences in gray scale with the room lights off. Mean resolution with a bar test pattern was measured as 1.5 line pairs per millimeter for systems with a 40-cm FOV. Measured pixel noise (in relative units) was 6–25. Mean threshold contrast with the lights off was 0.85%. The authors study concluded that once input exposure was normalized for FOV and image matrix size, soft-copy assessment of limiting resolution with either low-contrast detection or, preferably, an off-line noise metric (pixel SD) provides objective measurements of digital spot image quality. With the lights on, 10 systems with room-light sensors had an 11%loss of grayscale. For systems without sensors, the loss was 33%.(Peter et al., 2000)

T. Q. Simon et al, reported result of evaluated a digital X ray system for cardiology Quality control measurement performed in an image intensifier (II) and a dynamic flat panel detector (FD). Entrance surface air kerma (ESAK) to phantoms of 16, 20, 24 and 28 cm of polymethyl methacrylate (PMMA) and the image quality of a test object were measured. Images were evaluated directly on the monitor and with numerical methods (noise and signal-to-noise ratio). Information contained in the DICOM header for dosimetry audit purposes was also tested. ESAK values per frame (or kerma rate) for the most commonly used cine and fluoroscopy modes for different PMMA thicknesses and for field sizes of 17 and 23 cm for II, and 20 and 25 cm for FD.(Tapiovaara and Sandborg, 1995)

Padovani R, et al presented result of Survey on performance assessment of cardiac angiography systems and assess was performance of different cardiac angiographic systems. A questionnaire was sent to centers participating in SENTINEL Project to collect dosimetry data (typical entrance dose rate in fluoroscopy and imaging mode), image quality evaluations (low and high contrast resolutions) and KAP (kerma area product) calibration factors. There results from that survey could contribute to the explanation of patient dose variability in angiographic cardiac procedures and to derive reference levels for cardiac angiographic equipment performance parameters. Tests included measurement of air kerma dose rates in fluoroscopy and digital acquisition modes and a subjective assessment of image quality using the Leeds test object (TOR 18FG).Dose rates were measured under automatic exposure control in fluoroscopy and digital acquisition modes measured the entrance surface air kerma rate with a phantom of 20 cm PMMA thickness to simulate patient attenuation, and the field of view (FOV) on the detector had been sited at 22 cm or nearest with a focus-entrance phantom distance of 65 cm

and the image detector positioned at 5 cm from the exit phantom surface. With the purpose to use the KAP meter calibration factor to correct collected patient KAP values, the calibration procedure was performed taking into account the attenuation determined by the patient table and mattress. The calibration had been performed at60–80–100kVX-rayqualitieswithanionchamberonthe axis of the X-ray beam placed at minimum 10 cm away from the patient table and the image detector to avoid scatter. The different X-ray qualities were reached by inserting in the X-ray beam between the ion chamber and the image detector, attenuating material (copper and/or aluminum) simulating the patient attenuation with both kilo volt and added filtration use to typical clinical conditions. Surface area was calculated from field dimensions measured wit haradio-opaqueruler or an equivalent method. KAP calibration factor was assumed as the mean value of the calibration factor measured for the three X-ray qualities. The authors survey on the cardiac angiographic units in a sample of European centers demonstrates a large variability in entrance dose rates for fluoroscopy and image acquisition modes, image quality performance and KAP calibration. As an outcome of the study, a preliminary set of reference levels for the ESAK quantity was proposed, which can be adopted by centers and maintenance engineers to setup cardiac equipment at an acceptable dose performance level and by standardization bodies as an input to introduce proper standards. SENTINEL consortium is finally recommended as a European action directed to harmonize the level of performance of angiographic systems used in the daily cardiac practice.

Chapter three

3-Material and Methods

3. 1. Materials:

3. 1. 1 Test Phantom

TOR 18 FG is test object designed to be used quickly and easily on a regular basis to provide an ongoing check of imaging performance, particularly those aspects which are most liable to deterioration. After an initial grey-scale check, image quality is measured simply by counting the number of details detected and the number of bar patterns resolved in the image. An ongoing record of these numbers will reveal any trend towards deterioration in imaging performance. TOR 18FG used for fluoroscopy and fluorography. TOR 18FG enables the following checks to be made:

- Monitor brightness and contrast level adjustment (highlight and lowlight details).
- Resolution limit (0.5 to 5.0 LP/mm).
- Low-contrast large-detail detectability (18 details, 8mm diameter, contrast range 0.009 to 0.167 @ 70kVp 1mm Cu).
- Circular Geometry (Lead Circle).

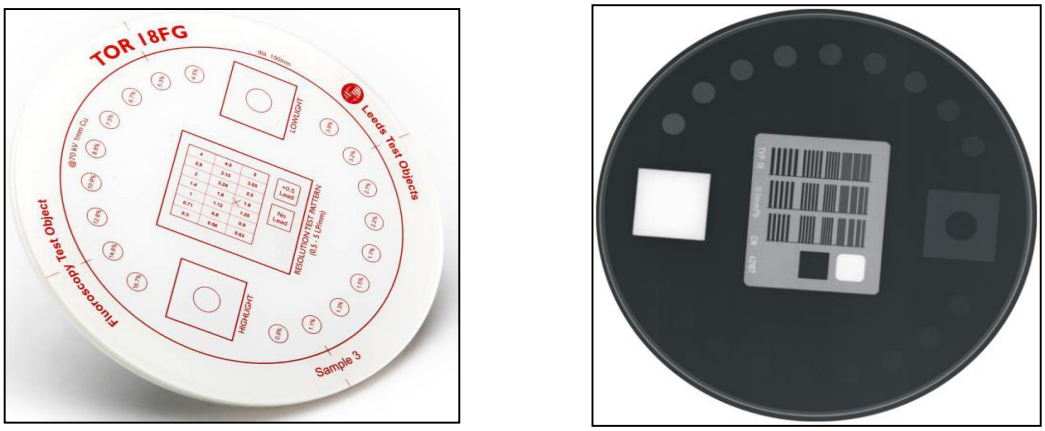


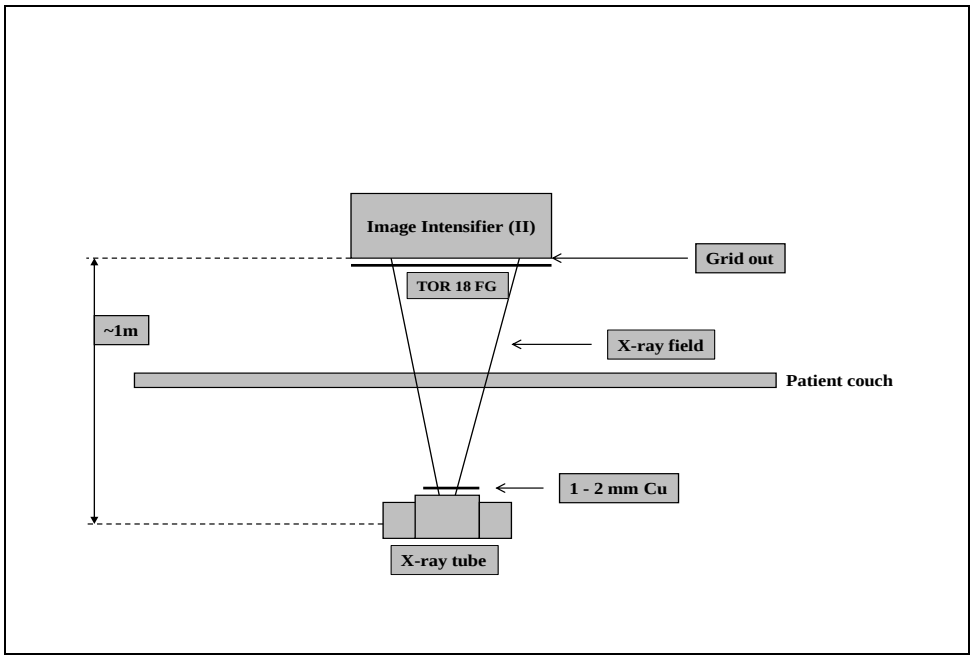
Figure (3-1): TOR 18 FG physical shape.

The physical shape of the phantom is showing in Figure (3-1)

3.2. Methods:

3. 2. 1 Procedure

The common clinical mode of operation was used i.e. pulse rate, dose level, total filtration, …etc. The phantom (TOR 18 FG) was placed at the center of the x-ray field close to the image receptor (image intensifier (II)) entrance plane. The fluoroscopy unit was operated with Automatic exposure control (AEC). 2 mm Cu placed close to the X-ray tube face. The images were acquired as the soft copy on the TV monitor. The image of the phantom was examined and evaluated for two criteria, low contrast detectability and limiting resolution. The evaluation was done visually to obtain the limiting resolution and the low contrast detectability of the system.



The setup of the phantom is showing in Figure (3-2).

Chapter four

Result and discussion

4.1. Results:

The results are presented for image quality measurements performed in three fluoroscopy units. Image quality was evaluated in terms of limiting resolution and low contrast detectability.

Table (4- 1). limiting Resolution

Filed size	Hospital I	Hospital 2	Hospital 3
Large	16	10	14
Medium	14	10	12
Small	11	6	10

Table1. Shows that the limiting resolution for Varity filed size (large medium and small)

Table (4-2).Low contrast detectable

Field Size	Large	Medium	Small
H1	11	11	11
H2	13	12	11
H3	12	13	11

Table2. Show that the low contrast detectable for Varity filed size (large medium and small)

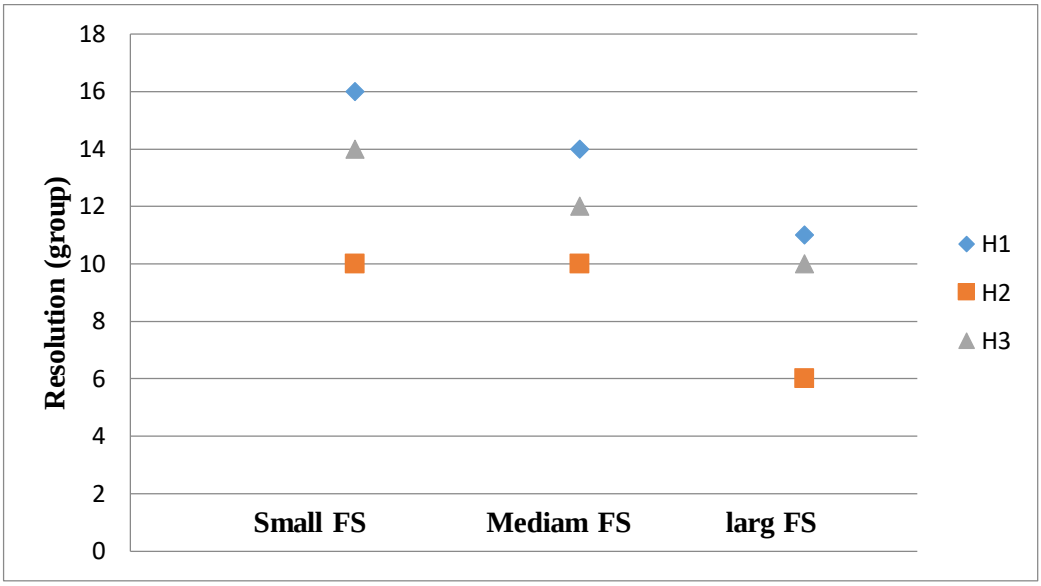


Figure (4-1) represent limiting resolution vs. filed size

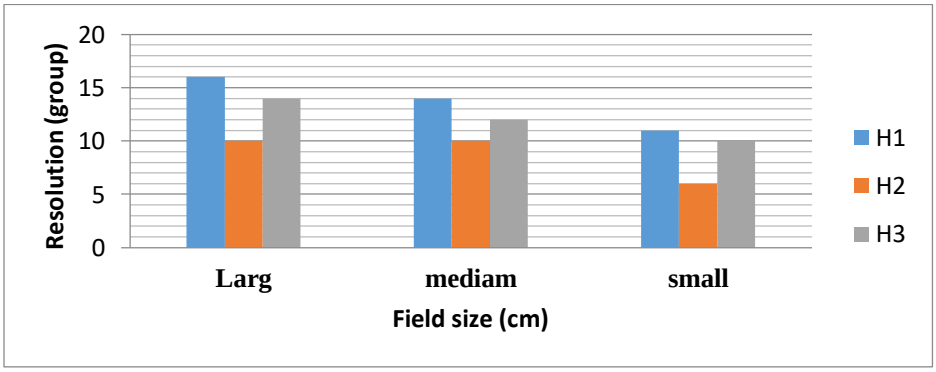


Figure (4-2). Show that resolution is inverse proportional to filed size in most hospital.

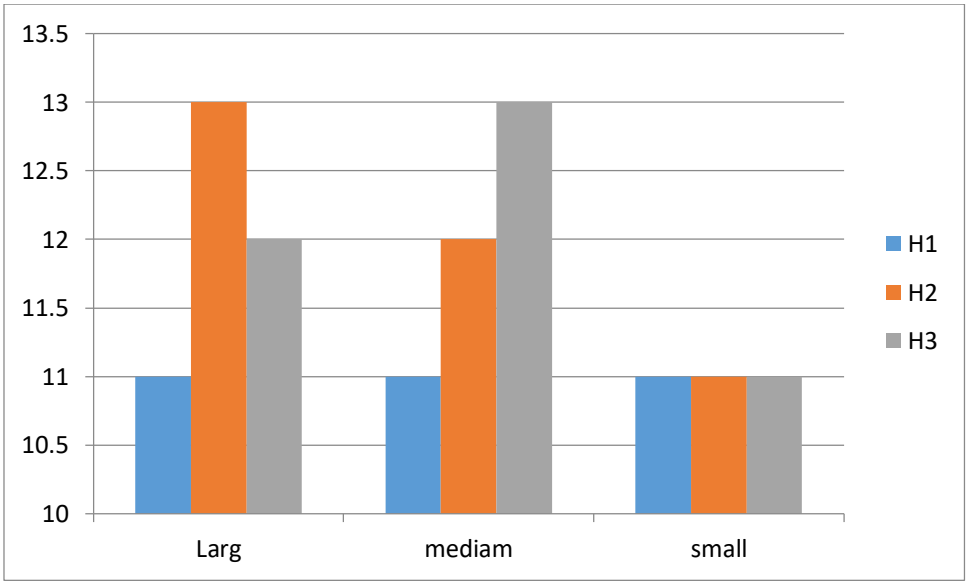


Figure (4.3) Show that low contrast detectability in all hospital

Chapter Five

Discussion, conclusion and Recommendation

5.1. Discussion:

5.1.1. Limiting resolution:

The results of image quality measurements in terms of limiting resolution shown in Table 1 indicate that for all systems the resolution was within recommended tolerance level (0.9 to 1.6 1Lp/mm for image intensifier FOV 19- 32) in this study the limiting resolution was less than some similar studies such as Conor U O, Mary Beth Peter, I. I. Suliman, Suliman I. I, and Dimov A. A, and limiting resolution were (1.25, 1.33, 1.5, 1.6, 1.6, and 2.2) respectively. in all hospital the limiting resolution is inverse proportional to filed size figure 1 and 2.

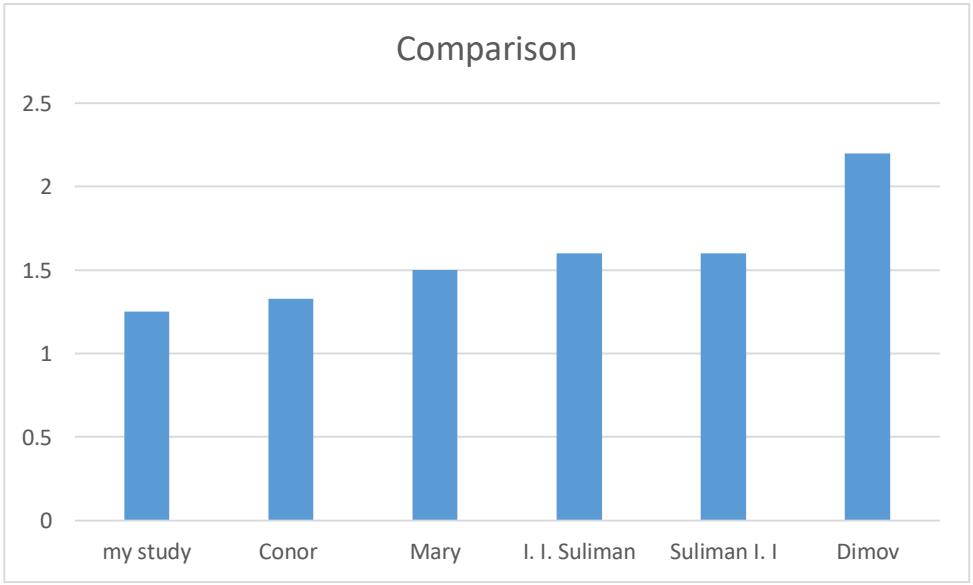


Figure (5-1) comparison between my study and similar studies in spatial resolution.

5.1.2. Threshold contrast detail delectability:

The threshold contrast detail detectability TCDD was measured using TOR FG18 in the table show that the H1 hospital in verity field size the number of desk seen as same and this result are not acceptable re compare the recommended value , in hospital H2 the field size are inverse proportional to low contrast delectability and all result are acceptable to compare the recommended value and hospital H3 the field size inverse proportional low contrast delectability accept the small field size and result are acceptable to compare recommended value accept small filed size as general show H2 as better than H_1 and H2 and H3 and better than H_1

5.2. Conclusion:

Results from subjective image quality tests using standard Leeds test objects were compared between the systems (threshold contrast detail detectability and limiting spatial resolution).Spatial resolution was similar approximately between systems (between1Lp/mmto2Lp/mm) for fields of view between 19 to 32cmThe actual reason for the measured weak or suboptimal performance in the low contrast sensitivity, limiting spatial resolution and low contrast detectability were compared with similar studies. Some basic parameters that can give appropriate measure of the image quality can be assessed easy with a combined phantom similartoFL18.

5.3 Recommendation:

Were commend that the image quality control of fluoroscopy machines she pus to make sure that the device performs proper way and will facilitate the repair, and they must be regular evaluation to fluoroscopy machines.

I hope in the future studies to perform quality control and measurement doses to patient and make survey area to protect staff and public.

The last recommended for use from the Health Inspectorates and from the local hospital staff at quick inspections and routine performance tests respectively. The other IQ tests need more specialized training of the medical physicists and availability of full range IQ standard test objects and have to be performed at the commissioning of the fluoroscopy units. The explored methods for QC tests have to be applied on more units typically used in the country. The results may indicate a need for revision of the nationally adopted acceptability criteria for the fluoroscopy equipment.

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