الآيـة

قـال تـعالـي:

اللَّهُ لَا إِلَٰهَ إِلَّا هُوَ الْحَيُّ الْقَيُّومُ ۚ لَا تَأْخُذُهُ سِنَةٌ وَلَا نَوْمٌ ۚ لَّهُ مَا فِي السَّمَاوَاتِ وَمَا فِي الْأَرْضِ ۗ مَن ذَا الَّذِي يَشْفَعُ عِندَهُ إِلَّا بِإِذْنِهِ ۚ يَعْلَمُ مَا بَيْنَ أَيْدِيهِمْ وَمَا خَلْفَهُمْ ۖ وَلَا يُحِيطُونَ بِشَيْءٍ مِّنْ عِلْمِهِ إِلَّا بِمَا شَاءَ ۚ وَسِعَ كُرْسِيُّهُ السَّمَاوَاتِ وَالْأَرْضَ ۖ وَلَا يَئُودُهُ حِفْظُهُمَا ۚ وَهُوَ الْعَلِيُّ الْعَظِيمُ ﴿٢٥٥﴾ لَا إِكْرَاهَ فِي الدِّينِ ۖ قَد تَّبَيَّنَ الرُّشْدُ مِنَ الْغَيِّ ۚ فَمَن يَكْفُرْ بِالطَّاغُوتِ وَيُؤْمِن بِاللَّهِ فَقَدِ اسْتَمْسَكَ بِالْعُرْوَةِ الْوُثْقَىٰ لَا انفِصَامَ لَهَا ۗ وَاللَّهُ سَمِيعٌ عَلِيمٌ ﴿٢٥٦﴾ اللَّهُ وَلِيُّ الَّذِينَ آمَنُوا يُخْرِجُهُم مِّنَ الظُّلُمَاتِ إِلَى النُّورِ ۖ وَالَّذِينَ كَفَرُوا أَوْلِيَاؤُهُمُ الطَّاغُوتُ يُخْرِجُونَهُم مِّنَ النُّورِ إِلَى الظُّلُمَاتِ ۗ أُولَٰئِكَ أَصْحَابُ النَّارِ ۖ هُمْ فِيهَا خَالِدُونَ ﴿٢٥٧﴾

صدق الله العظيم

Dedication

- *To my lovely parents and family, my mother the candle that burn to light others life.*
- *My father who did the best he can to make me what I became.*
- *To My family's love, patience, support and care were essential all the time to accomplish this search.*
- *To my friends and my colleagues.*
- *To my medical physics family.*

I dedicate this work to you

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Last but not the least, I would like to thank my parents, family and colleagues for supporting me all the way to accomplish this project.

I send deep appreciation and recognition to all of you.

Content

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Abstract

Imaging of the thoracolumbar area is difficult because of different composition and density in that region. The difference may cause non uniformity of optical density on radiograph. The aim of this research was to study the effect of compensating filter on image quality in lateral projection of thoracolumbar radiography. The study was performed by an X-ray unit exposed to the body phantom where different thicknesses of aluminum were used as compensating filter. The radiographs were processed by DICOM reader and being imported to KPACS software to analyze the pixel depth value, contrast and noise. it was found that the compensation filter increased radiographic image quality for thoracic lumber region by compensate the radiation energy that reach the thoracic region and appears dark without the filter and the lumber region which are low dense and shows good quality image and acceptable in image diagnosis without the filter.

It was also found that the appropriate filter thickness was 8 mm in the thoracic region and 2mm in the lumber region and Thickness of Aluminum compensating filter required to compensate the thorax and lumbar region was 5 mm.

Finally, the addition of aluminum compensating filter is advantageous in terms of efficiency which saving radiograph film, workload of the radiologic technologist and radiation dose to patient.

ملخص البحث:

يوجد صعوبه في التصوير الجانبي للمنطقة الصدريه والقطنيه بسبب اختلاف التركيب والكثافه في تلك المنطقه، هذا الاختلاف يمكن ان يسبب عدم انتظام الكثافه البصريه في صوره الأشعة.

الهدف من هذا البحث هو دراسة تاثير المرشح التعويضي على جوده الصوره الجانبيه في تصوير الفقرات الصدريه والقطنيه.

تم أداء الدراسه بواسطة تسليط جهاز الاشعه السينية على مجسم للمنطقه الصدريه والقطنيه بحيث تم استخدام سمك مختلف من الالمونيوم كمرشحات تعويض. وتمت معالجه صور الاشعه بواسطة قارئ يسمي Dicom وادخالها لبرنامج KPACS لتحليل قيمه عمق نقطه الشاشه وتباين الصوره والضوضاء في الصوره، ووجد ان مرشحات التعويض تزيد من جوده صوره الاشعه للمنطقه الصدريه والقطنية وذلك عن طريق تعويض طاقة الأشعة التي تصل للمنطقه الصدريه والتي تظهر مظلمه بدون مرشح والمنطقه القطنيه والتي لديها كثافه اقل وتظهر تفاصيل أكثر ومقبوله في التشخيص الطبي بدون مرشح. وأيضا وجد ان سمك المرشح المناسب لتصوير المنطقه الصدريه هو8 ملم وان سمك المرشح المناسب لتصوير المنطقه القطنيه هو 2 ملم وسمك المرشح المناسب لتصوير منطقتي الصدر والبطن معا هو5 ملم.

أخيرا وليس اخرا اضافه مرشحات التعويض المصنوعه من الالمونيوم هي ميزه لأنها تحافظ على فلم الصوره الاشعاعيه وتقلل حجم العمل على اخصائي الاشعه وتقلل الجرعه على المريض.

1. Chapter one

1.1 Introduction:

Diagnostic radiography is the practice of producing two-dimensional images using x-ray radiation. Radiographic exams are typically performed by Radiographers. Radiographers are trained, licensed medical professionals who specialize in the usage of radiographic positioning, patient care, and selection of technical factors, radiographic equipment, and radiation safety. Projection radiography is the cornerstone of modern medical imaging, and can be used to image almost every part of the human body, so producing quality radiographs with all the necessary diagnostic information is vital for patient diagnosis (Sutherland 2007).

Imaging of the thoracolumbar area is difficult because of different structure and density in that region. The difference may cause non uniformity of optical density on radiograph and hence effect on the contrast and the resolution (Vassileva J 2004).

Since the radiographic Image quality is depending on sharpness, contrast and the noise , the most significant issue in the radiographic image is the visibility of the details; we used the compensation filter to overcome this problem which removing a large proportion of the low energy photon before they reach the skin . This reduces the dose received by the patient while hardly affecting the radiation reach the film and then enhance the resolution, contrast and reduce the noise (Goncalves A et al, 2004).

There are several techniques to overcome the limitation in lateral thoracolumbar radiography such as using a high tube voltage technique, two exposure techniques and the use of compensating filter (Vassileva J 2004).

The usage of high kV technique is to give shorter exposure time. The higher the energy, the more photons will penetrate the body, the less that are absorbed. This will reduce the patient dose; but image quality is compromised due to reduced contrast of a preferentially forward scattered radiation (Martin C J 2007). The second technique used to overcome this issue is known as two exposure technique where the images may be taken twice with different exposure parameters. It produces two images in two exposures; one is thoracic and the other is lumbar. The disadvantage of this technique is doubling the dose to the patient since the exposure is taken twice, and it increases the time and workload of the radiographer (Almen A et al, 2000).

Filters are important for the purpose of patient protection and improving radiographic image quality (Shockley V E et al, 2008).

Compensating filter is a device, such as a wedge of aluminum, clay, or plastic that is placed over a body area during radiography to compensate for differences in radiopacity (Jerry Williams 2008). It is used in medical imaging to attenuate and hardened the x-ray beam spectrum increasing the thickness of the filter removes a great number of low energy photons that is absorbed in the body, thus reducing the radiation dose to patient (Uffmann M and Schaefer-Prokop C 2009).

The usage of compensating filter is effective as it is designed specifically for anatomical area that consists of varying thickness and density (Goncalves A et al, 2004). One advantage of compensating filter is anatomy of significant varying thickness can be imaged with a single exposure. Other advantages are quick and ease of use (Davidson R A 2001).

The research on the use of compensating filter in radiography has been conducted by several authors (Chamberlain C et al, 2000) ((Wieder S and Adams P L 1981). A special aluminum wedge filter has been used in scoliosis

radiography. The thickness of the filter is 19 mm in the chest region and 6.5 mm in the abdomen region (Chamberlain C C et al, 2000). Wieder and Adam use the trough filter made of aluminum for posteroanterior chest radiographs. It consists of a square plate of aluminum with a center trough that helps to produce high quality chest radiograph (Wieder S and Adams P L 1981).
Another study was conducted to study the effect of compensating filter on image quality in lateral projection of thoracolumbar radiography .This study was performed to verify the relationship between density, contrast and noise of lateral thoracolumbar radiography using various thickness of compensating filter and was conducted in Malaysia (N A A Daud et al, 2014).

In radiography, a filter that shields less dense areas to produce a more nearly uniform radiographic image .In this study, aluminum will be used as compensating filter as it reduces the proportion of low energy photons which is absorbed in the body. It is used due to its low atomic number thus a suitable filter material for the purpose of tissue compensation (Alcaraz M and Garcia-Vera M 2009).

A "diagnostic" radiograph is defined as one showing the tissues adequately penetrated, sharply outlined, and the variations in tissue opacity sufficiently demonstrated (Sutherland 2007). In image quality, radiographic density is the measure of overall darkening of the image. Density is a logarithmic unit that describes the ratio between light incident on the film and light being transmitted through the film. The higher the radiographic density shows the more opaque areas of the film (Watanabe PCA et al, 2007). The lower the density represents more transparent areas of the film. In term of pixel depth, the values varies from 0 (intensity for black pixel) to 255 (intensity for white pixel). In X-ray radiography, noise is the relevant comparison to contrast .In this study, noise properties of an image are described in terms of standard deviation measured within a region of interest (kitagawa H and Farman A2004).

1.2Problem of study:

Imaging body part such as lateral projection of thoracic lumber is challenging due to different structure and density of thoracic lumber area .this will lead to non-uniformity of optical density on radiograph and make it difficult to diagnosis.

1.3 Objectives of the study:

1.3.1General Objective:

To study the effect of compensation filter on the image quality in lateral projection of thoracic lumber radiography.

1.3.2 Specific objectives:

- To verify the relationship between density, contrast and noise of lateral thoracic lumber radiograph using various thickness of compensation filter.
- To determine the appropriate filter thickness with the thoracic lumber density.

1.4 Research Overview:

Chapter one is the introduction to this thesis. This chapter discusses the objectives and scope of work and introduces necessary background. It also provides an outline of the thesis.

Chapter two contains the background material for the thesis. Specifically it discusses the x-ray radiograph and the roll of compensation filter on image quality. This chapter also includes a summary of previous work performed in this field.

Chapter three describes the materials used to figure out the effect of compensation filter with different thickness on image quality and explains in details the methods used for imaging and measurement to accomplish this thesis.

Chapter four presents the results of this study.

Chapter five presents the discussion, conclusion, recommendations and references of the thesis.

2. Chapter Two

Literature review

2.1 X ray:

X-rays were discovered by Wilhelm Konrad Röntgen in 1895 while he was experimenting with cathode tubes. In these experiments, he used fluorescent screens, which start glowing when struck by light emitted from the tube. To Röntgen's surprise, this effect persisted even when the tube was placed in a carton box. He soon realized that the tube was emitting not only light, but also a new kind of radiation, which he called X-rays because of their mysterious nature. This new kind of radiation could not only travel through the box. Röntgen found out that it was attenuated in a different way by various kinds of materials and that it could, like light, be captured on a photographic plate (Paul Suetens 2009)

X-radiation (composed of X-rays) is a form of electromagnetic radiation. Most X-rays have a wavelength ranging from 0.01 to 10 nanometers, corresponding to frequencies in the range 30 petahertz to 30 exahertz (3×10^{16} Hz to 3×10^{19} Hz) and energies in the range 100 eV to 100 Kev (Novelline and Robert ,1997).

X-rays with photon energies above 5–10 Kev (below 0.2–0.1 nm wavelength) are called hard X-rays, while those with lower energy are called soft X-rays; Due to their penetrating ability, hard X-rays are widely used to image the inside of objects, e.g., in medical radiography(David Attwood 1999).

2.1.1Properties of x ray:

X-ray photons carry enough energy to ionize atoms and disrupt molecular bonds. This makes it a type of ionizing radiation, and therefore harmful to living tissue. A very high radiation dose over a short amount of time causes radiation sickness, while lower doses can give an increased risk of cancer. In medical imaging this increased cancer risk is generally greatly outweighed by the benefits of the examination. The ionizing capability of X-rays can be utilized in cancer treatment to kill malignant cells using radiation therapy (Denny P. P. et al, 1999).

Hard X-rays can traverse relatively thick objects without being much absorbed or scattered. For this reason, X-rays are widely used to image the inside of visually opaque objects. The penetration depth varies with several orders of magnitude over the X-ray spectrum. This allows the photon energy to be adjusted for the application so as to give sufficient transmission through the object and at the same time good contrast in the image (David et al ,2002).

2.1.2Interaction with matter:

X-rays interact with matter in three main ways, through photo absorption, Compton scattering, and Rayleigh scattering. The strength of these interactions depends on the energy of the X-rays and the elemental composition of the material. Photoelectric absorption is the dominant interaction mechanism in the soft X-ray regime and for the lower hard X-ray energies. At higher energies, Compton scattering dominates (Bushberg et al, 2002).

2.1.2.1Photoelectric absorption:

A photo absorbed photon transfers all its energy to the electron with which it interacts, thus ionizing the atom to which the electron was bound and producing a photoelectron that is likely to ionize more atoms in its path. An outer electron will fill the vacant electron position and produce either a characteristic photon or an electron. The probability of a photoelectric absorption per unit mass is approximately proportional to Z^3/E^3, where Z is the atomic number and E is the energy of the incident photon (Bushberg et al, 2002).

2.1.2.2Compton scattering:

Compton scattering is the predominant interaction between X-rays and soft tissue in medical imaging. Compton scattering is an inelastic scattering of the X-ray photon by an outer shell electron. Part of the energy of the photon is transferred to the scattering electron, thereby ionizing the atom and increasing the wavelength of the X-ray. The scattered photon can go in any direction, but a direction similar to the original direction is a bit more likely, especially for high-energy X-rays (Bushberg et al, 2002).

2.1.2.3 Classical or Coherent Scattering:

Low energy x-rays photon of about 10 Kev interact in this manner and Incident photon interacts with the atom. There is a change in direction, no loss of energy and no ionization and the Photon scattered forward (David Attwood 1999).

2.1.3 Production of x-ray:

Whenever charged particles (electrons or ions) of sufficient energy hit a material, x-rays are produced. X-rays can be generated by an X-ray tube, a vacuum tube that uses a high voltage to accelerate the electrons released by a hot cathode to a high velocity. The high velocity electrons collide with a metal target, the anode, creating the X-rays. In medical X-ray tubes the target is usually tungsten or a more crack-resistant alloy of rhenium (5%) and tungsten (95%) (Whaites et al, 2002).

The maximum energy of the produced X-ray photon is limited by the energy of the incident electron, which is equal to the voltage on the tube times the electron charge, so an 80 kV tube cannot create X-rays with an energy greater than 80 Kev. When the electrons hit the target, X-rays are created by two different atomic processes:

2.1.3.1Characteristic X-ray emission:

If the electron has enough energy it can knock an orbital electron out of the inner electron shell of a metal atom, and as a result electrons from higher energy levels then fill up the vacancy and X-ray photons are emitted. This process produces an emission spectrum of X-rays at a few discrete frequencies, sometimes referred to as the spectral lines. The spectral lines generated depend on the target (anode) element used and thus are called characteristic lines. Usually these are transitions from upper shells into K shell (called K lines), into L shell (called L lines) and so on (Bushburg et al, 2002).

2.1.3.2Bremsstrahlung:

This is radiation given off by the electrons as they are scattered by the strong electric field near the high-*Z* (proton number) nuclei. These X-rays have a continuous spectrum. The intensity of the X-rays increases linearly with decreasing frequency, from zero at the energy of the incident electrons, the voltage on the X-ray tube (Bushburg et al, 2002).

So the resulting output of a tube consists of a continuous bremsstrahlung spectrum falling off to zero at the tube voltage, plus several spikes at the characteristic lines. The voltages used in diagnostic X-ray tubes range from roughly 20 to 150 kV and thus the highest energies of the X-ray photons range from roughly 20 to 150 Kev(Bushburg et al, 2002).

Both of these X-ray production processes are inefficient, with a production efficiency of only about one percent, and thus most of the electric power consumed by the tube is released as waste heat. When producing a usable flux of X-rays, the X-ray tube must be designed to dissipate the excess heat (Camara et al, 2008).

2.2 X-ray detectors:

To produce an image from the attenuated X-ray beam, the X-rays need to be captured and converted to image information (Paul Suetens 2009).

2.2.1Screen–film system:

2.2.1.1Screen:

Photographic film is very inefficient for capturing X-rays. Only 2% of the incoming X-ray photons contribute to the output image on a film. This percentage of contributing photons corresponds to the probability that an X-ray photon (quantum) is absorbed by the detector. It is known as the absorption efficiency. The low sensitivity of film for X-rays would yield prohibitively large patient doses. Therefore, an intensifying screen is used in front of the film.

This type of screen contains a heavy chemical element that absorbs most of the X-ray photons. When an X-ray photon is absorbed, the kinetic energy of the released electron raises many other electrons to a higher energy state. When returning to their initial state they produce a flash of visible light, called scintillation. These light photons are scattered in all directions. Consequently, two intensifying screens can be used, i.e., one in front and one behind the film, to increase the absorption efficiency further. The portion of the light that is directed toward the film contributes to the exposure of the film.

X-ray intensifying screens consist of scintillating substances that exhibit luminescence. Luminescence is the ability of a material to emit light after excitation, either immediately or delayed (Paul Suetens 2009).

2.2.1.2 Film:

The film contains an emulsion with silver halide crystals (e.g., AgBr). When exposed to light, the silver Halide grains absorb optical energy and undergo a complex physical change. Each grain that absorbs a sufficient amount of photons contains dark, tiny patches of metallic silver called development centers. When the film is developed, the development centers precipitate the change of the entire grain to metallic silver. The more light reaching a given area of the film, the more grains are involved and the darker the area after development. In this way a negative is formed. After development, the film is fixed by chemically removing the remaining silver halide crystals. In radiography, the negative image is the final output image. Typical characteristics of a film are its graininess, speed, and contrast (Paul Suetens 2009).

2.2.2 Image intensifier:

An image intensifier works as follows:

A fluorescent screen converts the X-rays into visible light. The emitted light hits a photocathode, and the energy of the photons releases electrons from this cathode. A large potential difference between the cathode and the output accelerates the ejected electrons. The resulting electron beam is directed onto a small fluorescent screen by electrostatic or magnetic focusing and converted to light photons again. This focusing makes the system suitable to be coupled to a camera without any loss of light (Paul Suetens 2009).

Figure 2.1an x-ray tube

2.3Components of x-ray machine:

The complete radiographic imaging chain, which is consists of the following element:

2.3.1The X-ray source:

Basic elements of an x ray source assembly:

Anode: Also known as target electrode; the positively charged electrode in an X-ray tube.

Cathode: The negatively charged electrode in an X-ray tube (Paul Suetens 2009).

2.3.2Aluminum filter**:** often complemented by a copper filter. This filter removes low-energy photons, thus increasing the mean energy of the photon beam. Low-energy photons deliver doses to the Patient but are useless for the imaging process because they do not have enough energy to travel through the patient and never reach the detector.

Because low-energy photons are called soft radiation and high-energy photons hard radiation, this removal of low-energy photons from the beam is called beam hardening (Paul Suetens 2009).

2.3.3 A collimator: to limit the patient area to be irradiated (Paul Suetens 2009).

2.3.4A collimating scatter grid: This is a collimator that absorbs scatter photons. It stops photons with large incidence angle, whereas photons with small incidence angle can pass right through the grid. The grid can be made of lead, for example (Paul Suetens 2009).

2.3.5The detector: This can be a screen–film combination in which a film is sandwiched between two screens, an image intensifier coupled to a camera, a cassette containing a storage phosphor plate, or an active matrix flat panel detector (Paul Suetens 2009).

2.3.6The table: can be tilted in any orientation, from the horizontal to the vertical position (Paul Suetens 2009).

The X-ray system contains a tray for a conventional film-based or astorage phosphor cassette, as well as an image intensifier beneath the table (Paul Suetens 2009).

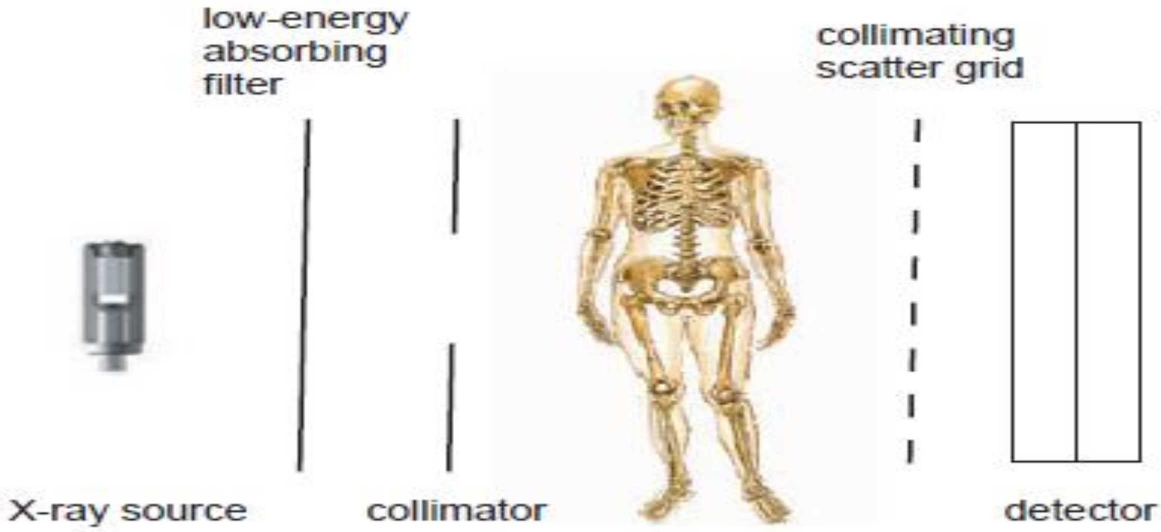

Figure 2.2 Schematic representation of the radiographic imaging chain (Paul Suetens 2009).

2.4 Radiographic image:

There are two basic ways to create images with x-radiation. One method is to pass an x-ray beam through the body section and project a shadow image onto the receptor. The second method, used in computed tomography (CT) (Spiegel PK 1995).

A radiograph is an X-ray image obtained by placing a part of the patient in front of an X-ray detector and then illuminating it with a short X-ray pulse. Bones contain much calcium, which due to its relatively high atomic number absorbs x-rays efficiently. This reduces the amount of X-rays reaching the detector in the shadow of the bones, making them clearly visible on the radiograph. The lungs and trapped gas also show up clearly because of lower absorption compared to tissue, while differences between tissue types are harder to see (**Roobottom** CA et al, 2010).

Radiographs are useful in the detection of pathology of the skeletal system as well as for detecting some disease processes in tissue. Traditional plain X-rays are less useful in the imaging of soft tissues such as the brain or muscle (Roobottom CA et al, 2010).

In medical diagnostic applications, the low energy (soft) X-rays are unwanted, since they are totally absorbed by the body, increasing the radiation dose without contributing to the image. Hence, a thin metal sheet, often of aluminum, called an X-ray filter, is usually placed over the window of the X-ray tube, absorbing the low energy part in the spectrum. This is called hardening the beam since it shifts the center of the spectrum towards higher energy (or harder) x-rays **(**Roobottom CA et al, 2010).

2.5Adverse effects (x-ray hazard):

Diagnostic X-rays (primarily from CT scans due to the large dose used) increase the risk of developmental problems and cancer in those exposed (Hall EJ, Brenner DJ2008). X rays are classified as a carcinogen by both the World Health Organization's International Agency for Research on Cancer and the U.S. government(Roobottom CA et al, 2010).

Experimental and epidemiological data currently do not support the proposition that there is a threshold dose of radiation below which there is no increased risk of cancer. It is estimated that the additional radiation will increase a person's cumulative risk of getting cancer by age 75 by 0.6–1.8%. The amount of absorbed radiation depends upon the type of X-ray test and the body part involved. CT and fluoroscopy entail higher doses of radiation than do plain X-rays *(*Brenner DJ and Hall EJ 2007).

To place the increased risk in perspective, An abdominal or chest CT would be the equivalent to 2–3 years of background radiation to the whole body, or 4–5 years to the abdomen or chest, increasing the lifetime cancer risk between 1 per 1,000 to 1 per 10,000(Giles D. et al, 1956).

The risk of radiation is greater to unborn babies, so in pregnant patients, the benefits of the investigation (X-ray) should be balanced with the potential hazards to the unborn fetus (Donnelly LF 2005).

2.6 X-ray Filters:

An X-ray filter is a device to block or filter out some or all wavelengths (low energy) in the X-ray spectrum. When a radiograph is taken, the lower energy photons in the x- ray beam are mainly absorbed by and deposit dose in the patient. Only a small fraction, if any, reaches the film and contributes to the image (Penelope Allisy et al, 2008).

The object of filtration is to remove a large proportion of the low energy photon before they reach the skin. This reduces the dose received by the patient while hardly affecting the radiation reach the film, and so the resulting image (Penelope Allisy et al, 2008).

This dose reduction is achieved by interposing between the x-ray tube and patient a uniform flat sheet of metal, usually aluminum, and called the added or additional filter .the predominant attenuation process in the filter should be photoelectric absorption. Because this varies inversely as the cube of the photon energy, the filter will attenuate the lower energy photons (mainly contribute to patient dose) much more than it does the higher energy photons (are mainly responsible for the image) (Penelope Allisy et al, 2008).

The x-ray photons produced in the target are initially filtered within the target itself, because they may be generated below its surface, and then by the window of the tube housing, the insulating oil and the glass insert. The combined effect of these disparate components' is expressed as an equivalent thickness of aluminum, typically1mm Al, and is called the inherent filtration. The total filtration is the sum of added filtration and inherent filtration for general diagnostic radiology, it should be at least 2.5mm Al equivalent (Penelope Allisy et al, 2008).

2.6.1 Choice of filter material:

The choice of filter material depends upon the choice of anode material in the X-ray tube as shown in the following table:

Anode	Cu	Co	Fe	Cr	Mo
Filter	Ni	Fe	Mn	V	Zr

From the table it can be seen that the ideal choice of material for an X-ray filter is a metal whose atomic number, *Z*, is one less than that of the anode target metal for first row transition metals (or two less for second row transition metals) (Jeremy Karl Cockcroft 2006).

The optimum thickness, *x* of the filter can be determined from the mass-absorption law:

$$I(\lambda) / I_o(\lambda) = \exp\{-(\mu / \rho)_\lambda \rho x\}$$

Where (μ / ρ) is the mass absorption coefficient at the wavelength λ, ρ is the density of the material, and $I(\lambda)$ and $I_o(\lambda)$ are the transmitted and incident X-ray intensities, respectively(Jeremy Karl Cockcroft 2006).

The atomic number should be sufficient high to make the energy dependent attenuating process, photoelectric absorption, predominate. It should not be too high, because the whole of useful x-ray spectrum should lie on the high energy side of the absorption edge. If not, the filter may actually soften the beam (Penelope Allisy et al, 2008).

2.6.2 K-edge filters:

Filter materials with K-edges in the higher energy part of the x- ray spectrum can be used .these remove both high and low energy x-rays but are relatively transparent to the energies just below the K-edges (Penelope Allisy et al, 2008).

2.6.3 Definition of compensating filter:

A device, such as a wedge of aluminum, clay, or plastic that is placed over a body area during radiography to compensate for differences in radiopacity.
In radiography, a filter that shields less dense areas to produce a more nearly uniform radiographic image (Penelope Allisy et al, 2008).

2.6.4 Compensating or wedge filter:

A shaped filter may be attached to the tube to make the exposure across the film more uniform and compensate for the large difference in transmission by, for example the upper and lower thoracic .these filters may be referred to as soft or wedges filters(Penelope Allisy et al, 2008).

Compensating filtration should be used in the following circumstances:

- for all AP full spine projections

Measure and compute technique for the thickest part of the abdomen and place the thick part of the filter in front of the thinner thorax. This avoids full spine films that are too dark at the top and too light through the pelvis at the bottom. (Lateral full spine projections should be avoided; rather, three separate sectional views should be produced to accompany the APFS projection [plus one additional AP cervical film; either APOM or 15° tilt-up] (Penelope Allisy et al, 2008).

- for virtually all AP thoracic projections

Procedure and rationale same as above. (The only exception to using a filter on an AP thoracic projection would be for a male patient with unusually developed pectoral muscles (as a body builder) and a very thin waist.) (Penelope Allisy et al, 2008).

- for most lateral thoracic projections

Measure side-to-side across the shoulders, on the outside of the arms, and compute technique sufficient to penetrate that thick and dense region. Then place the wedge filter inverted so that the thick part of the wedge attenuates the x-ray beam through the lower air-filled thorax, and the full amount of the beam is available to penetrate the dense upper thorax. This usually facilitates visualization of the difficult-to-see upper thoracic vertebrae without over penetrating the lower thoracic (Penelope Allisy et al, 2008).

- for most lateral lumbar projections on females

Measure side-to-side at widest part of hips and compute technique sufficient to penetrate that. Then place the wedge filter so that the thick part of the wedge attenuates the x-ray beam through the thinner waist, and the full amount of the beam is available to penetrate the dense pelvis. This avoids lateral lumbar films that are too dark at the top while being underpenetrated at the lumbosacral junction (Penelope Allisy et al, 2008).

- for all AP and oblique foot projections

Measure the thickest part of the foot, up near the ankle. Place wedge in front of the forefoot, thus avoiding over penetration of the toes, while allowing visualization of the tarsals (Penelope Allisy et al, 2008).

- for all swimmer's projections

Place filter to attenuate the beam through the cervical spine above the clavicle, while allowing visualization of the cervicothoracic junction (Penelope Allisy et al, 2008).

- for neutral AP shoulder projections

Avoids over penetrating the acromioclavicular joint while allowing visualization of the glenohumeral joint (Penelope Allisy et al, 2008)

2.7 Image quality:

The one comprehensive x-ray image quality characteristic is visibility. That is the visibility of anatomical structures, various tissues, and signs of pathology. Visibility depends on specific image characteristics that are:

2.7.1Resolution:

The image resolution of a radiographic system depends on several factors:

- 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.
- The patient. 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.

In most cases, spatial resolution is not a limiting factor in reader performance with film (Paul Suetens 2009).

2.7.2 Contrast:

The contrast is the intensity difference in adjacent regions of the image.

The image intensity depends on the attenuation coefficients 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, as used in mammography, yields a higher contrast than hard radiation (Paul Suetens 2009). 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 (Paul Suetens 2009).

A 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 range. In digital radiography, contrast can be adapted after the image formation by using a suitable gray value transformation (Paul Suetens 2009).

2.7.3 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 (Paul Suetens 2009).

2.7.4 Artifacts:

Although other modalities suffer more from severe artifacts than radiography, X-ray images are generally not artifact free. Scratches in the detector are not uncommon and deteriorate the image quality (Paul Suetens 2009).

2.8 Anatomy of vertebra column:

The human vertebral column is the vertebral column (backbone or spine) of the human skeleton, consisting of 24 articulating vertebrae and 9 fused vertebrae in the sacrum and the coccyx. The vertebrae in the column are separated from each other by intervertebral discs. It houses and protects the spinal cord in its spinal canal (Dorland's 2012).

There are normally thirty-three vertebrae; the upper twenty-four are articulating and separated from each other by intervertebral discs, and the lower nine are fused in adults, five in the sacrum and four in the coccyx or tailbone. The articulating vertebrae are named according to their region of the spine. There are seven cervical vertebrae, twelve thoracic vertebrae and five lumbar vertebrae. The number of vertebrae in a region can vary but overall the number remains the same. The number of those in the cervical region however is only rarely changed. There are ligaments extending the length of the column at the front and the back, and in between the vertebrae joining the spinous processes, the transverse processes and the vertebral laminae(Gray and Henry 1977)

2.8.1 Shape:

The upper cervical spine has a curve, convex forward, that begins at the axis (second cervical vertebra) at the apex of the odontoid process or dens, and ends at the middle of the second thoracic vertebra; it is the least marked of all the curves. This inward curve is known as a lordotic curve (Drake et al, 2005).

The thoracic curve, concave forward, begins at the middle of the second and ends at the middle of the twelfth thoracic vertebra. Its most prominent point behind corresponds to the spinous process of the seventh thoracic vertebra. This curve is known as a kyphotic curve (Drake et al, 2005).

The lumbar curve is more marked in the female than in the male; it begins at the middle of the last thoracic vertebra, and ends at the sacrovertebral angle. It is convex anteriorly, the convexity of the lower three vertebrae being much greater than that of the upper two. This curve is described as a lordotic curve(Drake, et al 2005).The sacral curve begins at the sacrovertebral articulation, and ends at the point of the coccyx; its concavity is directed downward and forward as a kyphotic curve (Drake et al, 2005).

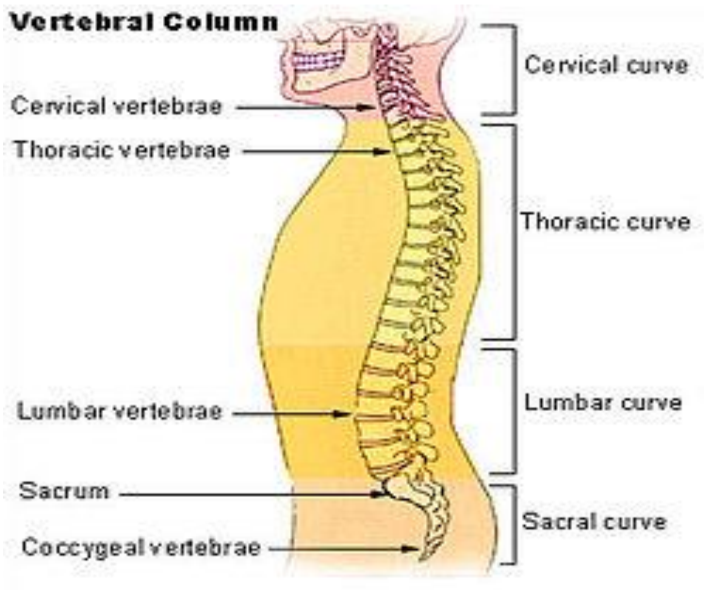

Figure2.3 the human vertebral column (Drake et al, 2005)

2.8.2 Thoracic Vertebrae:

Thoracic vertebrae form a transition between cervical vertebrae above and lumbar vertebrae below. The upper four thoracic vertebrae are like cervical vertebrae in some respects. They have vertically oriented articular facets and posteriorly directed spinous processes. The lower four thoracic vertebrae contain more lumbar features, like large bodies, robust transverse and spinous processes, and lateral projecting articular facets. The middle four thoracic vertebrae have characteristics between these two regions. These include vertically oriented articular processes and long, slender, and inferiorly inclined spinous processes (Dorland’s 2012)

The unique characteristic of thoracic vertebrae is articular facets for the ribs. Each vertebra contains two pairs of these costal demifacets on its body and one on each transverse process. Typical ribs articulate with the inferior demifacet and transverse process of a thoracic vertebra and the superior demifacet of the vertebra below it. The 11th and 12th thoracic vertebrae are sometimes considered atypical because they lack a superior costal demifacet. The 11th and 12th ribs thus articulate only with the 11th and 12th thoracic vertebrae, respectively (Dorland's 2012).

2.8.3 Lumbar Vertebrae:

The lumbar vertebrae consist of five individual cylindrical bones that form the spine in the lower back. These vertebrae carry all of the upper body’s weight while providing flexibility and movement to the trunk region. They also protect the delicate spinal cord and nerves within their vertebral canal (Patel et al, 2008).

Lumbar vertebrae are characterized by massive bodies and robust spinous and transverse processes. Their articular facets are oriented somewhat parasagittal, which is thought to contribute the large range of anteroposterior bending possible between lumbar vertebrae. Lumbar vertebrae also contain small mammillary and accessory processes on their bodies. These bony protuberances are sites of attachment of deep back muscles (Patel et al, 2008).

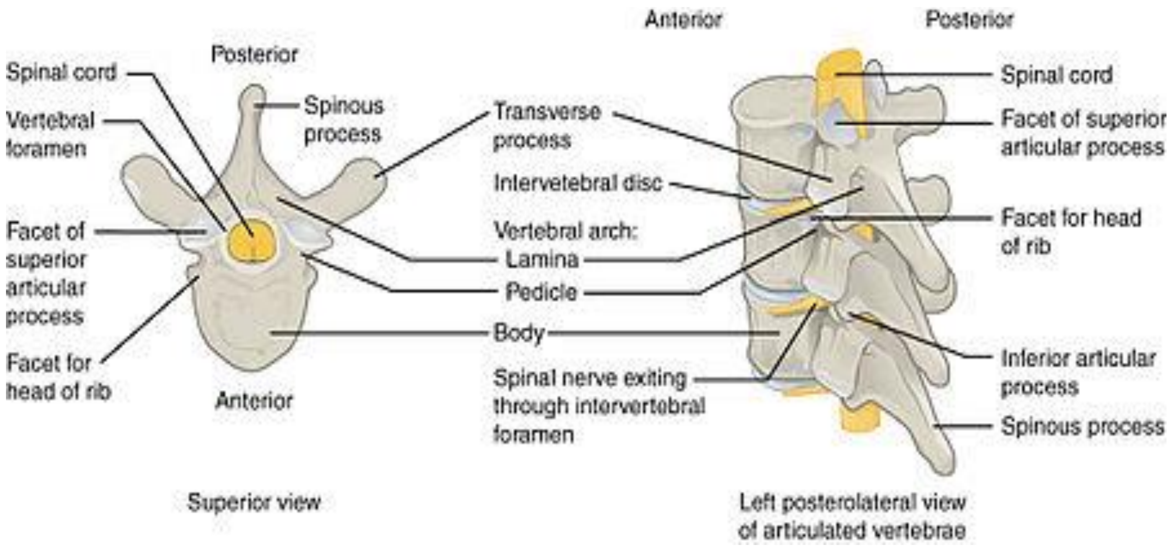

Figure 2.4 Anatomy of the vertebra (Drake et al, 2005)

2.9 Digital image processing:

Digital image processing is the use of computer algorithms to perform image processing on digital images(Solomon et al, 2010).

Image processing or digital image manipulation is one of the greatest advantages of digital radiography (DR). Preprocessing depends on the modality and corrects for system irregularities such as differential light detection efficiency, dead pixels, or dark noise. Processing is manipulation of the raw data just after acquisition. It is generally proprietary and specific to the DR vendor but encompasses manipulations such as unsharp mask filtering within two or more spatial frequency bands, histogram sliding and stretching, and gray scale rendition or lookup table application. These processing steps have a profound effect on the final appearance of the radiograph (Bushberg J et al, 2002)

Digital image processing allows the enhancement of the image's features of interest while attenuating details that are irrelevant in the given context, and then extract or increase the amount of useful information (Cho Z. H. et al 1993).
Digital image processing may be subdivided into three groups: preprocessing (low level of processing – preliminary operations used to filter noise, and to eliminate the unnecessary or unwanted details), data reduction (mid-level processing - extraction of useful information, segmentation), feature analysis (high level processing – ensuring semantic meaning extraction) (Gonzalez R. and Woods R.1992).
The basic task of digital image processing in medicine is source image enhancement. Images received from various medical devices (X-ray, Magnetic Resonance Imaging (MRI), X-Ray based Computer Tomography (CT) a.s.o.) - usually not very clear and containing noise - are filtered by using low level image processing algorithms (denoising, sharpening, edge detection) with parameter values tweaked for the specific problem and enlarging the amount of useful information offered to the medical specialist that interprets it. This is fully done by computers (Haralick R. M. and Shapiro L. G.1992).
The next processing operation is extraction of the useful information. On the enhanced image there are applied segmentation and edge linking algorithms for isolating certain body parts that are subject of interest (mid-level digital image processing algorithms). Simple analysis of the body parts can be done fully automatic (Russ J. C.1992).
One of the most complicated issues in digital image processing is automatic semantic meaning extraction from source images. This implies using of high-level digital image processing algorithms that are needed to improve medical diagnostic, operation planning and image guided surgery (Shung K. K.1992).
An already used technique is merging of several data acquisition and reconstruction of objects (3D object reconstruction from slices). An example of

using this technique is esthetical surgical planning (Sonka M. and Fitzpatrick J.2000).

Digital archiving of information offers a large number of advantages. Firstly, smaller space is needed for storing data. Secondly, the system allows flexible and easy access to data (Sonka M et al, 1999).

There are also many possibilities to interpret the stored information, easy observing of the evolution in time of patients with similar problems, the possibility to monitor the evolution in time of the body parts that are subject of interest for a patient .Another advantage offered by the digital storing of data is easy "sharing" of information (Webb S.1992).

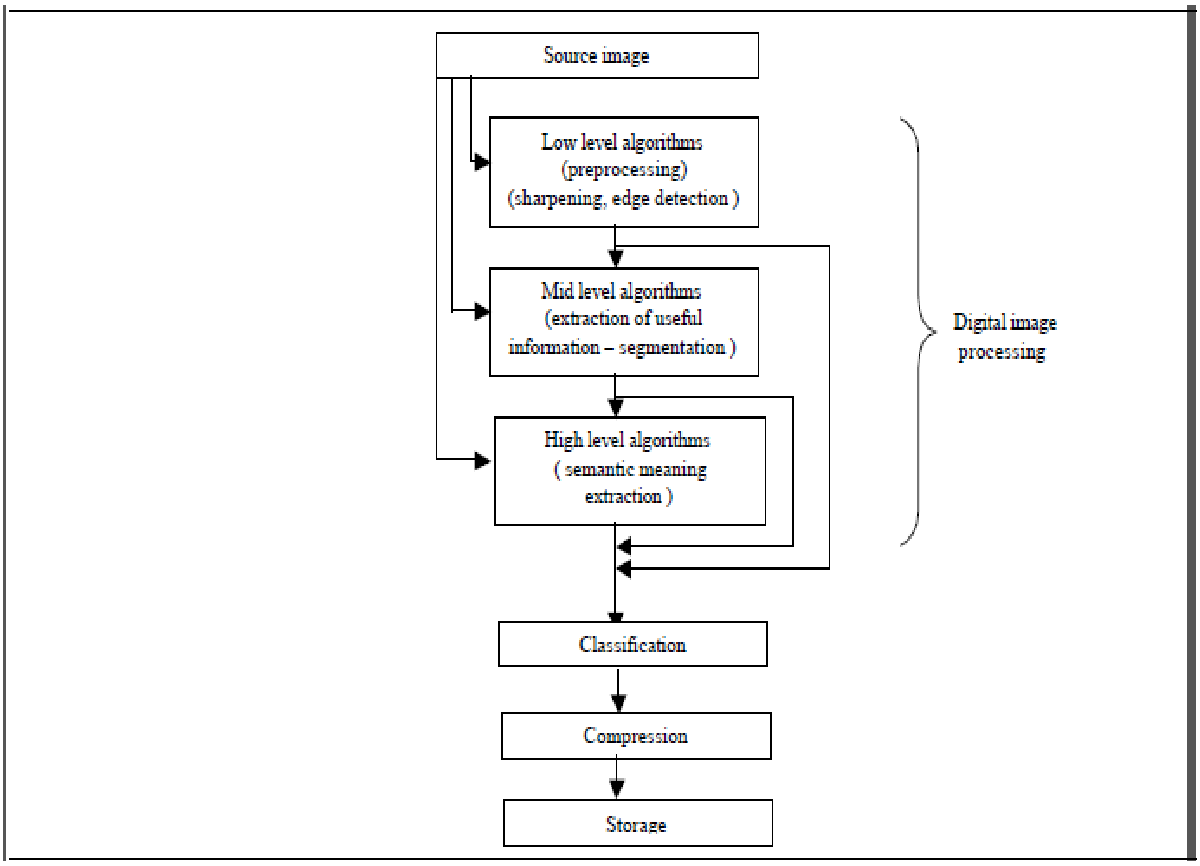

Figure 2.5 Digital image processing steps (Webb S.1992)

2.10 Previous study:

Various researches have been carried in the use of compensating filter in radiography by several authors. A study on the effect of compensating filter on image quality in lateral projection of thoracolumbar radiography was conducted to verify the relationship between density, contrast and noise of lateral thoracolumbar radiography using various thickness of compensating filter and to determine the appropriate filter thickness with the thoracolumbar density. The study was performed by an X-ray unit exposed to the body phantom where different thicknesses of aluminum were used as compensating filter. The radiographs were processed by CR reader and being imported to KPACS software to analyze the pixel depth value, contrast and noise. Result shows different thickness of aluminum compensating filter improved the image quality of lateral projection thoracolumbar radiography **(N** A A Daud et al, 2014).

A simple approach based on phantom measurements is proposed in this study to find the filtration, tube potential and ant scatter device that are optimal in respect of patient dose and image quality, at constant film–screen combination, film processing and viewing conditions. An original quasi-anthropomorphic chest phantom was exposed with 18 different beam qualities and three antiscatter devices. The entrance surface dose, organ doses and effective dose were estimated for each radiograph. The image quality was compared using two objective quality indexes, a contrast index and a scatter fraction, as well as two subjective indexes, a low contrast visualization index and a high contrast visualization index. It was found that for this X-ray unit, routinely using a 7:1 anti-scatter grid, the optimal imaging technique is added filtration of 0.1 mm Cu^{+1} mm Al at a tube potential 100 kvp (Vassileva J 2004).

Another study was performed Using aluminum trough filter for phototimed posteroanterior chest radiographs at 120 kvp provides excellent detail within the parenchyma of the lung as well as greatly enhanced visualization of mediastinal structures on a single exposure. An improvement in overall diagnostic value was noted by a panel of radiologists in 87% of 50 cases compared with conventional 120 kvp technique (Wieder S and Adams P L 1981)

A wedge-shaped aluminum filter was used in Radiation doses to patients undergoing scoliosis radiography to attenuated the X-ray beam in the"chest region" relative to the ``abdomen region "X-ray tube air kerma output factors (mGy $mAs^{-1}$) and half value layers (HVLs) were determined experimentally for the ``chest region" and ``abdomen region" (Chamberlain C C et al, 2000).

A study evaluated the performance of aluminum-copper alloy filtration, without the original aluminum filter, for dental radiography in terms of x-ray energy spectrum, air kerma rate and image quality. Comparisons of various thicknesses of aluminum-copper alloy in three different percentages were made with aluminum filtration. Tests were conducted on an intra-oral dental x-ray machine and were made on mandible phantom and on step-wedge. Depending on the thickness of aluminum-copper alloy filtration, the beam could be hardened and filtrated. The use of the aluminum-copper alloy filter resulted in reductions in air kerma rate from 8.40% to 47.33%, and indicated the same image contrast when compared to aluminum filtration (Goncalves A et al, 2004).

Chapter three

3. Materials and Method

3.1 Materials:

3.1.1 Phantom :

Thoracic lumber Phantom was placed in lateral projection. It is an anthropomorphic phantom consists skeleton and lungs. The positioning of lateral projection of thoracolumbar radiography was allowed using pillows.

3.1.2 Machine used :

The experiment was conducted in X-ray room examination of the Sudan University of science and technology faculty of medical radiological science. The study was performed using X-ray unit model Fuji film FDR smart with equivalent filtration of 3 mm aluminum, grid and cassette inside the table. The source to image distance (SID) was set to 100 cm.

3.1.3 Compensation filter:

The compensating filter was constructed from a single piece of aluminum of 8 cm x 8 cm dimensions. The aluminum thickness ranged between 1 mm to 11 mm.

3.2 Method:

3.2.1 Technique used:

A series of AEC exposure was made to determine the optimum kVp and mAs without the aluminum compensating filter. Another series of exposure then was made to determine the optimum kVp and mAs with the aluminum compensating filter. In this experiment the optimum kVp and mAs was 63 kVp and 12.8 mAs. The exposures parameters were kept constant throughout the experiment.

With the aid of a filter holder, the filter was placed in front of the light beam diaphragm while the compensating filter was inserted ranging from 1 mm thickness to 11 mm, in steps of 1 mm and a number of images were acquired with different thicknesses, so that the effect of increasing the thickness of aluminum compensating filter on image quality can be evaluated.

3.2.2 Measurements:

The images were processed and read by Dicom reader. The radiographs were imported to the KPACS software for the analysis purpose. The KPACS software was used to analyze the pixel value, contrast and noise of the images. The relationship between optical densities of thorax and lumbar with various thickness of compensating filter was determined and verified by the equation below:

$$\text{Optical density (OD)} = 2 - \left[220 - \frac{\text{pixel depth value}}{100}\right]$$

Besides, the Thickness of Aluminum compensating filter required to compensate the thorax and lumbar region was obtain by the equation:

$$\text{Appropriate filter thickness} = \frac{\text{Optical density of the thoracic}}{\text{Optical density of the lumber}} = 1$$

So the appropriate filter thicknesses with thoracic lumbar density was determined.

4. Chapter four

Results

In this chapter the results were presented to study the effect of compensation filter on the image quality in lateral projection of thoracic lumber radiography and to verify the relationship between density, contrast and noise of lateral thoracic lumber radiograph using various thickness of compensation filter. Compensation filter was used with different thickness after adjusting the kV and mA of an x-ray machine to a suitable value depending on the thickness of filters and the exposures parameters were kept constant throughout the experiment. Then the thoracic and lumber area of the phantom was imaged with each filter to find out the appropriate filter thickness with the thoracic lumber density and the results were as follow:-

Table 4.1 the relation between the thickness of the filter and the mean of the pixel value (density), stander deviation and the optical density.

Filters (mm)	T_8			T_9			T_{10}		
	Mean	SD	D_1	Mean	SD	D_2	Mean	SD	D_3
1	241.88	98.4375	2.41875	252	62.538	2.52	260.75	47.763	2.6075
2	225	90.5625	2.25	239.375	57.475	2.3938	244.5	56.3	2.5
3	234.38	103.288	2.34375	247	60.863	2.47	248.88	55.513	2.4888
4	222.13	94.7375	2.2212	248	49.775	2.48	256.13	52.163	2.5613
5	224.5	91.05	2.245	233.75	50.738	2.3375	246.75	50.888	2.4675
6	213.38	97.9875	2.13375	234.875	66.85	2.3488	248.63	54.225	2.4863
7	236.38	84.75	2.36375	253.375	33.525	2.5338	265.88	54.45	2.6588
8	247	94.0375	2.47	262.375	29.388	2.6238	265.13	96.2	2.6513
9	239.75	97.8375	2.3975	264.125	30.85	2.6413	267.13	111.09	2.6713
10	232.13	87.4625	2.32125	259.125	28.688	2.5913	279.75	96.713	2.7975
11	225.88	98.3375	2.25875	253.5	30.625	2.535	273.88	91.9	2.7388

Table 4.1 the relation between the thickness of the filter and the mean of the pixel value (density), stander deviation and the optical density (continue).

Filter (mm)	T_{11}			T_{12}			L_1		
	Mean	SD	D_4	Mean	SD	D_5	Mean	SD	D_6
1	270.5	54.15	2.705	271.25	43.438	2.7125	278.5	17.663	2.785
2	257.125	47.163	2.5713	258	37.063	2.58	269.25	22.25	2.6925
3	258.875	52.463	2.5888	260.25	36.413	2.6025	272.75	24.363	2.7275
4	261	43.675	2.61	267	39.85	2.67	272	31.888	2.72
5	260.375	42.263	2.6038	267.25	51.288	2.6725	269.25	28.613	2.6925
6	250.25	56.125	2.5025	260.38	34.4	2.6038	263.25	38.725	2.6325
7	270.125	68.375	2.7013	274.5	43.95	2.745	279.13	40.65	2.7913
8	281.375	43.375	2.8138	289.88	34.463	2.8988	293.38	25.463	2.9338
9	271.125	58.275	2.7113	285.88	48.713	2.8588	289.63	25.9	2.8963
10	280.125	46.625	2.8013	292	36.225	2.92	293.5	34.125	2.935
11	277.875	42.7	2.7788	281.63	54.725	2.8163	291.88	26.75	2.9188

	L_2			L_3		
Filters	Mean	SD	D_7	Mean	SD	D_8
1	287.88	31.85	2.8788	293.25	23.425	2.9325
2	281	25.525	2.81	283	32.075	2.83
3	286.13	61.475	2.8613	291.38	23.513	2.9138
4	282	32.338	2.82	283.5	31.613	2.835
5	282.25	26.438	2.8225	286.13	29.975	2.8613
6	273.75	35.888	2.7375	278.75	22.275	2.7875
7	283	33.375	2.83	286	24.313	2.86
8	303.75	26.725	3.0375	306.5	22.063	3.065
9	306	28.8	3.06	311.63	24.363	3.1163
10	305.5	27.075	3.055	307.88	24.813	3.0788
11	305.25	29.45	3.0525	308	26.713	3.08

Table 4.2 the relation between Thicknesses of the filters (mm) and mean pixel value for different region of Thorax and Lumbar.

Filters (mm)	T_8	T_9	T_{10}	T_{11}	T_{12}	L_1	L_2	L_3
	Mean	Mean	Mean	Mean	Mean	Mean	Mean	Mean
1	241.87	252	260.75	270.5	271.2	278.5	287.8	293.2
2	225	239.37	244.5	257.1	258	269.2	281	283
3	234.37	247	248.87	258.8	260.2	272.7	286.1	291.3
4	222.12	248	256.12	261	267	272	282	283.5
5	224.5	233.75	246.75	260.3	267.2	269.2	282.2	286.1
6	213.37	234.87	248.62	250.2	260.3	263.2	273.7	278.7
7	236.37	253.37	265.87	270.1	274.5	279.1	283	286
8	247	262.37	265.12	281.3	289.8	293.3	303.7	306.5
9	239.75	264.12	267.12	271.1	285.8	289.6	306	311.6
10	232.12	259.12	279.75	280.1	292	293.5	305.5	307.8
11	225.87	253.5	273.875	277.88	281.63	291.88	305.25	308

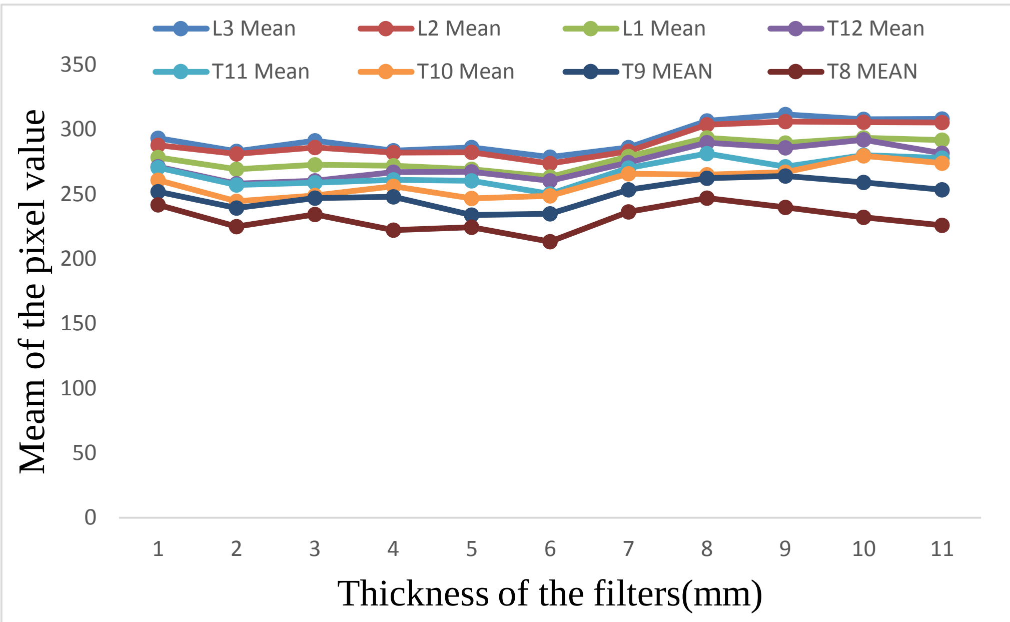

Figure 4.1 the relation between mean pixel value and thicknesses of the filters (mm) for different region of Thorax and Lumbar.

Table 4.3 the relation between thicknesses of the filters (mm) and optical density D for different region of Thorax and Lumbar.

Filters (mm)	T_8	T_9	T_{10}	T_{11}	T_{12}	L_1	L_2	L_3
	D_1	D_2	D_3	D_4	D_5	D_6	D_7	D_8
1	2.4187	2.52	2.607	2.705	2.712	2.785	2.87875	2.9325
2	2.25	2.3938	2.445	2.5713	2.58	2.6925	2.81	2.83
3	2.3437	2.47	2.488	2.588	2.602	2.727	2.8612	2.913
4	2.2212	2.48	2.561	2.61	2.67	2.72	2.82	2.835
5	2.245	2.337	2.467	2.603	2.672	2.692	2.8225	2.861
6	2.1337	2.348	2.486	2.502	2.603	2.632	2.7375	2.787
7	2.3637	2.533	2.658	2.701	2.745	2.791	2.83	2.86
8	2.47	2.623	2.651	2.813	2.898	2.933	3.0375	3.065
9	2.3975	2.641	2.671	2.711	2.858	2.896	3.06	3.116
10	2.3212	2.591	2.797	2.801	2.92	2.935	3.055	3.078
11	2.2587	2.535	2.738	2.778	2.816	2.918	3.0525	3.08

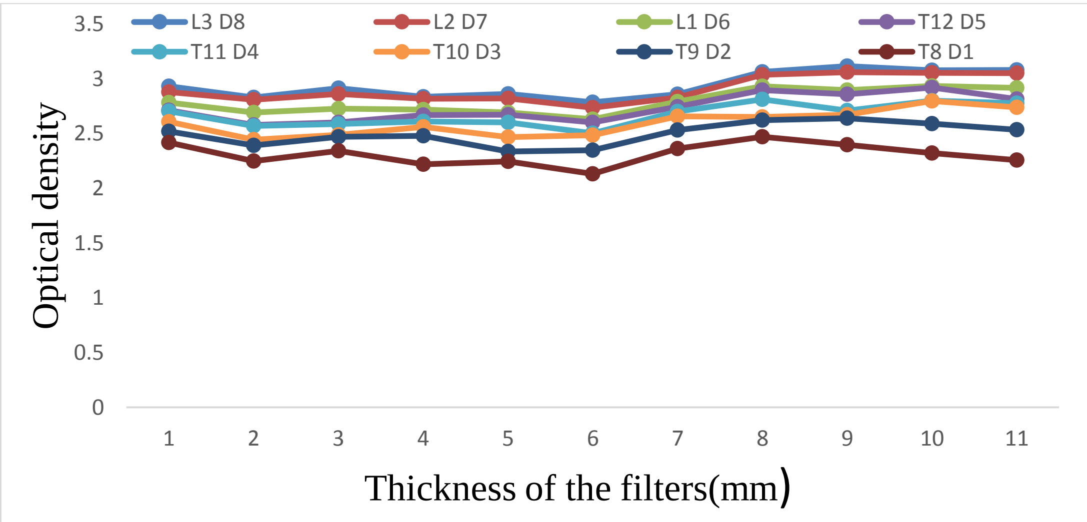

Figure 4.2 the relation between optical density D and thicknesses of the filters (mm) for different region of Thorax and Lumbar.

Table 4.4 the relation between thicknesses of the filters (mm) and stander deviation for different region of Thorax and Lumbar.

Filters (mm)	T_8	T_9	T_{10}	T_{11}	T_{12}	L_1	L_2	L_3
	SD	SD	SD	SD	SD	SD	SD	SD
1	9.8437	6.2537	4.7762	5.415	4.3437	1.766	3.185	2.3425
2	9.0562	5.7475	5.63	4.716	3.7062	2.225	2.552	3.2075
3	10.328	6.0862	5.5512	5.246	3.6412	2.436	6.147	2.3512
4	9.4737	4.9775	5.21625	4.367	3.985	3.188	3.233	3.1612
5	9.105	5.0737	5.08875	4.226	5.1287	2.861	2.643	2.9975
6	9.7987	6.685	5.4225	5.612	3.44	3.872	3.588	2.2275
7	8.475	3.3525	5.445	6.837	4.395	4.065	3.337	2.4312
8	9.4037	2.9387	9.62	4.337	3.446	2.546	2.672	2.2062
9	9.7837	3.085	11.108	5.827	4.871	2.59	2.88	2.4362
10	8.7462	2.868	9.6712	4.662	3.622	3.412	2.707	2.4812
11	9.83375	3.0625	9.19	4.27	5.4725	2.675	2.945	2.6712

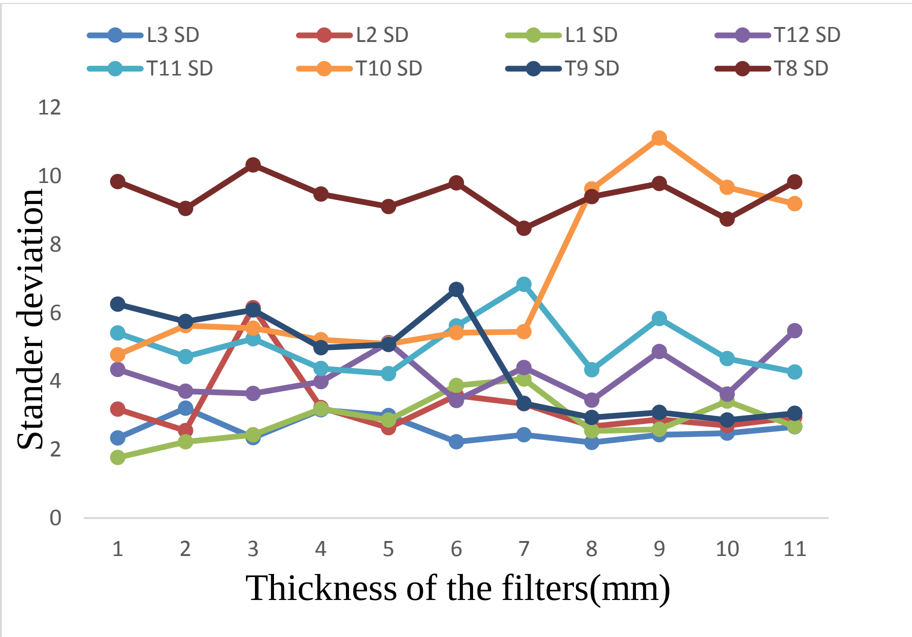

Figure 4.3 the relation between stander deviation and thicknesses of the filters (mm) for different region of Thorax and Lumbar.

Table4.5 the relation between the thickness of the filters and the ratio of the optical density of the thoracic and lumber.

Filters (mm)	T_{12}	L_1	D_5/D_6
	D_5	D_6	
1	2.7125	2.785	0.974
2	2.58	2.6925	0.9582
3	2.6025	2.7275	0.9542
4	2.67	2.72	0.9816
5	2.6725	2.6925	0.9926
6	2.6038	2.6325	0.9891
7	2.745	2.7913	0.9834
8	2.8988	2.9338	0.9881
9	2.8588	2.8963	0.9871
10	2.92	2.935	0.9949
11	2.8163	2.9188	0.9649

5. Chapter Five

5.1 Discussion:

This study was conducted to find the effect of compensation filter on the image quality in lateral projection of thoracic lumber radiography. Using an x-ray machine and compensation filters, the experiment was carried out by inserting the filters in front of the light beam diaphragm from 1mm thickness up to 11mm and the images were acquired for each filter.

Our results showed that For the Pixel value analysis, as the filter thickness increased, the value of pixel slightly increased (density reduced). There was sudden change of pixel values in figure 4.1 between 6 to 8 mm of aluminum thickness due to the characteristics of x-ray. At this point, the compensation was effective as the filter compensate more than others (see figure4.1). The results obtained in this study is in agreement with the previous study which state that the pixel value also increase with the thickness of the filter (N A A Daud et al, 2004).

The compensating filter absorbs radiation according to the thickness of the filter. Without compensating filter, pixel depth value (optical density) of lumbar L_1-L_3 is higher (low density) compared to the pixel depth value of thorax T_8-T_{12} (high density) means that the thorax region was too dark (black) while the lumbar region shows the good quality image and acceptable in image diagnosis. More filter thickness is required to compensate the thorax region to decrease its density while maintaining the density of the lumbar region. However, increasing too much thickness of the filter improved the image of thorax but reduced the image quality of lumbar (N A ADaud et al, 2004). Due to this reason we need to design one compensating filter with appropriate thickness in order to compensate both thorax and lumbar region so that the thoracic lumbar region can be imaged in one exposure (N A A Daud et al, 2004).

It’s clear that from the results of Optical density (pixel depth value)calculated at different vertebra of thoracic lumber that the optical density(pixel depth value) was increased at the thoracic region from T_8 up to $T_{12}$and at lumber region from L_1 up to L_3 as the filter thickness increased(see table 4.3 and figure 4.2) that means there was good compensation of the radiation by the filter between the thoracic and lumber region that makes the image showed better visibility of the bone structure at that area. the results obtained in this study is agree with the previous study which state that increasing thickness of compensation filter improved the image quality of thorax and lumbar region(N A A Daud et al,2004).

The results were illustrated for the standard deviation which represents the noise of an image. In relation to the contrast, noise increased as the contrast increased. The lower density difference means there is lower contrast. Contrast within a film increases with increasing density difference. our results show that in thorax region, the noise level slightly increased with increasing of filter thickness, in the lumbar region, noise decreased as the thickness of filter increased (see figure 4.3) this is due to the used of the filter and agree with the previous study which stated that noise decrease at the lumber region and increase in the thoracic region (N A A Daud et al, 2004).

The Thickness of Aluminum compensating filter required to compensate the thorax and lumbar region is the value which the ratio of optical density between thorax and lumbar equal to 1.

Our results were showed that the appropriate filter thickness was 8 mm in the thoracic region, 2 mm in the lumber region and Thickness of aluminum compensating filter required to compensate the thorax and lumbar region was 5mm.

Thickness of 8 mm was preferable in term of dose reduction in which it reduced more dose in this filter thickness than other, and this was agree with the previous study which state that the appropriate thickness of filter was 8 mm maximum in the thorax region but disagree with the value of appropriate filter in the lumber region which was1mm maximum in the lumbar region (N A A Daud et al, 2004).

The reason of the difference between the values of the appropriate filter thickness for imaging of the thoracic and lumber region of our study and the previous study may be due to the using of different phantom or the output of the x ray machine was not accurate or not quality controlled or due to the purity of the material filter that used.

5.2 Conclusion:

This study was performed to investigate the effect of compensation filter on the image quality in lateral projection of thoracic lumber radiography and it was found that the compensation filter increased radiographic image quality for thoracic lumber region by compensate the radiation energy that reach the thoracic region and appears dark without the filter and the lumber region which are low dense and shows good quality image and acceptable in image diagnosis without the filter.

However, increasing too much thickness of the filter improved the image of thorax but reduced the image quality of lumbar. .

There are many advantages when using compensating filter for lateral projection of thoracolumbar radiography. It is efficient in imaging the part of varying tissue thickness and density. It also improves the image quality of the radiography and saves the patient from being exposed to a large radiation; because the design of compensating filter allows the single exposure instead of two, therefore, saving the time and workload of the radiographer. In this study, it was also found that the appropriate filter thickness was 8 mm in the thoracic region, 2 mm in the lumber region and Thickness of aluminum compensating filter required to compensate the thorax and lumbar region was 5mm.

5.3 Recommendations:

- A research on the effect of other types and shapes of filters on the image quality and dose reduction purpose can be conducted.
- The possibility of manufacturing a filter from available and cheap materials.
- Studies about the effect of the compensation filter on the radiographic image at different organs and structure of the body like skull, limps and cervical can be carry out too.
- Investigation about the effect of the filter on the imaging of pediatric to reduce the kilo voltage, mill ampere and hence the dose for radiation protection purpose for pediatric.
- Study about the relation between the dose and filters can be carry out especially for the imaging of region that have more sensitivity of radiation.

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Appendix:

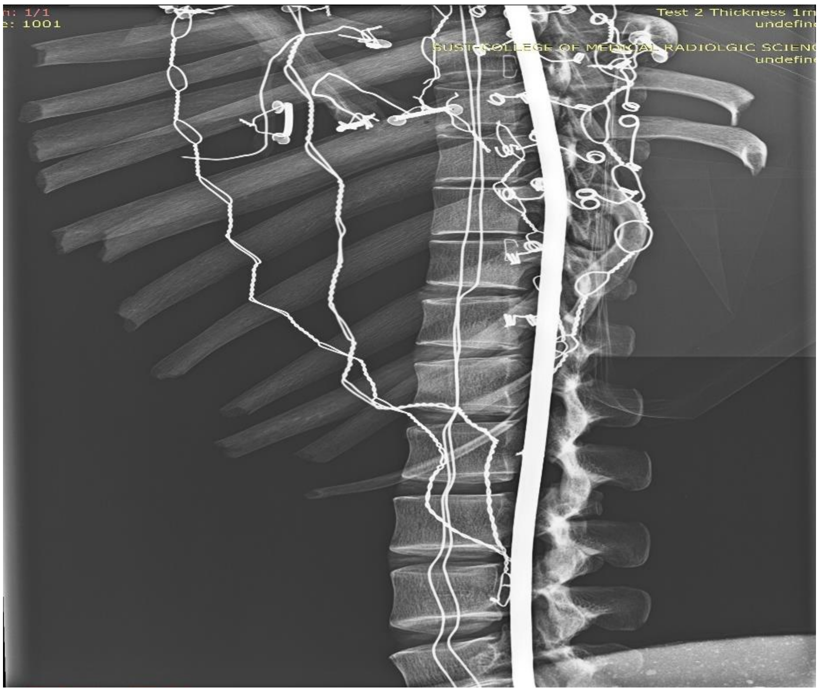

Image 5.1 for thoracic lumber region with 1mm thickness of AL filter

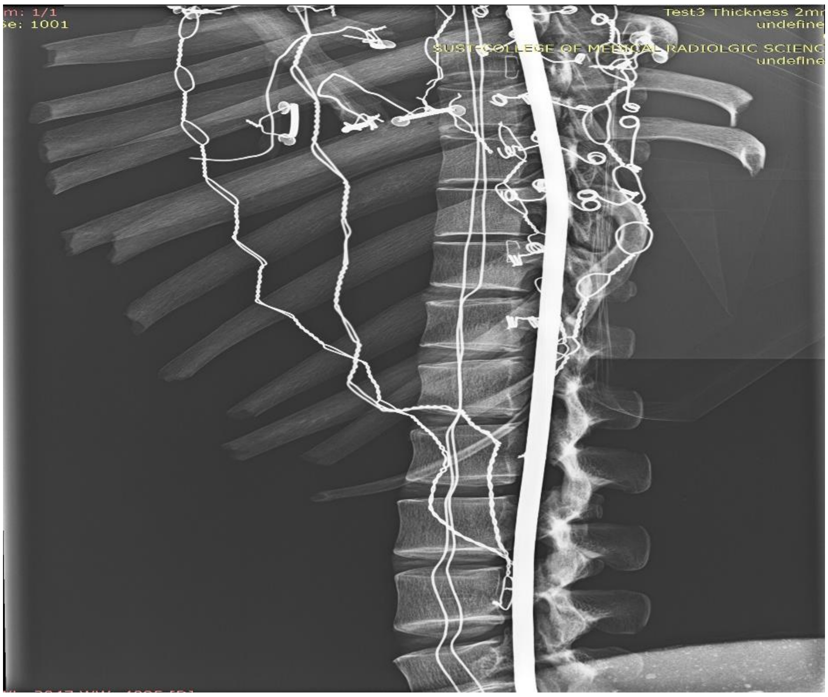

Image5.2 for thoracic lumber region with 2mm thickness of AL filter

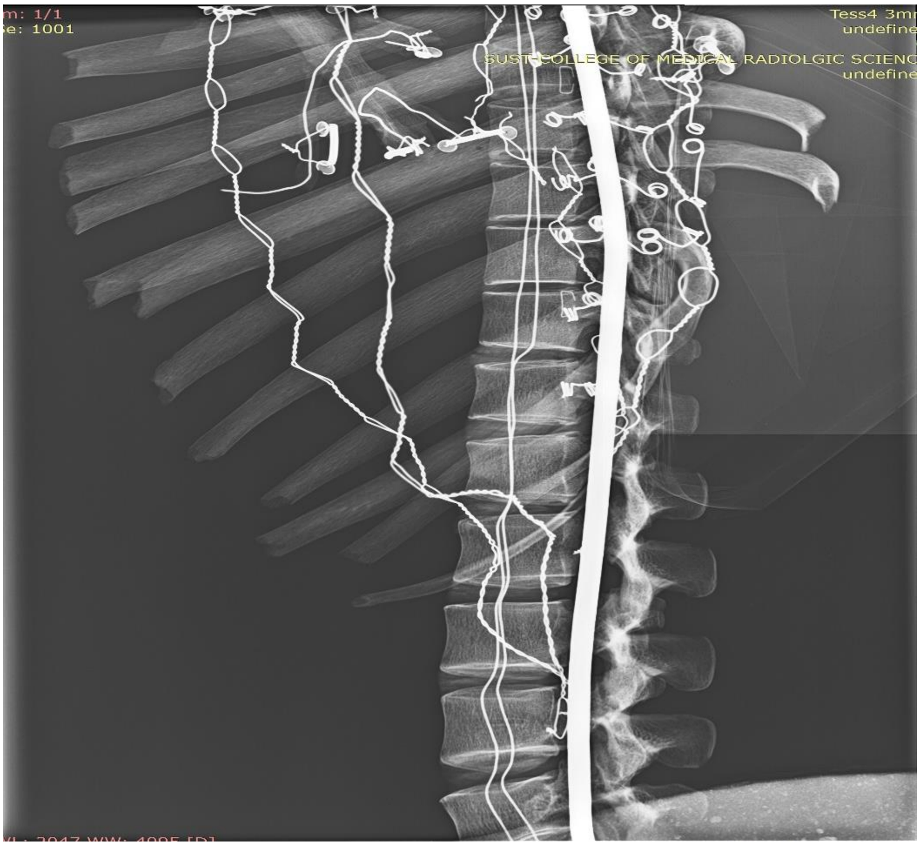

Image 5.3for thoracic lumber region with 3mm thickness of AL filter

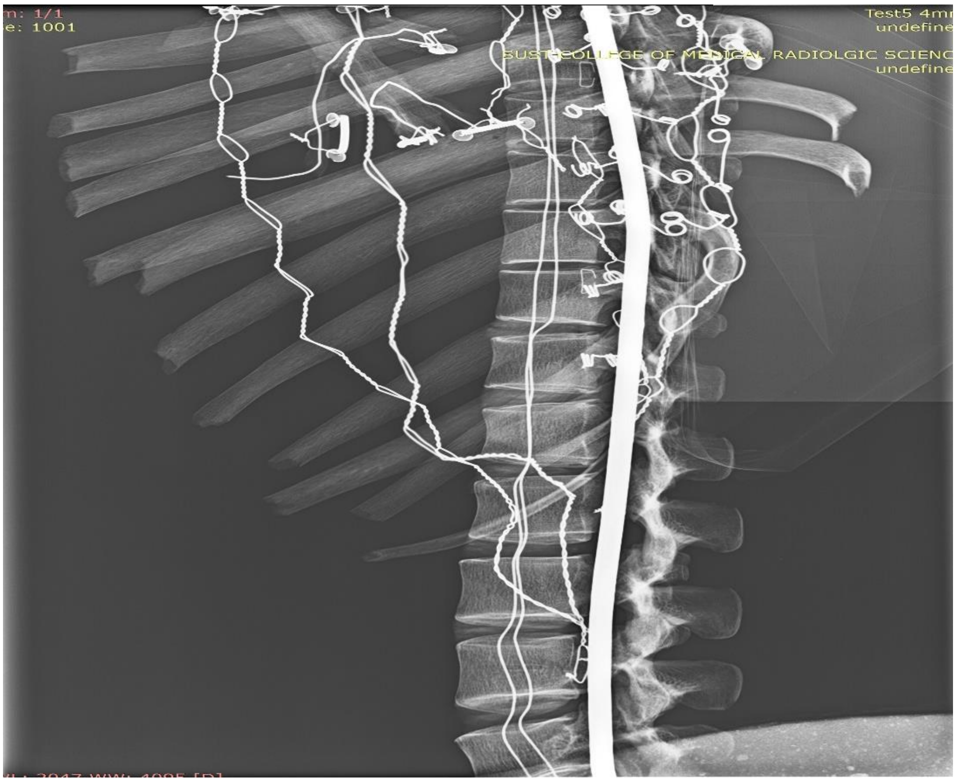

Image5.4 for thoracic lumber region with 4mm thickness of AL filter

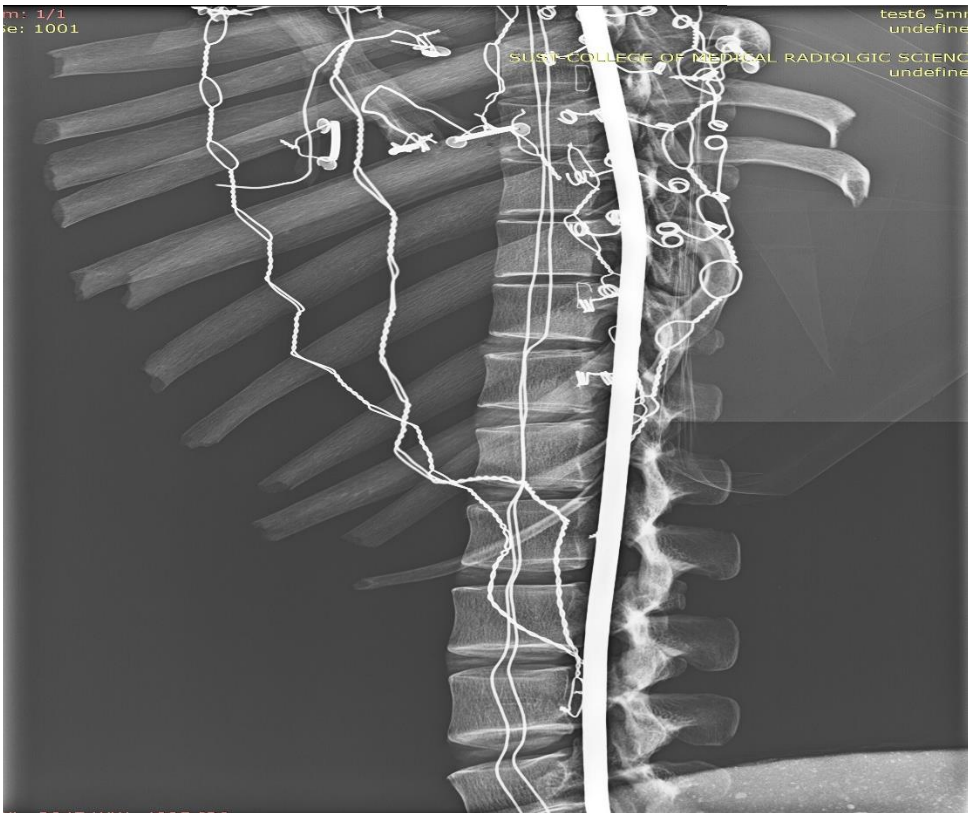

Image 5.5for thoracic lumber region with 5mm thickness of AL filter

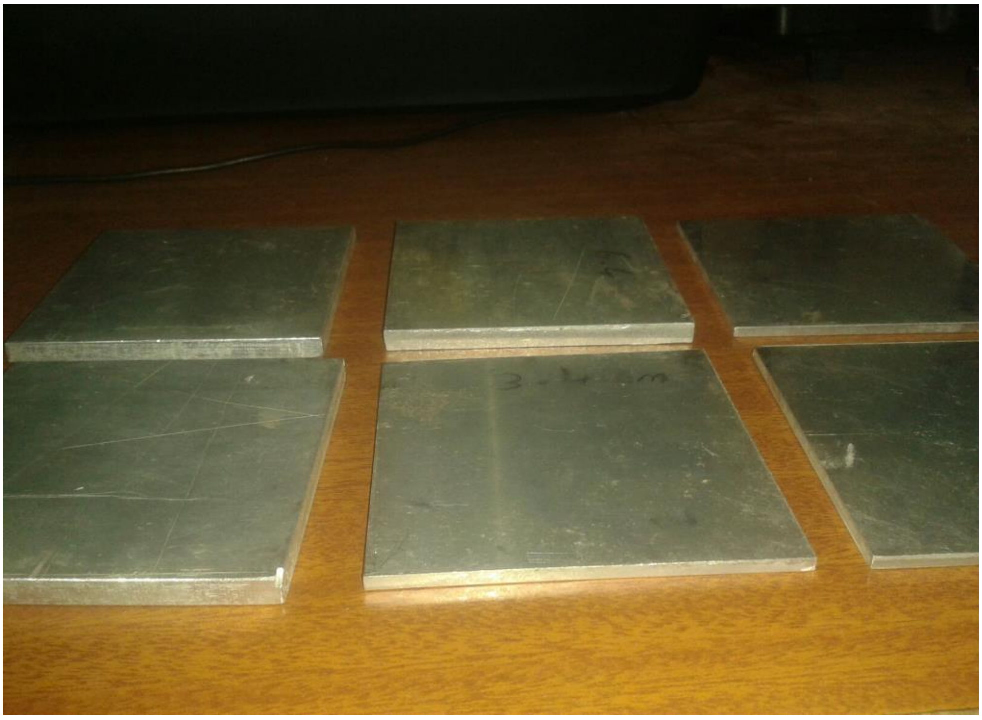

Image 5.6 different thicknesses of AL filters