Sudan University of Science and Technology

Collage of Graduate Studies

Estimation of Entrance Skin Dose (ESD) in Intravenous Urography

تقدير دخول جرعة الجلد في فحص الجهاز البولي

By:

Hanadi Mohammed Alawad jadalla

Supervisor:

Dr: Ahmed Mostafa Abukonna

May 2015

الايـــــة

(والله أخرجكم من بطون امهاتكم لا تعلمون شيئا وجعل لكم السمع والأبصار والأفئدة لعلكم تشكرون)

صدق الله العظيم

سورة النحل الآية (78)

Dedication

This work is dedicated to my parents, brothers and sisters,

To my teachers,

To my friends,

And colleagues

Acknowledgements

My acknowledgements and gratefulness at the beginning and at last is to God who gave us the gift of the mind.

My gratitude is extended to my supervisor **Dr. Ahmed Mostafa Abukonna** for his support and guidance, without his help this work could not have been accomplished.

My gratitude is also extended to my colleagues in Radiology Department at Military hospital for their continuous help and support.

Finally, my profound thanks and gratitude to everyone who encouraged me to complete this thesis.

ABSTRACT

Intravenous urography (IVU) is efficient radiological examinations for the evaluation of the urinary system disorders. However patients are exposed to a significant radiation dose.

The objective of this study was to estimate skin dose for IVU. A total of 44 patients from two public hospitals underwire using a Shimadzu X ray machine. The result of the study revealed that Patients ESDs value for IVU was 1.496 ± 0.29 mGy for total sample. Inter-hospital variation was reported; this is could be possibly due to the number of films taken and due to variations in patients' size. Furthermore the output of the machine could be contributed to this variation.

In this study the radiation dose is considered low compared to previous studies.

الخلاصة:

تصوير الجهاز البولي عن طريق الوريد هو من الفحوصات الاشعاعية الفعالة لتقييم اضرابات الجهاز البولي ، ومع ذلك ييتعرض المرضى الي جرعة اشعاعية كبيرة.

الهدف من هذه الدراسة هو قياس دخول الجرعة الاشعاعية للمرضي الخاضعين لفحص الجهاز البولي.

قد تم التحقق من 44 مريض من مستشفيين حكوميين باستخدام جهاز الاشعة السينية.

وكشفت الدراسة ان متوسط قيم الجرعة التى تدخل لجلد المريض الخاضع لفحص الجهاز البولي لكل العينات هو (0.29 ± 1.495) .

وجد ان هناك اختلاف في قيم كل من المستشفيين ،ربما يرجع الي عدد الافلام التي اخذت او لاختلاف في في حجم المريض. وكذلك يمكن ان يساهم في هذا الاختلاف .

في هذه الدراسة تعتبر الجرعة الاشعاعية منخفضة مقارنة مع الدراسات السابقة .

Table Contents

Chapter One

INTRODUCTION

1.1 Introduction

The urinary system is one of the most important systems in the body, and disturbance of it, leads to major problems all over the body. Intravenous urography (IVU) is one of a radiographic study of the urinary system using an intravenous contrast agent (Ballinger, et al 1999)

Diagnostic x-ray radiology is a common diagnostic practice and there has been a substantial increase in the number of examinations recently (Bushong, 2001). Inspite of the increasing hazard of diagnostic x-rays to human beings, studies aimed at achieving low patient doses with sufficient image quality have continued to be of interest in research (ICRP, 1991 and United Nations Scientific Committee on the Effects of Atomic Radiation, 2000). All exposures to ionizing radiation needs to justified and optimized in terms of the benefit and risks (ICRP, 1991).

Entrance skin dose (ESD) is an important parameter in assessing the dose received by a patient in a single radiographic exposure. The European Union has identified this physical quantity as one to be monitored as a diagnostic reference level in the hopes of optimizing patient dose (Bushong, 2001 and ICRP, 1991).

Patient doses in diagnostic x-ray examinations can be best estimated in terms of entrance surface dose (ESD) per radiograph or dose area product (DAP) for the complete examination (European Commission, 1996). On the other hand, the effective dose is the best quantity for estimating radiation risks to the patients. The major benefit of using the effective dose is that this parameter accounts for the absorbed doses and relative radiosensitivities of the irradiated organs in the patients and, therefore, better quantifies the patient risk (ICRP, 1991). In this study, the patient is exposed to high doses in a short time several times more than four exposure making some organs such as the kidneys and the bladder and ureter affected by this radiation .

The aim of this study was to estimate of dose received by patients in IVU using the entrance skin doses (ESD) of patients during routine x-ray examinations of the kidney, bladder, ureters in two public hospitals.

1.2 Research problems

The size of irradiated volume relative to the treated volume (and integral dose) may increase with increase numbers of beams, but both volumes can be reduced by beam shaping and conformal therapy. The organs at risk are critical normal tissues whose radiation sensitivity may significantly influence treatment planning and/or prescribed dose.

In this study, the patient is exposed to high doses in a short time several times making some organs such as the kidneys and the bladder and ureter affected by this radiation

1.3 Objectives

1.3.1 General objective

To estimate the entrance skin dose for the patients undergoing IVU.

1.3.2 Specific objectives

- To investigate the correlation between ESD & age, FSD, kVp and mAs.
- To identify the difference between hospitals in ESD received by the patient

1-4 Definition of Entrance skin dose (ESD)

The entrance skin dose ESD is defined as the absorbed dose to air on X-ray beam axis at the point where X-ray beam enters the patient or a phantom, including the contribution of the backscatter (Toivonen M, 2001)

1-5 ESD calculation

In this study the entrance skin doses ESDs calculated using the equation:

$$ESD = OP \times \left(\frac{Kv}{80}\right)^2 \times mAs \times \left(\frac{100}{FSD}\right)^2 \times BSF \qquad (1\text{-}1)$$

Where

OP ≡ Out Put Factor

FSD ≡ Focal Skin Distance

BSF ≡ Back Scatter Factor

The tube output obtained from the calibration curve. The focal skin distance (FSD) obtained by subtracting the nominal patient thickness from the focal film distance (FFD) by equation (2-1). The back scatter factor (BSF) used in this study was 1.4 [8, 9]

$$\mathrm{FSD} = \mathrm{F}FD - Thickness \quad (2-1)$$

To calculate the ESD, the following X-ray tube exposure parameters were recorded for each patient take diagnostic examination: peak tube voltage (kVp), exposure current–time product (mAs) and focus-to-film distance (FFD). In addition to, the patient information i.e. patient age and weight.

1.6 Thesis layout:

This study contains five chapters, chapter one contains introduction and objectives of the study. Chapter two contains literature review, chapter three contains material and method, chapter four contains results and in chapter five includes discussion, conclusion and recommendations.

Chapter two

Theoretical Background and Previous study

2.1 Radiation dose quantities and unit

Radiation dosimetry deals with methods for a quantitative determination of energy deposited in a given by directly or indirectly ionizing radiation   that as (IAEA 2005)

2.1.1 Absorbed dose

The fundamental physical quantity of energy deposition average over specified organs and tissues (ICRP 2006). So it energy (joules or ergs) deposited per unit mass (martin 2006). Absorbed dose derived from the mean value of stochastic quantities, the SI unit J/Kg the traditional unit is (Rad) and the special unit is Gray (Gy) (Thayain; 2001).

2.1.2 Gray

The basic unit of radiation called the gray (Gy) and it defined as: One gray is an absorbed radiation dose of one joule per kilogram. The gray is universal applicable to all type of ionizing radiation dosimetry (Lapp and Andrews; 1972).

2.1.3 Rad

Before the universal absorption of SI unit, radiation dose was measured by a unit called the rad (Radiation Absorbed Dose). One rad is an absorbed radiation dose of 100 ergs per unit gram (Lapp and Andrews 1972).

2.1.4 Dose equivalent

The radiation protection unit is called dose equivalent use to specify exposure limits for keeping the occurrence of stochastic health effects below acceptable levels and for avoiding tissue reactions in workers and public. The unit of dose equivalent is joules per kilograms (J/Kg) and its special name is sievert (SV).and its traditional unit is REM (Radiation Equivalent man) (Thayain; 2001).

2.1.5 Effective dose

Since the basic protection standard were derived from risk data based on whole body doses, a mechanism is necessary to adjust doses to internal organs to one effective whole body dose (Martin 2006). In practice the effective dose itself cannot measure directly in body tissues. The protection system therefore includes operational quantities that can be measured and from which the effective dose can be assessed, the unit of effective dose J/Kg with special name sievert (SV) (Thayain; 2001).

2-2 X- radiation:

2-2-1 Discovery of X-rays:

In November 1895 Wilhelm Roentgen, a physicist at the University of Wurzburg, was experimenting with cathode rays. These rays were obtained by applying a potential difference across a partially evacuated glass tube. Rontgen observed the emission of light from crystals of barium platinocyanide some distance away, and he recognized that the fluorescence had to be caused by radiation produced by his experiments. He called the radiation X-rays and quickly discovered that the new radiation could penetrate various materials and could be recorded on photographic plates. Among the more dramatic illustrations of these properties was a radiograph of a hand. Within a month of their discovery, X-rays were being explored as medical tools in several countries. Two months after Rontgen's discovery, Poincare demonstrated to the French academy of sciences that X-rays were released when cathode rays struck the wall of a gas discharge tube (William and Ritenour; 2002).

2-2-2 Physics of X-ray production

There are two different mechanisms by which X-rays are produced. One gives rise to bremsstrahlung X-rays and the other characteristic X-ray (Faiz et al, 2003).

2-2-2-1 Bremsstralung X-rays

The process of bremsstrahlung is the result of radioactive interaction between a high-speed electron and a nucleus. The electron while passing near a nucleus may be deflected from in path by the action of Coulomb forces of attraction and loses energy as bremsstrahlung, a phenomenon predicted by Maxwell's general theory of electromagnetic radiation. According to this theory, energy is propagated through space by electromagnetic fields. As the electron, with its associated electromagnetic field, passes in the vicinity of a nucleus, it suffers a sudden deflection and acceleration. As a result, a part or all of its energy is dissociated from it and propagates in space as electromagnetic radiation. Since an electron may have one or more bremsstrahlung interactions in the material and an interaction may result in partial or complete loss of electron energy, the resulting bremsstrahlung photon may have any energy up to the initial energy of the electron. Also, the direction of emission of bremsstrahlung photons depends on the energy of the incident electrons. At electron energies below about 100 KeV, X-rays are emitted more or less equally in all directions. As the kinetic energy of the electrons increases, the direction of X-ray emission becomes increasingly forward. Therefore, transmission-type targets are used in megavoltage X-ray tubes accelerators in which the electrons bombard the target from one side and the X-ray beam is obtained on the other side. In the low voltage X-ray tubes, it is technically advantageous to obtain the X-ray beam on the same side of the target i.e. at 90 degrees with respect to the electron beam direction. The energy loss per atom by electrons depends on the square of the atomic number(Z^2).Thus the probability of bremsstrahlung production varies with(Z^2) of the target material. However the efficiency of X-ray production depends on the first power of atomic number and the voltage applied to the tube. The term efficiency is defined as the ratio of output energy emitted as X-rays to the input energy deposited by electrons. It can be shown in equation (2.1) that:

$$\text{Efficiency} = 9 \times 10^{-10}\, Z \times V \qquad (2-1)$$

where V is tube voltage in volts ,from the above equation it can be shown that the efficiency of X-ray production with tungsten target $(Z = 74)$ for electrons accelerated through 100 kV is less than 1% the rest of the input energy (~99%) appears as heat, the accuracy of above equation is limited to a few MV (Faiz et al, 2003).

2-2-2-2 Characteristics X-rays

Electrons incident on the target also produce characteristic X-rays. An electron with kinetic energy E_0, may interact with the atoms of the target by ejecting an orbital electron, such as a K, L, or M electron, leaving the atom ionized. The original electron will recede from the collision with energy when a vacancy is created in an orbit; an outer orbital electron will fall down to fill that vacancy. In doing so, the energy is radiated in the form of electromagnetic radiation. This is called characteristic radiation, i.e., characteristic of the atoms in the target and of the shells between which the transitions took place. With higher atomic number targets and the transitions involving inner shells such as K, L, M, and N, the characteristic radiations emitted are of high enough energies to be considered in the X-ray part of the electromagnetic spectrum (Faiz et al, 2003).

2-2-3 Spectrum of X-ray

X-ray photons produced by an X-ray machine are heterogeneous in energy. The energy spectrum shows a continuous distribution of energies for the bremsstrahlung photons superimposed by characteristic radiation of discrete energies. If no filtration, inherent or added for the beam, the calculated energy spectrum will be a straight line and mathematically given by Kramer's equation.

$$I_E = KZ(E_m - E) \quad (2-2)$$

Where:

I_E ≡The intensity of photons with energy E

$K \equiv$ Constant

$Z \equiv$ the atomic number of the target

$E_m \equiv$ The maximum photon energy

The inherent filtration in conventional X-ray tubes is usually equivalent to about 0.5 to 1.0 mm aluminum. Added filtration, placed externally to the tube, further modifies the spectrum. It should be noted that the filtration affects primarily the initial low-energy part of the spectrum and does not affect significantly the high energy photon distribution. The purpose of the added filtration is to enrich the beam with higher-energy photons by absorbing the lower energy components of the spectrum. As the filtration is increased, the transmitted beam hardens, i.e., it achieves higher average energy and therefore greater penetrating power. Thus the addition of filtration is one way of improving the penetrating power of the beam. The other method, of course, is by increasing the voltage across the tube. Since the total intensity of the beam decreases with increasing filtration and increases with voltage, a proper combination of voltage and filtration is required to achieve desired hardening of the beam as well as acceptable intensity. The shape of the X-ray energy spectrum is the result of the alternating voltage applied to the tube, multiple bremsstrahlung interactions within the target and filtration in the beam. However, even if the X-ray tube were to be energized with a constant potential, the X-ray beam would still be heterogeneous in energy because of the multiple bremsstrahlung processes that result in different energy photons. Because of the X-ray beam having a spectral distribution of energies, which depends on voltage as well as filtration, it is difficult to characterize the beam quality in terms of energy, penetrating power, or degree of beam hardening. A rule of thumb is often used which states that the average X-ray energy is approximately one-third of the maximum energy or kVp. Of course, the one-third rule is a rough approximation since filtration significantly alters the average energy. Another quantity, known as half-value layer, has been defined to describe the quality of an X-ray beam (Faiz et al, 2003)

2-2-4 X-ray interaction with matter

When traversing matter, photons will penetrate, scatter, or be absorbed. There are four major types of interactions of X- ray photons with matter, the first three of which play a role in diagnostic radiology: Rayleigh scattering, Compton scattering, photoelectric absorption and pair production (Jerrold et al; 2002)

2-2-4-1 Rayleigh scattering

In Rayleigh scattering, the incident photon interacts with and excites the total atom, as opposed to individual electrons as in Compton scattering or the photoelectric effect. This interaction occurs mainly with very low energy diagnostic X- rays, as used in mammography (15 to 30keV). During the Rayleigh scattering event, the electric field of the incident photon's electromagnetic wave expends energy, causing all of the electrons in the scattering atom to oscillate in phase. The atom's electron cloud immediately radiates this energy, emitting a photon of the same energy but in a slightly different direction. In this interaction, electrons are not ejected and thus ionization does not occur. In general, the scattering angle increases as the X-ray energy decreases. In medical imaging, detection of the scattered X-ray will have a deleterious effect on image quality. However, this type of interaction has a low probability of occurrence in the diagnostic energy range. In soft tissue, Rayleigh scattering accounts for less than 5% of X-ray interactions above 70 keV and at most only accounts for 12% of interactions at approximately 30 keV. Rayleigh interactions are also referred to as "coherent" or "classical" scattering (Jerrold et al; 2002).

2-2-4-2 Compton scattering

Compton scattering is the predominantinteraction of X-ray and gamma photons in the diagnostic energy range with soft tissue. In fact, Compton scattering not only predominates in the diagnostic energy range above 26keV in soft tissue, but continues to predominate well beyond diagnostic energies to approximately 30Mev.This interaction is most likely to occur between photons and outer shell electrons. The electron is ejected from the atom, and the photon is scattered with some reduction in energy. As with all types of interactions, both energy and momentum must be conserved. Thus the energy of the incident photon (E_0) is equal to the sum of the energy of the scattered photon (E_{SC}) and the kinetic energy of the ejected electron (E_e), the binding energy of the electron that was ejected is comparatively small and can be ignored.

$$E_0 = E_{Sc} + E_e \qquad (2-3)$$

Compton scattering results in the ionization of the atom and a division of the incident photon energy between the scattered photon and ejected electron; the ejected electron will lose its kinetic energy via excitation and ionization of atoms in the surrounding material. The Compton scattered photon may traverse the medium without interaction or may undergo subsequent interactions such as Compton scattering, photoelectric absorption, or Rayleigh scattering. The energy of the scattered photon can be calculated from the energy of the incident photon and the angle of the scattered photon.

$$E_{sc} = \frac{E_0}{1 + \frac{E_0}{511KeV}(1 - \cos\theta)} \qquad (2-4)$$

Where:

$E_{sc} \equiv$ the energy of the scattered photon.

$E_0 \equiv$ the incident photon energy.

$\theta \equiv$ the angle of the scattered photon.

As the incident photon energy increases, both scattered photons and electrons are scattered more toward the forward direction. In X-ray transmission imaging, these photons are much more likely to be detected by the image receptor, thus reducing image contrast. In addition for given scattering angle, the fraction of energy transferred to the scattered photon decreases with increasing incident photon energy. Thus, for higher energy incident photons, the majority of the energy is transferred to the scattered electron. When Compton scattering does occur at the lower X-ray energies used in diagnostic imaging (18 to 150kev) ,the majority of the incident photon energy is transferred to the scattered photon which, if detected by the image receptor, contributes to image degradation by reducing the primary photon attenuation differences of the tissues. For example, following the Compton interaction of an 80 keV photon, the minimum energy of the scattered photon is 61 keV. Thus, even with maximal energy loss, the scattered photons have relatively high energies and tissue penetrability. The laws of conservation of energy and momentum place limits on both scattering angle and energy transfer. For example the maximal energy transfer to the Compton electron occurs with a 180-degree photon backscatter. In fact, the maximal energy of the scattered photon is limited to 511 keV at 90 degrees scattering and to 255 keV for 180-degree scattering (backscatter) event. These limits on scattered photon energy hold even for extremely high-energy photons (e.g., therapeutic energy range).The scattering angle of the ejected electron cannot exceed 90 degrees, whereas that of the scattered photon can be any value including a 180-degree backscatter. In contrast to the scattered photon, the energy of the ejected electron is usually absorbed near the scattering site. The incident photon energy must be substantially greater than the electron's binding energy before a Compton interaction is likely to take place. Thus, the probability of a Compton interaction increases, compared to Rayleigh scattering or photoelectric absorption, as the incident photon energy increases. The probability of Compton interaction also depends on the electron density. The probability of Compton scattering per unit volume is approximately proportional to the density of the material. Compared to other elements, the absence of neutrons in the hydrogen atom results in an approximate doubling of electron density.Thus, hydrogenous materials have a higher probability of Compton scattering than a non-hydrogenous material of equal mass (Jerrold et al; 2002).

2-2-4-3 The photoelectric effect

The photoelectric effect is a phenomenon in which a photon interacts with an atom and ejects one of the orbital electrons from the atom. In this process, the entire energy of the photon is first absorbed by the atom and then transferred to the atomic electron. The kinetic energy of the ejected electron called the photoelectron is equal to $h\upsilon - E_b$ where E_b is the binding energy of the electron.Interactions of this type can take place with electrons in the K, L, M, or N shells (Faiz et al, 2003)

In the photoelectric effect, all of the incident photon energy is transferred to an electron, which is ejected from the atom. The kinetic energy of the ejected photoelectron (E_e) is equal to the incident photon energy (E_0) minus the binding energy of the orbital electron (E_b).

$$E_e = E_0 - E_b \qquad (2-5)$$

In order for photoelectric absorption to occur, the incident photon energy must be greater than or equal to the binding energy of the electron that is ejected. The ejected electron is most likely one whose binding energy is closest to, but less than, the incident photon energy. For example, for photons whose energies exceed the K-shell binding energy, photoelectric interactions with K-shell electrons are most probable. Following a photoelectric interaction, the atom is ionized, with an inner shell electron vacancy. This vacancy will be filled by an electron from a shell with a lower binding energy. This creates another vacancy, which, in turn, is filled by an electron from an even lower binding energy shell. Thus, an electron cascade from outer to inner shells occurs.

The difference in binding energy is released as either characteristic X-rays or auger electrons. The probability of characteristic X-ray emission decreases as the atomic number of the absorber decreases and thus does not occur frequently for diagnostic energy photon interactions in soft tissue (Jerrold et al; 2002)

2-2-4-4 The pair production

Pair production can only occur when the energies of X-ray and gamma rays exceed 1.02 MeV. In pair production, an X or gamma ray interacts with the electric field of the nucleus of an atom. The photon's energy is transformed into an electron-positron pair. The rest mass energy equivalent of each electron is 0.511MeV Photon energy in excess of this threshold is imparted to the electrons as kinetic energy. The electron and positron lose their kinetic energy via excitation and ionization. When the positron comes to rest, it interacts with a negatively charged electron, resulting in the formation of two oppositely directed 0.511Mev annihilation photons.

Pair production is of no consequence in diagnostic X-ray imaging because of the very high energies required for it to occur. In fact; pair production does not become significant unless the photon energies greatly exceed the 1.02 Mev energy thresholds (Jerrold et al; 2002)

2-2-5 Interactions of Ionizing Radiation:

When an x- ray beam passes through a medium, interaction between photons and matter can take place with the result that energy is transferred to the medium. The initial step in the energy transfer involves the ejection of electrons from the atoms of the absorbing medium. These high-speed electrons transfer their energy by producing ionization and excitation of the atoms along their paths. If the absorbing medium consists of body tissues, sufficient energy may be deposited within the cells, destroying their reproductive capacity.

2-2-5-1 Ionizing Radiation:

Ionizing (high-energy) radiation has the ability to remove electrons from atoms; i.e., to ionize the atoms. Ionizing radiation can be electromagnetic or particulate radiation (Fig. 2.1). Clinical diagnostic radiation uses photons (electromagnetic) as radiation in the diagnosis of disease and some benign conditions (UNSCEAR. 2000)

Table2-4

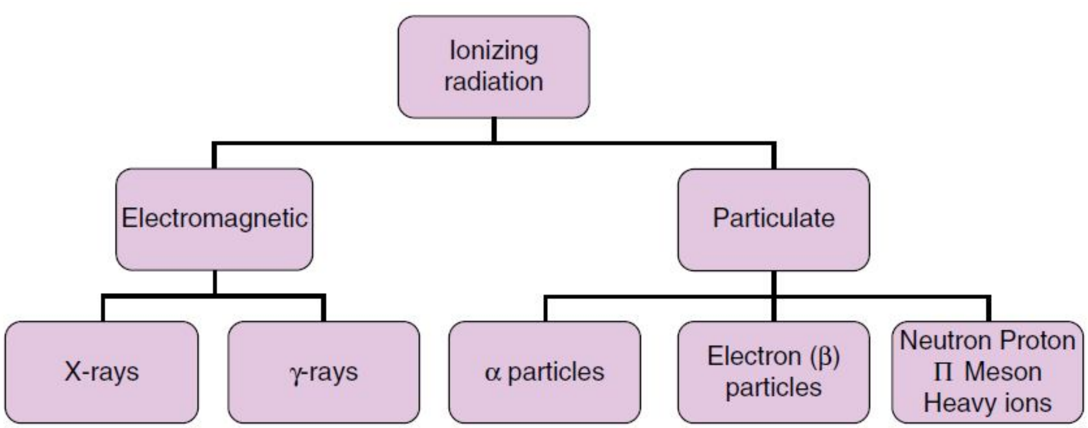

Table2-4 Shows the types of Ionizing radiation

2-2-5-2 Biological Effect of Ionizing Radiation:

The collective to the population resulting from medical exposure has been estimated to represent the largest single man-made contribution to both the somatic and the genetically significant dose equivalent in European countries. A similar distribution was found in other developed countries. The largest contribution being from Diagnostic Radiology, which estimated at ten times the sum of contributions from nuclear medicine and radiotherapy. It is desirable to reduce the patient dose in diagnostic radiology to minimum with good image quality to keep the dose as low as reasonably achievable (ALARA principle)

Biological effects may occur in the exposed individual as somatic radiation damage to the cell or alternatively this genetic damage could be passed on to individual's descendants. The effects are classified into stochastic and deterministic effects.

2-2-5-3 Stochastic Effects:

Short-term effects may arise soon after exposed to radiation. The effect may subside quickly. These effects generally manifest many years, even decades, after the radiation exposure (and were once called “late effects”). Their major characteristics, in direct contrast with those for non stochastic effects

- A threshold may not be observed.
- The probability of the effect increases with dose.
- It is difficult definitively to associate the effect with the radiation exposure.

2-2-5-4 Deterministic effects:

In deterministic effect there is a threshold dose of radiation below, which the effect does not occur. Doses significantly above the threshold will produce effects which severity is proportional to the absorbed dose received. Short-term effects are: radiation erythema, gastro-intestinal syndrome where the cell renewal system is damaged [13]. The major identifying characteristics of non stochastic Effects are:

- There is a threshold of dose below which the effects will not be observed.
- Above this threshold, the magnitude of the effect increases with dose.
- The effect is clearly associated with the radiation exposure.

2-2-5-5 Entrance surface air kerma (ESAK)

The entrance surface air kerma (ESAK) is defined as the kerma in air at the point where the central radiation beam axis enters the hypothetical object, i.e. patient or phantom, in the absence of the specified object.

The entrance surface dose, or alternatively the entrance skin dose (ESD) is defined as the absorbed dose to air on the x-ray beam axis at the point where x-ray beam enters the patient or a phantom, including the contribution of the backscatter. The ESD is to be expressed in mGy. Some confusion exists in the literature with regard to the definition of the ESD. That is, whether the definition should refer to the absorbed dose to the air as defined above or absorbed dose to tissue

2-2-6 Equivalent dose HT:

Accounts for biological effect per unit dose

$$H_T = W_R x D$$

Table2-5 Radiation weighting factors (W_R):

Radiation type and energy range	weighting factors (W_R):
Photons (X-rays and gamma-rays) all Energies	1
Electron all Energies Neutrons	1
<10 keV	5
10-100 keV	10
>100 kev to 2 MeV	20
2-20 MeV	10
>20 MeV	5
Protons >20MeV	5
Alpha particles , Fission fragments	20

2-2-6-1 Effective dose

Risk related parameter, taking relative *radiosensitivity* of each organ and tissue into account:

$E\ (Sv) = \Sigma_T\ W_T \times H_T$

W_T : tissue weighting factor for organ T

H_T: eequivalent dose received by organ or tissue

Table 2-6 Tissue and organ weighting factors (UNSCEAR 2008) :

weighting factors for different organs			
Tissue	Tissue weighting factors		
	ICRP 30(136) 1979	ICRP 60(13) 1991	ICRP 103(16) 2008
Gonads	0.25	0.20	0.08
Red bone marrow	0.12	0.12	0.12
Colon	-	0.12	0.12
Lung	0.12	0.12	0.12
Breast	0.15	0.05	0.12
Esophagus	-	0.05	0.04
Thyroid	0.03	0.05	0.04
Skin	-	0.01	0.01
Bone surfaces	0.03	0.01	0.01
Salivary glands	-	-	0.01
Brain	-	-	0.01
Remainder	0.30	0.05	0.12

2-2-6-2 Radiation Units:

Roentgen:

The roentgen is a unit used to measure a quantity called exposure. It can only be used to describe an amount of gamma and X-rays, and only in air. One roentgen is equal to depositing in dry air enough energy to cause 2.58 x 10-4 coulombs per kg. It is a measure of the ionizations of the molecules in a mass of air. The main advantage of this unit is that it is easy to measure directly, but it is limited because it is only for deposition in air, and only for gamma and x-ray

(UNSCEAR 2000).

2-2-6-3 Radiation absorbed dose (Rad):

The rad is a unit used to measure a quantity called absorbed dose. This relates to the amount of energy actually absorbed in some material, and is used for any type of radiation and any material. One rad is defined as the absorption of 100 ergs per gram of material. The unit rad can be used for any type of radiation, but it does not describe the biological effects of the different radiations.

2-2-6-4 Rem (roentgen equivalent man):

The rem is a unit used to derive a quantity called equivalent dose. This relates the absorbed dose in human tissue to the effective biological damage of the radiation. Not all radiation has the same biological effect, even for the same amount of absorbed dose. Equivalent dose is often expressed in terms of thousandths of a rem, or mrem. To determine equivalent dose (rem), you multiply absorbed dose (rad) by a quality factor (Q) that is unique to the type of incident radiation.

2-2-6-5 Gray (Gy):

The gray is a unit used to measure a quantity called absorbed dose. This relates to the amount of energy actually absorbed in some material, and is used for any type of radiation and any material. One gray is equal to one joule of energy deposited in one kg of a material. The unit gray can be used for any type of radiation, but it does not describe the biological effects of the different radiations. Absorbed dose is often expressed in terms of hundredths of a gray, or centi-grays. One gray is equivalent to 100 rads.

2-2-6-6 Sievert (Sv):

The sievert is a unit used to derive a quantity called equivalent dose. This relates the absorbed dose in human tissue to the effective biological damage of the radiation. Not all radiation has the same biological effect, even for the same amount of absorbed dose. Equivalent dose is often expressed in terms of millionths of a sievert, or micro-sievert. To determine equivalent dose (Sv), you multiply absorbed dose (Gy) by a quality factor (Q) that is unique to the type of incident radiation. One sievert is equivalent to 100 rem (UNSCEAR; 2002).

2-2-7 Radiation measurements:

With respect to measurement, three separate features of an X-ray beam must be identified .The first consideration is the flux of photons travelling through air from the anode towards the patient .The ionization produced by this flux is a measure of the Radiation exposure. If expressed per unit area per second it is the intensity of more fundamental importance as far as the biological risk is concerned is the absorbed dose of radiation. This is a measure of the amount of energy deposited as a result of ionization processes (UNSCEAR; 2002).

2-2-7-1 Dose measurement:

There are several ways of measuring doses from ionizing radiation. Workers who come in contact with radioactive substances or may be exposed to radiation routinely carry personal dosimeters. In the United States, these dosimeters usually contain materials that can be used in thermo luminescent dosimeter (TLD) or optically stimulated luminescence (OSL). Outside the

United States, the most widely used type of personal dosimeter is the film badge dosimeter, which uses photographic emulsions that are sensitive to ionizing radiation. The equipment used in radiotherapy (linear particle accelerator in external beam therapy) is routinely calibrated using ionization chambers or the new and more accurate diode technology (UNSCEAR; 2002).

2-2-7-2 Ionization chamber:

In medical x-ray imaging the Free-in-air air kerma measurements are best made with suitably designed ionization chambers of typically between 0.6 and 180 cm3 volume. The chambers should have 'air equivalent' walls so that their energy response in terms of air kerma is substantially uniform for all relevant x-ray spectra. The leakage current should be very small compared with the ionization current produced by the minimum dose rate to be measured and the response should not be affected appreciably by ion recombination at high dose rates. Dosimeters should be calibrated in a manner traceable to a national primary standard of air kerma as described; there are special requirements for ionization chambers used for air-kerma measurements in mammography: these are a thin entrance wall to reduce attenuation at low photon energies, and ideally a structure that does not appreciably disturb the primary radiation field. Thin entrance window chambers with small volumes generally have a rather massive construction on the exit side, which implies that the charge produced in the cavity contains a significant contribution from scattered radiation

2.2 X-ray imaging system

An X-ray imaging system consist of an X-ray source or generator (x-ray tube),an image detection system which can be a film (analog technology) or a digital capture system, Xray photons are produced by an electron beam that is accelerated to a very high speed and strikes with a target. The electrons that make up the beam are emitted from a heated cathode filament. The electrons are then focused and accelerated by an electrical field to wards and angled anode target. The point where the electron beam strikes the target is called the focal spot. Most of the kinetic energy contain in electron beam is converted to heat, but around 1% of the energy is converted into X-ray photons, the excess heat is dissipated via a heat sink. At the focal spot, X-ray photons are emitted in all direction from the target surface, the highest intensity being around 60deg to 90deg from the beam due to the angle of the anode target to the approaching X-ray photons. There is small round window in the X-ray tube directly above the angled target. This window allows X-Ray to exit the tube with little attenuation while maintaining a vacuum seal required for X-ray tube operation. X-ray machines work by apply controlled voltage and current to the X-ray tube, which results in a beam of X-ray. The beam is projected on matter. Some of the X-ray beam will pass through the object, while some are absorbed. The resulting pattern of radiation is then ultimately detected by a detection medium including rare earth screen (which surround photographic film).

2.3 Computed Radiography (CR)

Uses very similar equipment to conventional radiography except that in place of a film to create the imaging plate housed in a special cassette and placed under the body part or object to be examined and x-ray exposure is made. Hence, instead of taking an exposed film into a darkroom for developing in chemical tanks or an automatic film processor, the imaging plate is run through a special laser scanner, or CR reader , that reads and digitize the image the digital image can be viewed and enhanced using software that has functions very similar to other conventional digital image processing software, such as contrast, brightness , and filteration and zoom (Faiz et al, 2003)

2.4 Image Quality:

2.4.1 Spatial Resolution:

Spatial resolution refers to the minimum resolvable separation between high-contrast objects. In digital detectors, spatial resolution is defined and limited by the minimum pixel size. The image resolution of a radiographic system depends on several factors such as the size of the focal spot.

The anode tip should make a large angle with the electron beam to produce a nicely focused X-ray beam. Thicker patients cause more X-ray scattering, deteriorating the image resolution. Patient scatter can be reduced by placing a collimator grid in front of the screen. The grid allows only the photons with low incidence angle to reach the screen. The light scattering properties of the fluorescent screen. The film resolution, which is mainly determined by its grain size. For image intensifier systems and digital radiography, the sampling step at the end of the imaging chain is an important factor.

The resolving power (i.e., the frequency where the MTF is 10%) of clinical screen–film combinations varies from 5 up to 15 lp/mm. In most cases, spatial resolution is not a limiting factor in reader performance with film. For images with storage phosphors, a resolving power of 2.5 up to 5 lp/mm (at 10% contrast) is obtained. This corresponds to a pixel size of 200 to 100 µm, which is mostly sufficient except for mammography, for which more recent detector technology is needed. Depending on the size of the object, it is clear that images with 2000 by 2000 pixels and even more are needed to obtain an acceptable resolution [Bushberg et.al 2002].

2.4.2 Contrast:

The contrast is the intensity difference in adjacent regions of the image. Image intensity depends on the attenuation coefficients μ(E, x) and thicknesses of the different tissue layers encountered along the projection line. Because the attenuation coefficient depends on the energy of the X-rays, the spectrum of the beam has an important influence on the contrast. Soft radiation yields a higher contrast than hard radiation. Another important factor that influences the contrast is the absorption efficiency of the detector, which is the fraction of the total radiation hitting the detector that is actually absorbed by it. Higher absorption efficiency yields a higher contrast. In systems with film, the contrast is strongly determined by the contrast of the photographic film. The higher the contrast, the lower the useful exposure ranges. In digital radiography, contrast can be adapted after the image formation by using a suitable gray value transformation. Note however that such a transformation also influences the noise, thus keeping the CNR unchanged [Bushberg et.al 2002].

2.4.3 Noise:

Quantum noise, which is due to the statistical nature of X-rays, is typically the dominant noise factor. A photon-detecting process is essentially a Poisson process (the variance is equal to the mean). Therefore, the noise amplitude (standard deviation) is proportional to the square root of the signal amplitude, and the SNR also behaves as the square root of the signal amplitude. This explains why the dose cannot be decreased unpunished. Doing so would reduce the SNR to an unacceptable level. Further conversions during the imaging process, such as photon–electron conversions, will add noise and further decrease the SNR.

To quantify the quality of an image detector the measure detective quantum efficiency (DQE) is often used. The image detector is one element in the imaging chain and to quantify its contribution to the SNR, the DQE is used, which expresses the signal-to-noise transfer through the detector. The DQE can be calculated by taking the ratio of the squared SNR at the detector output to the squared SNR of the input signal as a function of spatial frequency. It is a measure of how the available signal-to-noise ratio is degraded by the detector. Several factors influence the DQE, particularly the absorption efficiency of the detector, the point spread function of the detector and the noise introduced by the detector [Bushberg et.al 2002].

2.4.4 Artifacts:

Although other modalities suffer more from severe artifacts than radiography, X-ray images are generally not artifact free. Scratches in the detector, dead pixels, unread scan lines, inhomogeneous X-ray beam intensity (heel effect), afterglow, etc., are not uncommon and deteriorate the image quality [Bushberg et.al 2002].

2.4.5 Modulation Transfer Function:

Modulation transfer function (MTF) is the capacity of the detector to transfer the modulation of the input signal at a given spatial frequency to its output [Spahn 2005]. At radiography, objects having different sizes and opacity are displayed with different gray-scale values in an image. MTF has to do with the display of contrast and object size. More specifically, MTF is responsible for converting contrast values of different-sized objects (object contrast) into contrast intensity levels in the image (image contrast). For general imaging, the relevant details are in a range between 0 and 2 cycles/mm, which demands high MTF values.

MTF is a useful measure of true or effective resolution, since it accounts for the amount of blur and contrast over a range of spatial frequencies. MTF values of various detectors were measured and further discussed by Illers et.al [Illers et.al 2005].

2.5 IVU TECNIQUE

2.5.1 Definition

Intravenous urography is a test that uses X-ray and a special dye to help assess the kidneys, ueters , bladder and uretra. The kidneys excrete the contrast into the urine, which becomes visible when x rayed (radio opaque), creating images of urinary collection system. The procedure has several variation and many names, including: Intravenous Pyelography (IVP), Intravenous urography (IVU) and execratory urography.

2.5.2 Clinical indication for IVU

The urinary tract does not show up well on ordinary X-ray pictures . however , with inteavenous urography a contrast dye is injected into a vein (intravenous injection) the dye travels in the bloodstream , concentrates n the kidneys, and is passed out into the ureters with urine made by the kidneys. A series of X-ray pictures is then taken over your tummy (abdomen), usually every 5-10 minutes. You stay on couch between each X-ray picture, but you may be asked to get up to empty your bladder befor the final X-ray picture is taken. The procedure usually takes about 30-60 minutes. However some pictures may be taken hours later in certain circumstances.

The X-ray pictures produced are called an intravenous urogram (IVU),but can also be called an intravenous pyelgram(IVP).

An intravenous urogram is order to demonstrate the structure and function of the kidney, ureters, and bladder. patients complainingof abdominal pain radiating to the back may require this exam to rule out kidneys stones.

Previous study

The first study deal with this issue has been conducted by **Suliman et.al (2007).** The aim of the authors in this study was to evaluate the entrance surface doses (ESDs) to patients undergoing selected diagnostic X-ray examinations in major Sudanese hospitals. That work was carried out in four major hospitals in the Sudanese capital Khartoum. Eight X-ray units were included in the study. ESD per examination was estimated from X-ray tube output parameters in four hospitals comprising eight X-ray units and a sample of 346 radiographs. The mean ESDs were found to be within the established international reference doses.

Another study was performed by: Suleiman A et.al (2014) to determine the Entrance Surface dose (ESD) for adults patients undergoing common X- ray examinations in two Hospitals in Khartoum, namely Khartoum teaching hospital and academy teaching hospital. The study was performed in four X-ray machines. A total of 191 patients were included in this study. Patient's data such as (age and weight) and exposure parameters (kV and mAs) were recorded. The results of ESD have been obtained with the use of the Dose Cal software which developed by the radiological protection center in saint gorges hospital London.. The results obtained in this work, range from (10.3) for lumbosacral lat to (0.004) for Elbow, was not exceeding the reference value and also the values obtained by previous studies. But when comparison made between the four machines using some selected tests, the mean dose value at Khartoum teaching hospital by (shimadzu (1)) was found to be higher than other machines. This may be due the fact that the machine is old one and also it is output is greater than outputs of other machines, but in general the competency of technicians in Khartoum teaching hospital is less than in the academy teaching hospital, and also the number of patients in this hospital is more than academy hospital.

The third study by : H. Osman et.al (2013) The main aim of the current study was to determine Entrance Surface Dose (ESD) to paediatric patients as the result of imaging procedure, in main paediatric hospital in Taif city –Saudi Arabia for the first time. 110 patients underwent different examinations (chest, abdomen, skull, and extremities), age range from 0-15 years .The patients bio data (age, weight, height, gender) were recorded. The exposure factors, focal skin distance, tube output and back scatter factor were entered in special software known by DOS CAL in order calculate the ESDs. The mean ESD obtained ranged 0.18 -0.32 mGy per radiograph for different ages and groups. The results for paediatric radiation dose were agreed and compatible with literature.

According to A.K. Sam et.al (2010) patient's radiation dose in routine X-ray examinations in Omdurman teaching hospital Sudanwas evaluted.110 patients was examined (134) radiographs in two X-ray rooms. Entrance surface doses (ESDs) were calculated from patient exposure parameters using Dose Cal software. The mean ESD for the chest, AP abdomen, AP pelvis, thoracic spine AP, lateral lumber spine, anteroposterior lumber spine, lower limb and for the upper limb were; 231±44 μGy,453±29 μGy, 567±22μGy, 311±33 μGy,716±39 μGy, 611±55μGy, 311±23 μGy, and 158±57 μGy, respectively. Data shows asymmetry in distribution. The results of were comparable with previous study in Sudan.

Several study were done to estimate the enter skin dose. Chin Med J ,et,al,practical was founded the value of intravenous urography combined with add-on CT in diagnosing ureteral abnormalities, intravenous urography (IVU) can help to provide more diagnostic information for determining the localization and nature of ureteral abnormalities with less irradiation dose .this study aimed to determine the value of IVU-CT for diagnosis of ureteral disease , where IVU is an insufficient to determine the diagnosis .method two hundred and eighty patents underwent IVU for suspected ureteral disorders, which identified a diagnosis in 184 cases and was insufficient for definite diagnosis in 96 cases designated as indeterminate diagnosis .Subsequently 90 patents (six patient declined CT) with indeterminate diagnosis consented to undergo immediate or delayed helical CT scan . the CT scan data were transferred to the workstation for post- processing ,and the cost and mean effective dose for each imageing method were calculated and compared indirectly .they conclude IVU-CT provides valuable information fo r the localization and diagnosis of ureteral abnormalities and may be considered as an efficient ,cost-effective and low-dose diagnostic technique in this setting (D.Tack 2007).

Another study by: M.A. Halato et al ,2008 estimated the entrance skin dose ESDs for patients undergoing selected diagnostic X- ray examinations in two large public hospitals in Khartoum state ,Sudan.

The study included the examinations of the chest postero-anterio (PA), skull anterio–posterior (PA) skull lateral (LAT), lumber spine AP/LAT , abdomen intravenous urogram (IVU)and pelvic AP .totally ,241 patents were included in this study .ESDs were estimated from patients specific exposure parameters using established relation between output (μG/mAS) and tube voltage (Kvp).the estimated ESDs ranged from 0.18 -1.05mGy for chest PA, 0.98 – 3.48 mGy for skull (AP) ,0.66 – 2.75 mGy for skull (LAT) ,1.22 – 5.35 mGy for abdomen (IVU), 1.18 -5.75 mGy for pelvis , 1.52 – 5.01 mGy lumber spine AP and 1.48 – 10.41 mGy for lumber spine (LAT). These values compare well with the international reference dose levels. This study provides additional data that can help the regulatory to authority to establish reference dose level for diagnostic radiology in Sudan.

Chapter three

Materials & Methods

3.1Materials:

3.1.1 population study

This study was conducted at Military hospital and Ibn Sinaa, data were collected in the period from January 2015 – March 2015

A total of 50 patients underwent intravenous urogram were examined.

3.1.2 Machine use

In the present study, major x-ray machine were used

- Manufacturer : Sibmbns
- Manufacturing date :1985
- Focal spot: small (0.6) mm/large (1.2) mm.
- Maximum KV: 150 KVp.
- Maximum mA 500 mA.
- Year of installation: 1990.

For Ibn Sinaa hospital:

- Manufacturer : Shimadzu
- Manufacturing date : 1983
- Focal spot: small () mm/large () mm.
- Maximum KV: 150 KVp.

- Maximum mA : 500 mA.

- Year of installation: 1984

3.1.3 ESD calculations

ESD which is defined as the absorbed dose to air at the center of the beam including backscattered radiation, measured for all patients using mathematical equation in addition to output factor and patient exposure factors. The exposure to the skin of the patient during standard radiographic examination or fluoroscopy can be measured directly or estimated by a calculation to exposure factors used and the equipment specifications from formula below.

$$ESD = OPx \left(\frac{kV}{80}\right)^2 xmAsx \left(\frac{100}{FSD}\right)^2 BSF$$

Where:

(OP) is the output in mGy/ (mA s) of the X-ray tube at 80 kV at a focus distance of 1 m normalized to 10 mA s, (kV) the tube potential,(mA s) the product of the tube current (in mA) and the exposure time(in s), **(FSD)** the focus-to-skin distance (in cm).

(BSF) the backscatter factor, the normalization at 80 kV and 10 mAs was used as the potentials across the X-ray tube and the tube current are highly stabilized at this point. BSF is calculated automatically by the Dose Cal software after all input data are entered manually in the software. The tube output, the patient anthropometrical data and the radiographic parameters (kVp, mA s, FSD and filtration) are initially inserted in the software. The kinds of examination and projection are selected afterwards.

FSD is the focus –to- patient entrance surface distance and BSF is the backscatter factor, with a value of (1.29-1.35) in this study.

3.1.4 ESD mothods

The Kv and mAs was changed according to the type of examination and patient age .the ESD was used to estimate radiation dose to the patient for the following selected examination: abdominal, kidney, ureter, and bladder. The result were pick up with different figures , tables , groups and graphs using excel software of windows Microsoft and correlation found between kv and ESD , mAs and ESD and FSD and ESD. The results were picked up with different figures, tables, groups and graphs using excel software of windows Microsoft and correlation was found between FSD and ESD

Chapter four

Results

The following tables and figures represent the data obtained from 44 patients underwent routine x-ray examinations for (abdominal, ureter , kidney , bladder) with digital imaging in the x-ray department in Military hospital and Ibn Sinaa in Khartoum state.

Table 4-1: Summary of patients ' characteristics and technical parameters selected for the various examinations in two Governmental hospitals considered for the study.

Male patients	Female patients	Age (mean)	Kvp (mean)	mAs (mean)	FSD (mean)
18	26	39.02	79.34	31.15	82.98

Table 4-2: statistical exposure for selected X ray examination.

	Range	Minimum	Maximum	Mean
Kvp	27.00	62.00	89.00	75.8947
MAs	44.00	12.00	56.00	30.7105
FSD	11.00	78.00	89.00	82.7368
ESD		0.29	2.74	1.495

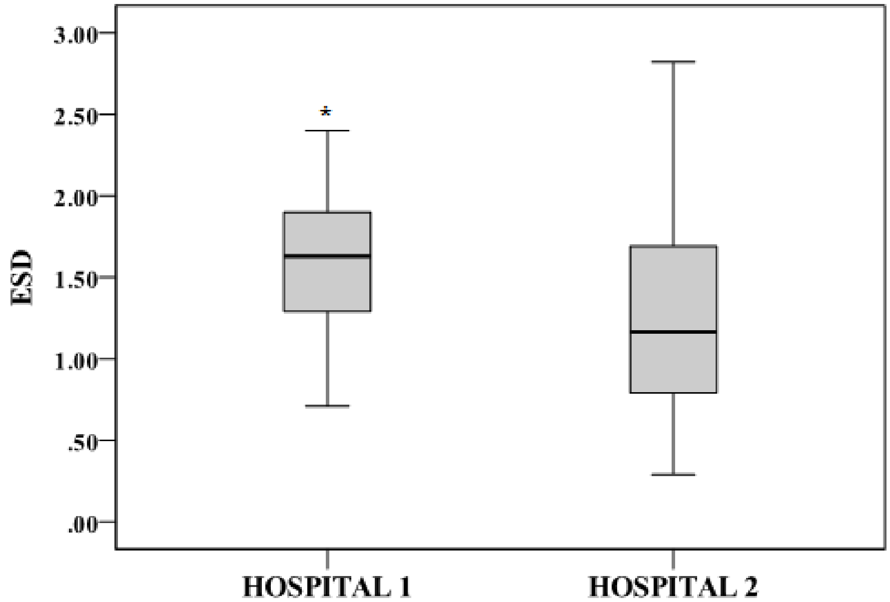

Figure 4-1: The difference between hospitals in ESD received by the patient

Table 4-3 : Correlations between Kv and ESD

		Kv	ESD
Kv	Pearson Correlation	1	.860**
	Sig. (2-tailed)		.000
	N	44	44
ESD	Pearson Correlation	.860**	1
	Sig. (2-tailed)	.000	
	N	44	44

****. Correlation is significant at the 0.01 level (2-tailed).**

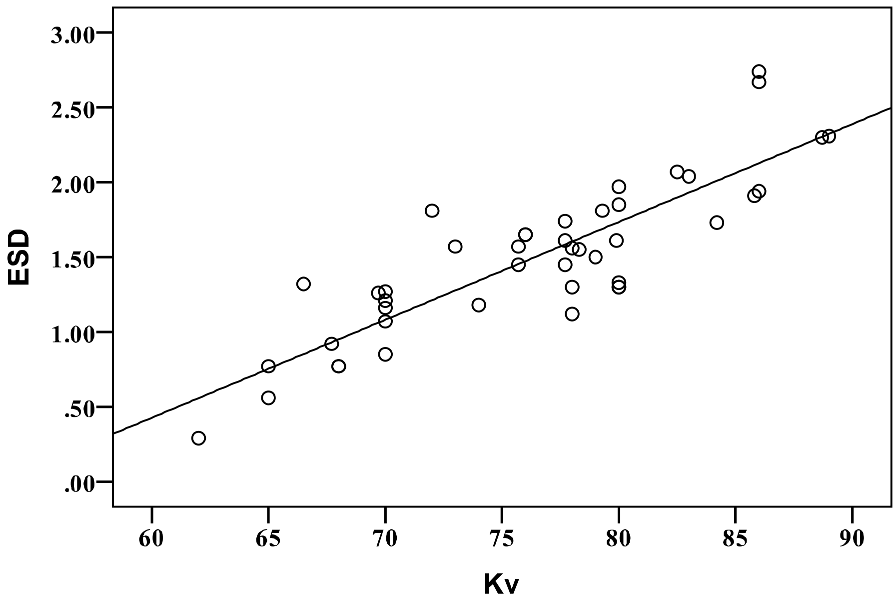

Figure 4-2: the relation between ESD and KV

Table 4-4 : Correlations between ESD and mAs

		ESD	mAs
ESD	Pearson Correlation	1	.844**
	Sig. (2-tailed)		.000
	N	44	44
mAs	Pearson Correlation	.844**	1
	Sig. (2-tailed)	.000	
	N	44	44

**. Correlation is significant at the 0.01 level (2-tailed).

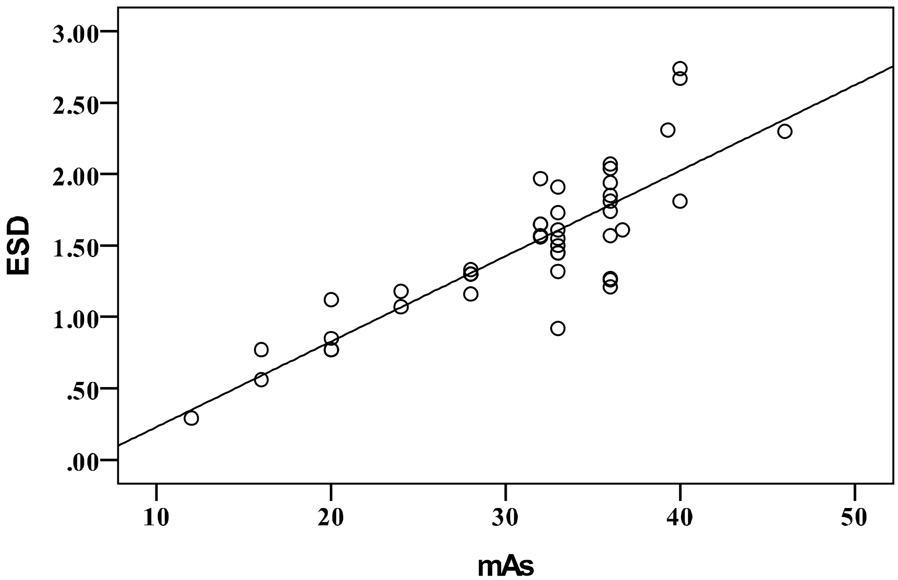

Figure 4-3: the relation between ESD & mAs

Table 4-5 : Correlations between ESD & FSD

		ESD	FSD
ESD	Pearson Correlation	1	-.791**
	Sig. (2-tailed)		.000
	N	44	44
FSD	Pearson Correlation	-.791**	1
	Sig. (2-tailed)	.000	
	N	44	44

****. Correlation is significant at the 0.01 level (2-tailed).**

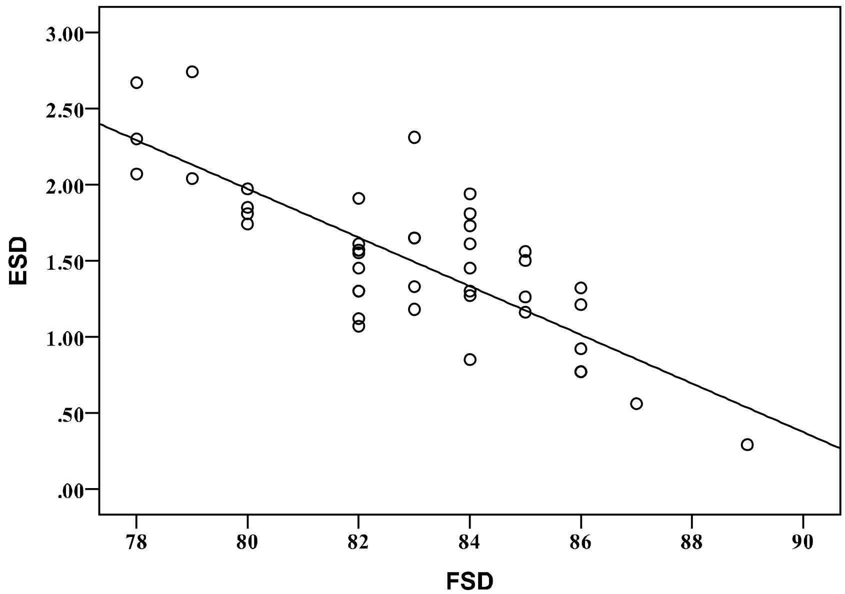

Figure 4-4: the relation between ESD & FSD

Table 4-6 Comparison of ESD with previous studies

	This study	SamuelW. Gordon	Abd almonim suliman
ESD	1.495± 0.29	0.99±7.65	3.0 ±1.5

Chapter five

Discussions, Conclusion and Recommendations

5.1 Discussion:

This study intended to estimate the radiation entrance skin doses for patients undergoing IVU examinations by digital radiography. It was anticipated that the study would help in the optimization of radiation protection of the patient. The data were collected from Military hospital and Ibn Sinaa hospital.

Shown in Table 4-2 is the estimated entrance surface dose (ESD) for all projections of IVU examination. For all examinations and projections, the ESD ranged from a minimum of 0.29 mGy/mAs to the maximum of 1.495 mGy/mAs. When comparing this ESD values with reference levels and with previous studies in Sudan and other countries, it revealed that the mean ESDs was relatively in agree with the previous studies (M.A. Halato et al, 2008).

The correlation between ESD and exposure factors (Kvp, mAs and FSD) has been carried out to show the effect of each factor on the ESD, it was clearly seen that ESD depend on exposure factors. Therefore, adjustment of exposure parameters according to dose saving protocols is necessary for optimization purposes. Dose variations could possibly be reduced to a minimum by using automatic exposure control (AEC) or exposure chart technique in condition where the AEC is not provided. This will ensure that the exposure parameters used will give radiation doses enough for the required image.

Inter-hospital variation was reported (figure 4-1), this is could be possibly due to the number of films taken and due to variations in patients' size. Furthermore the output of the machine could be contributed to this variation.

5.2 Conclusion:

The study has estimated the ESD for 44 patients who underwent IVU procedures in two hospitals in Sudan. The results of this dose survey provide essential data for patient dose levels for IVU radiography and the performance of the equipment used. Dose values for IVU radiography were accepted as compared with previous studies in Sudan or other countries.

ESD values depend on exposure parameters and number of films obtained. These findings support the importance of the ongoing quality assurance program to ensure that doses should be kept to a level consistent with optimum image quality.

5.3 Recommendation

- Computed Radiography operator must optimize the patient dose by use the best strategies available for reducing radiation dose to allow for mAs reduction in relation to patient size and weight.
- Implementation of Automatic Exposure Control System AECS by manufactures should be encouraged
- Computed Radiography must be used with high level of training for medical staff.

References:

AntonukL. E, Yorkston J, Huang W, et al. A real-time, flat-panel, amorphous silicon, digital x-ray imager. RadioGraphics1995

Bushong, S. C. (2001). Radiologic science for technologists. St. Louis: Mosby, ISBN 0-323-01337-6.

Ciraj, O., Markovi_c, S., & Ko_suti_c, D. (2005). First results on patient dose measurements from conventional diagnostic radiologys procedures in Serbia and Montenegro. Radiation Protection Dosimetry, 113(3), 330e335.

Choquette M, Demers Y, Shukri Z, et al. Performance of a real-time selenium-based x-ray detector for fluoroscopy. Proc SPIE2001.

Colbeth R, Boyce S, Fong R, et al. 40 × 30 cm flat-panel imager for angiography, R&F, and cone-beam CT applications. Proc SPIE2001.

Davidson RA. Radiographic contrast-enhancement masks in digital radiography. [PhD Thesis] New South Wales, Australia: The University of Sydney; 2006.

Davies, M., McCallum, H., White, G., Brown, J., & Helem, M. (1997). Patient dose audit in diagnostic radiography using custom designed software. Radiography, 3, 17e25. Diretrizes de Protecaeo Radiol_o gica.

European Commission. (1996). European guidelines on quality criteria for diagnostic radiographic images. EUR 16260 EN. Brussels. International Commission on Radiological Protection. (1991). Recommendations of the ICRP publication 60. Annals of ICRP. Oxford: Pergamon Press.

FAIZ M. KHAN, THE PHYSICS OF RADIATlON THERAPY-3thEd, Philadelphia (2003).

JERROLD, ANTHONY, EDWIN AND JOHN, THE ESSENTIALPHYSICS OF MEDICAL IMAGING -2th Ed, USA (2002).

Gallet J, Van Metter R. Image capture chain performance of Kodak's next generation of DIRECTVIEW DR Systems (Kodak Medical Systems white paper) [Online] Available at http://www.carestreamhealth.com/awp-dr.html. (Accessed 3 September 2009).

Illers H, Buhr E, Gunther-Kohfahl S, et al. Measurement of the modulation transfer function of digital X-ray detectors with an opaque edge-test device. RadiatProt Dosimetry 2005.

International Atomic Energy Agency. International basic safety standards for protection against ionizing radiation and for the safety of radiation sources. (1996). IAEA Safety Series no. 115 (Vienna: IAEA).

International Commission on Radiological Protection. (2002). Basic anatomical and physiological data for use in radiological protection reference values (pp. 3e4). ICRP Publication 89. Ann. ICRP 32.

International Commission on Radiological Protection-103. (2007). Recommendations of the international commission on radiological protection. Annals of ICRP. Oxford, UK: Pergamon Press.

J. T. Bushberg, J. A. Seibert, E. M. Leidholdt, J. M. Boone (2002) The Essential Physics of Imaging, 2ndedn.

KandarakisI, Cavouras D, Panayiotakis G. S, et al. Evaluating x-ray detectors for radiographic applications: a comparison of ZnSCdS:Ag with Gd2O2S:Tb and Y2O2S:Tb screens. Phys Med Biol1997.

KrugerR. A, Mistretta C. A, Crummy A. B, et al. Digital K-edge subtraction radiography. Radiology1977.

Kramer, R., Khoury, H. J., & Vieira, J. W. (2008). CALDose_X a software tool for the assessment of organ and tissue doses, effective dose and cancer risk in diagnostic radiology. Physics in Medicine and Biology, 53, 6437e6459.

Ma, C. M., Coffey, C. W., DeWerd, L. A., Liu, C., Nath, R., Seltzer, S. M., et al. (2001). AAPM protocol for 40-300 kV X-ray beam dosimetry in radiotherapy and radiobiology. Medical Physics, 28(6), 868e889.

OvittT. W, Christenson P. C, Fisher H. D 3rd, et al. Intravenous angiography using digital video subtraction: x-ray imaging system. AJR Am J Roentgenol1980.

P. Allisy, R. J. Williams (2008) Farr's Physics for Medical Imaging 2ndedn. Philadelphia.

P. Suetens (2009) Fundamentals of MedicalImaging 2ndedn. Cambridge University.

PuigS. Digital radiography of the chest in pediatric patients [in German]. Radiologe2003.

Padovani, R., Contento, G., & Fabretto, M. (1987). Patient doses and risks from diagnostic radiology in north-east Italy. The British Journal of Radiology, 60, 155e165.

Revised radiation doses for typical Xray examinations: report on a recent review of doses to patients from medical X-ray examinations in the UK by NRPB. British Journal of Radiology, 70, 437 - 439

. Shrimpton, P. C., Wall, B. F., & Jones, D. G. (1986). Doses to patients from routine diagnostic X-ray examinations in England. The British Journal of Radiology, 59, 749e758.

Schaefer-Prokop C, De Boo D, Uffmann M, et al. DR and CR: Recent advances in technology. Eur J Radiol. 2009 (in press). [PubMed].

SpahnM. Flat detectors and their clinical applications. Eur Radiol2005.

Toivonen M, (2001) Patient dosimetry protocol in digital and interventional radiology Radiat. Port. Dosim.94,1-2,105-8.

United Nations Scientific Committee on the Effects of Atomic Radiation. (2000). Sources and effects of ionizing radiation. New York: United Nations. Wall, B. F., & Hart, D. (1997).

W. R. Hendee, E. R. Ritenour (2002) Medical Imaging Physics 4thedn. New York.

YaffeM. J, Rowlands J. A. X-ray detectors for digital radiography. Phys Med Biol1997.

ZhaoW, Rowlands J. A. X-ray imaging using amorphous selenium: feasibility of a flat panel self-scanned detector for digital radiology.