

# Chapter One

## Introduction

### 1.1 Introduction:

A bone scan or bone scintigraphy is a nuclear scanning test to find certain abnormalities in bone. It is primarily used to help diagnose a number of conditions relating to bones, including: cancer of the bone or cancers that have spread (metastasized) to the bone, locating some sources of bone inflammation (e.g. bone pain such as lower back pain due to a fracture), the diagnosis of fractures that may not be visible in traditional X-ray images, and the detection of damage to bones due to certain infections and other problems. Nuclear medicine bone scans are one of a number of methods of bone imaging, all of which are used to visually detect bone abnormalities. Such imaging studies include magnetic resonance imaging (MRI), X-ray computed tomography (CT) and in the case of 'bone scans' nuclear medicine. However, a nuclear bone scan is a functional test: it measures an aspect of bone metabolism or bone remodeling, which most other imaging techniques cannot. The nuclear bone scan competes with the FDG-PET scan in seeing abnormal metabolism in bones, but it is considerably less expensive. Nuclear bone scans are not to be confused with the completely different test often termed a "bone density scan," DEXA or DXA, which is a low-exposure X-ray test measuring bone density to look for osteoporosis and other diseases where bones lose mass, without any bone-rebuilding activity. The nuclear medicine scan technique is sensitive to areas of unusual bone-rebuilding activity because the radiopharmaceutical is taken up by osteoblast cells that build bone. The technique therefore is sensitive to fractures and bone reaction to infections and bone tumors, including tumor metastases to bones,

because all these pathologies trigger osteoblast activity The bone scan is not sensitive to osteoporosis or multiple myeloma in bones; therefore, other techniques must be used to assess bone abnormalities from these diseases.

In the nuclear medicine technique, the patient is injected (usually into a vein in the arm or hand, occasionally the foot) with a small amount of radioactive material such as 740 MBq of technetium-99m-MDP and then scanned with a gamma camera, a device sensitive to the radiation emitted by the injected material. Two-dimensional projections of scintigraphy may be enough, but in order to view small lesions (less than 1cm) especially in the spine, single photon emission computed tomography (SPECT) imaging technique may be required. In the United States, most insurance companies require separate authorization for SPECT imaging. A disruption of bone turnover by a pathologic process on the order of 5 to 15% from normal can be detected by bone scintigraphy. Specificity of bone scintigraphy can be increased by performing an indium 111-labeled white blood cell test combined with a technetium-99m-MDP injection. About half of the radioactive material is localized by the bones. The more active the bone turnover, the more radioactive material will be seen. Some tumors, fractures and infections show up as areas of increased uptake. Others can cause decreased uptake of radioactive material. Not all tumors are easily seen on the bone scan. Some lesions, especially lytic (destructive) ones, require positron emission tomography (PET) for visualization. About half of the radioactive material leaves the body through the kidneys and bladder in urine. Anyone having a study should empty their bladder immediately before images are taken. In evaluating for tumors, the patient is injected with the radioisotope and returns in 2–3 hours for imaging. Image acquisition takes from 30 to 70 minutes, depending if SPECT images are required. If the physician wants to evaluate for osteomyelitis (bone infection) or fractures, then a

Three Phase/Triphasic Bone Scan is performed where 20–30 minutes of images (1st and 2nd phases) are taken during the initial injection. The patient then returns in 2–3 hours for additional images (3rd Phase). Sometimes late images are taken at 24 hours after injection. The three phase bone scan detects different types of pathology in the bone. The first phase is also known as the nuclear angiogram or the flow phase. During this phase, serial scans are taken during the first 2 to 5 seconds after injection of the Technetium-99m-MDP. This phase typically shows perfusion to a lesion. Cellulitis shows up more in phase 1 and phase 2 scan, but not in phase 3. Pathology that is more moderate to severe will show more in the first two phases. Pathology that is chronic or partially treated will be more pronounced in the third phase of a triphasic scan.<sup>[1]</sup> The second phase image, also known as the blood pool image is obtained 5 minutes after injection. This shows the relative vascularity to the area. Areas with moderate to severe inflammation have dilated capillaries, which is where the blood flow is stagnant and the radioisotope can "pool". This phase shows areas of intense or acute inflammation more definitively compared with the third phase.<sup>[1]</sup> The third phase, delayed phase, is obtained 3 hours after the injection, when the majority of the radioisotope has been metabolized. This phase best shows the amount of bone turnover associated with a lesion. A typical radiation dosage obtained during a bone scan is 6.3 mSv.

These techniques give medical images where they are analysis and enhancement by image processing (Image processing is the study of any algorithm that takes an image as input and returns an image as output) image processing give Image enhancement, noise removal, restoration, feature detection, compression and image analysis give Segmentation, image registration, matching.

## **1.2 Problem of Study:**

Medical images are often deteriorated by noise due to various of interference and other factor associated with imaging process and data acquisition system. Digital images are prone to a variety of types of noise. There are several ways that noise can be introduced into an image, depending on how the image is created. The nature of the physiological system under investigation and procedures used in imaging also diminish the contrast and the visibility of details. Sometimes information is not available a priori to identify the useful intensity bands. Radiation is a major risk in diagnostic in medical image. The problem is caused from overdose during the diagnosis due to improve the image.

## **1. 3 Objectives:**

The main objective of this study is to enhance of bone scintigraphy Image by using image processing technique

### **1.3.1. Specific Objectives:**

- To enhance of bone scintigraphy image by using MatLab global and local enhancement techniques.
- To evaluate contrast enhancement pattern in different panoramic images such as grey color.
- To evaluate the usage of new nonlinear approach for contrast enhancement of soft tissues in panoramic images.
- To highlight the importance of image processing in improvement of image quality in nuclear medicine.

## **1.4 Overview of the study**

This thesis is concerned with the Application of Preprocessing Correction of Nuclear Medicine Images using Local Adaptive Filters. This study falls into five chapters, Chapter one, which is an introduction, It presents the statement of the study problems, objectives of the study, chapter two, contains the background material for the thesis. Specifically it discusses the using of image processing in bone scan and the different methods use to enhance of bone scan image and how can do these techniques, Chapter three describes the materials and a method used to enhancement of bone scintigraphy and explains in details the methods used, Chapter fours deals with results and Chapter five discussions, conclusions, recommendations and references, Appendix.

## **Chapter Two**

### **Literature Review**

## 2.1 Nuclear Medicine Image Acquisition:

Nuclear Medicine is the section of science that utilizes the properties of radiopharmaceuticals in order to derive clinical information of the human physiology and biochemistry. The radiopharmaceutical follows its physiological pathway and it is concentrated on specific organs and tissues for short periods of time. Then, the patient is positioned under nuclear medicine equipment which can detect the radiation emitted by the human body resulting in images of the bio-distribution of the radiopharmaceutical. In Nuclear Medicine, there are two main methods of patient imaging, the imaging with Planar Imaging, Dynamic Imaging or SPECT and the PET. During the last decade, hybrid systems have been developed integrating the CT technique with either SPECT or PET resulting in SPECT/CT and PET/CT respectively. This chapter will concentrate on the implementation of MatLab code in gamma camera planar imaging, SPECT and SPECT/CT methods [1,7]. The gamma camera is composed of a collimator, scintillator crystal usually made of NaI (or CsI), the photomultiplier tubes, the electronic circuits and a computer equipped with the suitable software to depict the nuclear medicine examinations. In planar imaging, the patient, having been delivered with the suitable radiopharmaceutical, is sited under the gamma camera head. The gamma camera head remains stable at a fixed position over the patient for a certain period of time, acquiring counts (disintegrations). These will constitute the radiopharmaceutical distribution image. The counts measured in a specific planar projection originate from the whole thickness of patient [2]. In SPECT, the gamma camera head rotates around the patient remaining at well-defined angles and acquiring counts for specific periods of time per angle. What makes SPECT a valuable tool in nuclear medicine is the fact that information in the three dimensions of the patient can be collected in a number of slices with a finite

known volume(in voxels).Thus SPECT technique is used to display the radiopharmaceutical distribution in a single slice removing the contribution from the overlying and underlying tissues. In order to obtain the most accurate quantitative data from SPECT images, two issues that have to be resolved are the attenuation correction and the Compton scattering that the photons are undergone until reach and interact with the slice of interest tissues. As an examining organ has certain dimensions, each slice along the axis of the gamma camera has different distance from the detector. Thus, each photon experiences different attenuation. These two phenomena usually lead to distortion of the measured activity concentration (Abdallah Y, 2011), (Abdallah Y et al, 2014). The acquired data are processed in order to correct and compensate the undesired effect of these physical phenomena. The projection data of each slice constitute the sinogram. As a result, a series of sinograms is the files acquired. However, this kind of files needs reconstruction in order to get an image with diagnostic value. The most known reconstruction methods are the Filtered Back-Projection (FBP) and the Iterative methods. Advanced techniques of image processing and analysis find wide spread use in medicine. In medical applications, image data are used to gather details regarding the process of patient imaging whether it is a disease process or a physiological process. Information provided by medical images has become a vital part of today's patient care. The images generated in medical applications are complex and vary notably from application to application. Physiological properties of the structures-organs. In order to have high quality medical images for reliable diagnosis, the processing of image is necessary. The scope of image processing and analysis applied to medical applications is to improve the quality of the acquired image and extract quantitative information from medical image data in an efficient

and accurate way. MatLab (Matrix Laboratory) is a high performance interactive software package for scientific and engineering computation developed by Math Works. MatLab allows matrix computation, implementation of algorithms, simulation, plotting of functions and data, signal and image processing by the Image Processing Toolbox. It enables quantitative analysis and visualization of nuclear medical images of several modalities, such as Single Photon Emission Computed Tomography (SPECT), Positron Emission Tomography (PET) or a hybrid system (SPECT/CT) where a Computed Tomography system (CT) is incorporated to the SPECT system (Abdallah Y et al, 2014), (Abdallah Y, 2011).

The range of nuclear medicine examinations is fairly wide. It includes, among others, patients' studies, as myocardium perfusion by  $^{99m}\text{Tc}$ -Tetrofosmin or  $^{99m}\text{Tc}$ -Sestamibi, striatum imaging in brain by  $^{123}\text{I}$ -Ioflupane (DaTSCAN), renal parenchyma imaging by  $^{99m}\text{Tc}$ -De-Methylo-Sulfo-Acid (DMSA) and  $^{99m}\text{Tc}$ -Methylo-Di-Phosphonate (MDP) for bone scintigraphy. Fundamental image analysis methods of myocardium, brain, kidneys, thyroid, lungs and oncological (e.g. neuroblastoma) nuclear medicine studies include regions' properties, boundary analysis, curvature analysis or line and circle detection. Image processing serves in reconstruction of images acquired using SPECT techniques, in improvement of the quality of images for viewing and in preparation of images for quantitative results (Peter F, 2009).

### **2.1.1 Image types:**

The image types supported from the Image Processing Toolbox are listed below:

- **Binary Images:** In these, pixels can only take 0 or 1 value, black or white.



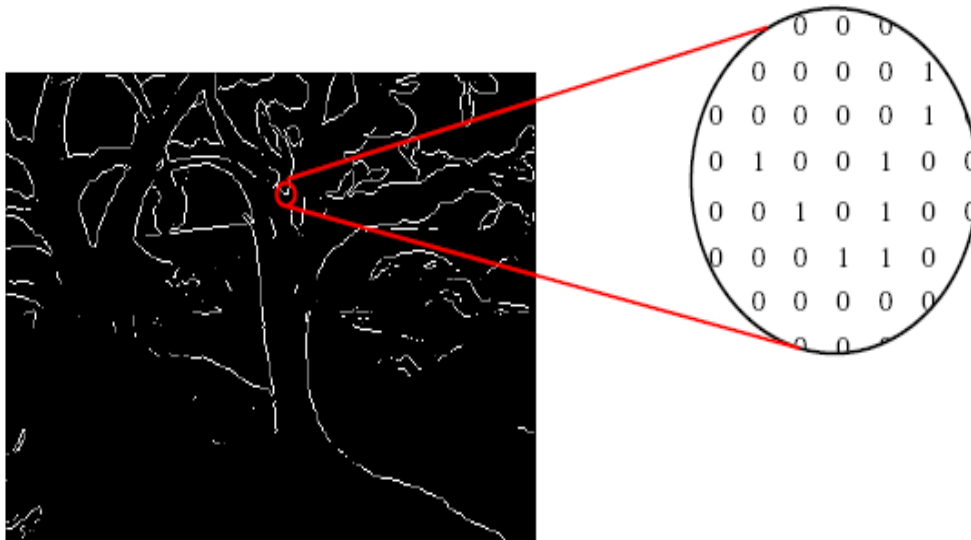


Figure 2-1 Binary image (MathWorks, 2001)

• **Grey scale or intensity images:** The image data in a grey scale image represent intensity or brightness. The integers' value is within the range of  $[0 \dots 2^k - 1]$ , where  $k$  is the bit depth of the image. For a typical grey scale image each pixel can be represented by 8 bits and intensity values are in the range of  $[0 \dots 255]$ , where 0 corresponds to black and 255 to white.

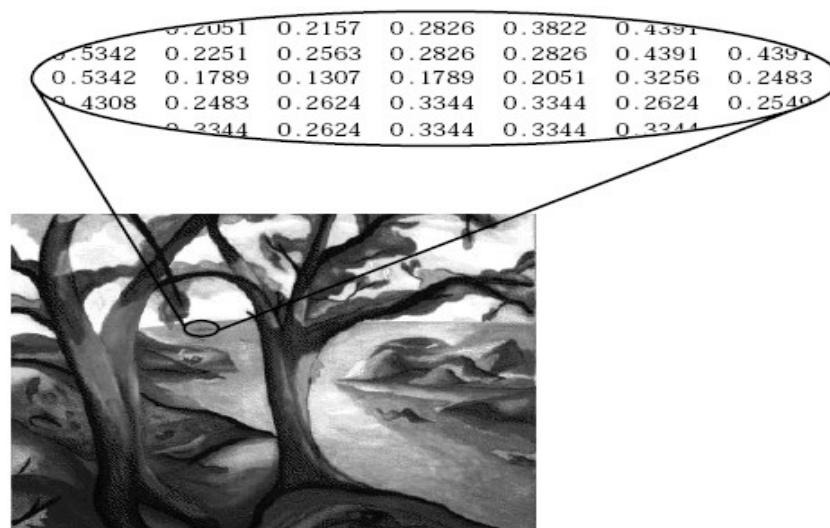


Figure 2-2. Intensity image of class double (MathWorks, 2001)

- **True color or RGB:** In these, an image can be displayed using three matrices, each one corresponding to each of red-green-blue colour. If in an RGB image each component uses 8 bits, then the total number of bits required for each pixel is  $3 \times 8 = 24$  and the range of each individual colour component is  $[0 \dots 255]$ .

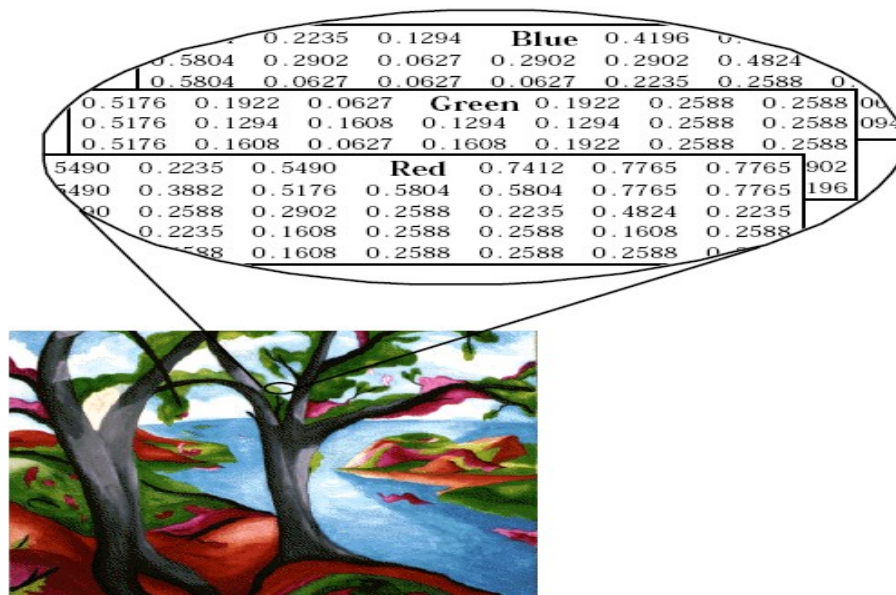


Figure 2-3. RGB image of class double (MathWorks, 2001)

- **Indexed images :** Indexed images consist of a 2D matrix together with an  $m \times 3$  colour map ( $m$ = the number of the columns in image matrix). Each row of map specifies the red, green, and blue components of a single colour. An indexed image uses direct mapping of pixel values to colour map values. The colour of each image pixel is determined by using the corresponding value of matrix as an index into map. The grey scale image is the most convenient and preferable type utilised in nuclear medicine image processing. When colouring depiction is needed, the RGB one should be used and processed. The indexed type images should be converted to any of the two other types in order to be processed. The functions used for image type conversion are: `rgb2gray`, `ind2rgb`, `ind2gray` and reversely. Any image can be also transformed to binary one using the command: `im2bw`. Moreover, in any image, the

function `impixelinfo` can be used in order to detect any pixel value. The user can move the mouse cursor inside the image and the down left corner appears the pixel identity (x, y) as well as the (RGB) values. The pixel range of the image can be displayed by the command `imshow range`.

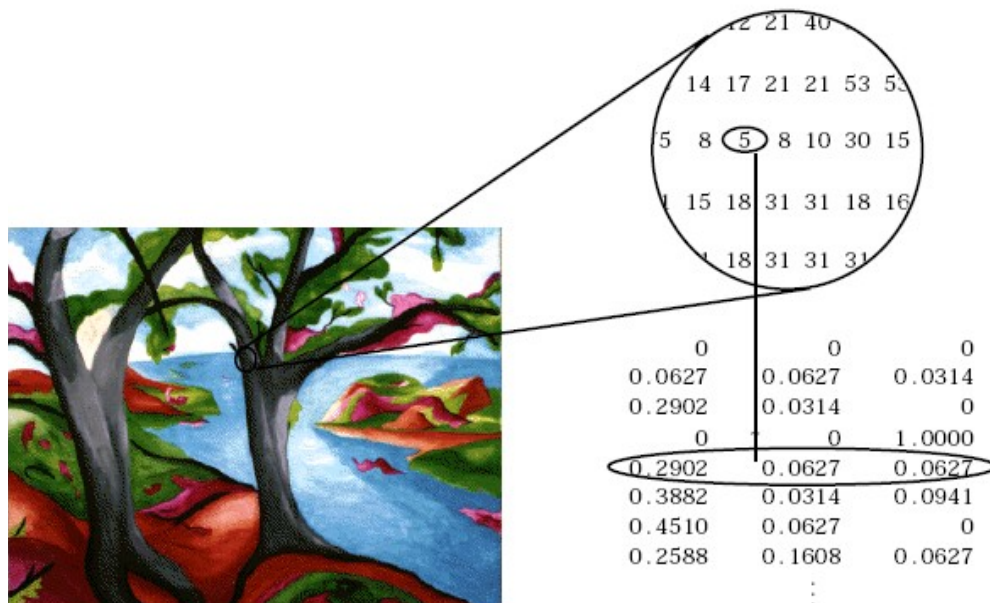


Figure 2-4. Indexed image of class double (MathWorks, 2001)

### 2.1.2 Image quality in nuclear medicine:

Image quality plays an important role in nuclear medicine imaging as the goal is a reliable image of the projected organ to be provided, for accurate diagnosis or therapy. The physical characteristics that are used to describe image quality are (1) contrast, (2) spatial resolution and (3) noise.

**Image contrast** is the difference in intensity corresponding to different concentration of activity in the patient. For high diagnostic accuracy, nuclear medicine images must be of high contrast. The image contrast is principally affected by the radiopharmaceutical that is used for imaging and the scattered radiation. In general, it is desirable to use radiopharmaceutical which has a high uptake within the target organ.

**Spatial resolution** is defined as the ability of the imaging modality to reproduce the details of a nonuniform radioactive distribution. The spatial resolution is separated into intrinsic resolution (scintillator, photomultiplier tubes and electronic circuit) and system resolution (collimator, scintillator, photomultiplier tubes and electronic circuit). The intrinsic resolution depends on the thickness of scintillation crystal while the system resolution depends mainly on the distance from the emitting source to collimator. The resolution of a gamma camera is limited by several factors. Some of these are the patient motion, the statistical fluctuation in the distribution of visible photons detected and the collimators geometry (Wernick & Aarsvold, 2004).

**Noise** refers to any unwanted information that prevents the accurate imaging of an object.

**Noise** is the major factor in the degradation of image quality. Image noise may be divided into random and structured noise. Random noise (also referred as statistical noise) is the result of statistical variations in the counts being detected. The image noise is proportional to  $N^{1/2}$  where  $N$  is the number of detected photons per pixel. Therefore, as the number of counts increases the noise level reduces. Image noise is usually analysed in terms of signal-to-noise-ratio (SNR). SNR is equal to  $N / N^{1/2}$ . If the SNR is high, the diagnostic information of an image is appreciated regardless of the noise level. Structured noise is derived from non-uniformities in the scintillation camera and overlying structures in patient body.

### **2.1.3 Acquiring data:**

The signal used in nuclear medicine is basically that of emitted ionising radiation from internal sources of radioactively labelled material. In general these will be gamma rays of energies from 30KeV up to 500KeV where the radioactive material has been administered by injection or orally. The original classical example was the

oral absorption of I131 which then goes to the thyroid, decays to produce 364KeV gammas, which are then detected to form an image of the thyroid. The most commonly used radioisotope is that of Tc-99m, which is used to label a wide variety of substances, and which decays to produce a 140KeV gamma. Other interesting isotopes are the positron emitters C-11, N-13 O-15, F-18, all of which generate coincident pairs of 511KeV gammas and are used for PET. The most common device used for imaging in nuclear medicine is the gamma camera. Here, a collimator being a sheet of lead with holes, is used to select photons arriving from a particular (set of) directions, and then the gamma events are detected from a scintillation occurring most commonly in a sheet of a material such as NaI(Tl) producing light which is then converted to electrical signals by an array of photomultipliers. Data are acquired on an event by event basis. The position (and energy) of each individual detected gamma event is recorded or used to create images. Thus two basic detection modes are used, frame mode and list mode. Additional information such as from the ECG may be employed as in so called gated studies to create time-lapse averaged images. Quite significant pre-processing is included as part of the acquisition system to correct for errors notably in energy, uniformity and of spatial distortion.

Frame mode acquisition is the most commonly used. The form of the data coming from the detector is as a sequence of x,y coordinates. At some point we must form images. If we form images directly as part of the acquisition, this is called frame mode, whereas if we store the list of x,y coordinates, this is called list mode. In frame mode, a buffer exists, which could be a slab of memory in the acquisition interface, or could be part of the computer memory, which has been allocated to an image (a frame). As each event arrives, after being digitized and corrected, the x,y coordinate is used to form an address, and the corresponding location in the frame incremented by

one. The frame therefore acts as a counter, and (provided that it was initialised to zero) after some interval of time, corresponds to the (digital) image. The parameters of the frame size which can vary are the frame size, and the bit depth.

Frames are usually in powers of two for nuclear medicine applications, going from as little as 32x32 up to 512x512. Sampling theory states that we should sample at twice the frequency of the maximum frequency in the signal, depending on depth etc, almost certainly not less than 6mm. Thus we should sample at 3mm intervals, that is, for a 40cm field of view, the matrix size should be about 133x133. Thus 128x128 is a reasonable matrix size, and there is little point in theory to going to finer matrices than this. The only justification might be for display or for correction of errors such as spatial distortion.

The frame time is the length of time for which data is collected. The maximum count which might occur at a pixel is dependent on the total number of counts collected (which depends on count rate and frame time), and the activity distribution. If the activity is distributed over a small area, the maximum count rate will be much higher for example as might occur with a point source. The number of bits available should be enough to accommodate the maximum number of counts. The normal choices are between byte mode where 8 bits per pixel are allowed, word mode where typically 16 bits per pixel are allowed, and certain system which use intermediate values such as 12 bits per pixel. The data is normally considered as being an unsigned integer and thus the maximum number of counts in a byte mode acquisition is 256, and 65536 for a word mode acquisition. Action must be taken on over flow, that is when the maximum value is exceeded. One possibility is to start a new frame, but that would make the frame timing irregular. The interframe time is the time gap between two frames, and is normally very short, when double buffering is used. The term Gated

study is used to indicate a study where a set of images are acquired for a whole range of time delays, for example, throughout the cardiac cycle. Gating is equivalent to time lapse averaging that is, data is averaged together for some time delay after some signal. The signal used here, in cardiac studies, is the R wave of the ECG. Thus essentially, a gated image is formed by acquiring data in frame mode, but only accepting events which occurred during some time interval after the R wave of each cardiac cycle. Thus we could gate an image coincident with the R wave to give an image of the heart at end-diastole, or use an appropriate delay to obtain an image at endsystole. There are many details of importance, primarily resulting from the fact that the cardiac rhythm is unlikely to be completely regular. The first choice for a Gated acquisition is to acquire a set of frames at fixed time intervals after the R wave, for example every 10 milliseconds. When this is chosen, when a shorted beat occurs, fewer counts will be added into the last frames, and the count density of these frame will be less. When a time activity curve is drawn, it will tend to droop at the end. An alternative is phase gating where the cardiac cycle is divided into a fixed number of frames, for example 32, and the time per frame is chosen from the R-R interval of the current beat. However, the R-R interval cannot be known in advance, and this implies either buffering the data on input, and storing at least one beats worth of information, or using list mode.

In addition, the components of the cardiac cycle, for example time between e-d and e-s may not change proportionally to the change in R-R interval. If the data is buffered, it is also possible to perform backwards gating, that is forming images with respect to fixed time intervals before the next R wave. Indeed it is possible to perform hybrid gating, forward gating the start of the cardiac cycle, and backwards gating the end (and hoping it matches in the middle. Bad beat exclusion also needs to be performed,

that is, rejecting data coming from an R-R interval which is either too long or too short, or following a bad beat, when the heart may not have filled adequately. There is some clinical disagreement on what is the best technique to use. List mode acquisition is a very important extra to have in addition to frame mode. The data is directly stored to disk and then replayed to create images and enables 'impossible' acquisitions to be created by choosing frame times after the acquisition. Although frame mode data collection is used for the bulk of acquisitions in nuclear medicine, there are occasions when list mode is essential. List mode acquisition is that mode where the individual x and y coordinates of each pulse recorded is sent directly to some backing store (such as a hard disk) without being turned into image frames. It therefore can occupy quite a lot of space on disk, approximately 2bytes per event recorded. Usually the acquisition system is double buffered. Having stored the data, one usually wants at some time, to do something with it. This involves replay of the data, that is, recalling it so as to be able to form image frames. The way in which image frames are formed is just like the way in which they were created in frame mode acquisition, except that the source of the data is the backing store, rather than the camera. The reason for doing this is that the parameters of the image frames created can then be selected AFTER the acquisition. Thus we do not need to choose in advance the frame times, this could be chosen a posteriori, when we know what frame times are appropriate. However, in order to be able to reframe at all, we must store timing information with the raw data. This is usually performed by inserting codes into the list mode data, for example time ticks and physiological markers when appropriate. There are several other reasons why list mode data could be useful. It is possible using list mode to create image frames which could not have been created directly. For example, given the available amount of memory available for acquisition, there is a maximum number of frames



that could be created for a gated study. Using list mode data, and replaying the data, possibly several times, this limit can be exceeded. Similarly, backwards gating, that is forming images at fixed time with respect to some event in the future, can be easily handled by list mode acquisition. Handling bad beat rejection can be much more sophisticated in list mode. The drawbacks of list mode acquisition is that the maximum count rate that can be handled is low, limited by the speed of writing onto disk, the space required is usually rather large, typical several megabytes, and finally the replay process is time consuming. List mode data seems to be of particular interest in PET imaging.

## **2.2. Nuclear Medicine Image Enhancement Technique:**

Image Enhancement Technique are used to refine given image, so that desired image features become easier to perceive for the human visual system or more likely to be detected by automated image analysis (Graaf,C, 1988). Image Enhancement allows the observable to see details in image that may not be immediately observable in the original image . This may be the case , for example , when the dynamic range of the data and that of the display are not commensurate , when the image has high level of noise or when contrast is insufficient (Graaf,C, 1988). Fundamentally, Image enhancement is the transformation or mapping of one image to another (Niblack, 1986). This transformation is not necessarily one-to-one, so that two different input image may transform into the same or similar output image after enhancement. More commonly, one may want to generate multiple enhanced versions of given image. This aspect also means that enhancement techniques may be irreversible. Often the enhancement of certain features in image is accompanied by undesirable effects. Valuable image information may be lost or the enhanced image may be poor representation of the original. Furthermore, enhancement algorithms cannot be

expected to provide information that is not present in the original image. If the image does not contain the feature to be enhanced, noise or other unwanted image components may be inadvertently enhanced without any benefit to the user.

Image Enhancement Involves taking an image and improving it visually, typically by taking advantages of human Visual Systems responses. One of the simplest enhancement techniques is to simply stretch the contrast of an image. Enhancement methods tend to be problem specific. For example, a method is used to enhance satellite images may not be suitable for enhancing medical images (Niblack, 1986). Although enhancement and restoration are similar in aim, to make an image look better. They differ in how they approach the problem.

### **Why perform enhancement?**

The basic goal of image enhancement is to process the image so that we can view and assess the visual information it contains with greater clarity. Image enhancement, therefore, is rather subjective because it depends strongly on the specific information the user is hoping to extract from the image.

The primary condition for image enhancement is that the information that you want to extract, emphasize or restore must exist in the image. Fundamentally, 'you cannot make something out of nothing' and the desired information must not be totally swamped by noise within the image. Perhaps the most accurate and general statement we can make about the goal of image enhancement is simply that the processed image should be more suitable than the original one for the required task or purpose. This makes the evaluation of image enhancement, by its nature, rather subjective and, hence, it is difficult to quantify its performance apart from its specific domain of application.

### **2.2.1 Enhancement via image filtering:**

The main goal of image enhancement is to process an image in some way so as to render it more visually acceptable or pleasing. The removal of noise, the sharpening of image edges and the ‘soft focus’ (blurring) effect so often favoured in romantic photographs are all examples of popular enhancement techniques. These and other enhancement operation scan be achieved through the process of spatial domain filtering. The term spatial domain is arguably somewhat spurious, but is used to distinguish this procedure from frequency domain procedures. Thus, spatial domain filtering simply indicates that the filtering process takes place directly on the actual pixels of the image itself (John Wiley et al, 2011). Therefore, we shall refer simply to filtering in this chapter without danger of confusion. Filters act on an image to change the values of the pixels in some specified way and are generally classified into two types: linear and nonlinear. Linear filters are more common, but we will discuss and give examples of both kinds. Irrespective of the particular filter that is used, all approaches to spatial domain filtering operate in the same simple way. Each of the pixels in an image – the pixel under consideration at a given moment is termed the target pixel – is successively addressed. The value of the target pixel is then replaced by a new value which depends only on the value of the pixels in a specified neighborhood around the target pixel.

## **2-3 Application of Image processing in Nuclear Medicine:**

Image processing is a set of techniques in which the data from an image are analyzed and processed using algorithms and tools to enhance certain image information that is more useful to human interpretation (Nailon, 2010). The processing of an image permits the extraction of useful parameters and increases the possibility of detection of small lesions more accurately. Image processing in nuclear medicine serves three

major purposes: a) the reconstruction of the images acquired with tomographic (SPECT) techniques, b) the quality improvement of the image for viewing in terms of contrast, uniformity and spatial resolution and, c) the preparation of the image in order to extract useful diagnostic qualitative and quantitative information. Advanced techniques of image processing and analysis find widespread use in medicine. In medical applications, image data are used to gather details regarding the process of patient imaging whether it is a disease process or a physiological process. Information provided by medical images has become a vital part of today's patient care. The images generated in medical applications are complex and vary notably from application to application. Nuclear medicine images show characteristic information about the physiological properties of the structures-organs. In order to have high quality medical images for reliable diagnosis, the processing of image is necessary. The scope of image processing and analysis applied to medical applications is to improve the quality of the acquired image and extract quantitative information from medical image data in an efficient and accurate way.

### **2-3-1 Image processing techniques – MatLab:**

Image processing techniques include all the possible tools used to change or analyze an image according to individuals' needs. This subchapter presents the most widely performed image processing techniques that are applicable to nuclear medicine images. MatLab (Matrix Laboratory) is a high performance interactive software package for scientific and engineering computation developed by Math Works (Mathworks Inc., 2009). MatLab allows matrix computation, implementation of algorithms, simulation, plotting of functions and data, signal and image processing by the Image Processing Toolbox. It enables quantitative analysis and visualization of

nuclear medical images of several modalities, such as Single Photon Emission Computed Tomography (SPECT), Positron Emission Tomography (PET) or a hybrid system (SPECT/CT) where a Computed Tomography system (CT) is incorporated to the SPECT system. The Image Processing Toolbox (Mathworks Inc., 2009). is a comprehensive set of reference-standard algorithms and graphical tools for image processing, analysis, visualisation and algorithm development. It offers the possibility to restore noisy or degraded images, enhance images for improved intelligibility, extract features, analyse shapes and textures, and register two images. Thus, it includes all the functions that MatLab utilizes in order to perform any sophisticated analysis needed after the acquisition of an image. Most toolbox functions are written in open MatLab language offering the opportunity to the user to inspect the algorithms, to modify the source code and create custom functions (Wilson et al., 2003, Perutka, 2010).

### **2.3.2 Contrast enhancement:**

One of the very first image processing issues is the contrast enhancement. The acquired image does not usually present the desired object contrast. The improvement of contrast is absolutely needed as the organ shape, boundaries and internal functionality can be better depicted. In addition, organ delineation can be achieved in many cases without removing the background activity. The command that implements contrast processing is the `imadjust`. Using this, the contrast in an image can be enhanced or degraded if needed. Moreover, a very useful result can be the inversion of colours, especially in grey scale images, where an object of interest can be efficiently outlined. The general function that implements contrast enhancement is the following:

```
J = imadjust(I,[low_in high_in],[low_out high_out],gamma);
```

while the function for colour inversion is the following:

```
J = imadjust(I,[0 1],[1 0],gamma); or J = imcomplement(I);
```

suppose that J, is the new image, I, is the initial image and gamma factor depicts the shape of the curve that describes the relationship between the values of I and J. If the gamma factor is omitted, it is considered to be 1.

### 2.3.3 Organ contour:

In many nuclear medicine images, the organs' boundaries are presented unclear due to low resolution or presence of high percentage of noise. In order to draw the contour of an organ in a nuclear medicine image, the command `imcontour` is used. In addition, a variable `n` defines the number of equally spaced contours required. This variable is strongly related with the intensity of counts. For higher `n` values, the lines are drawn with smaller spaces in between and depict different streaks of intensity. The type of line contouring can be specified as well. For example, when a contour of 5 level contours, drawn with solid line, is the desirable outcome, the whole function is:

```
Example 1 I = imread('kindeys.jpg');
```

```
figure, imshow(I)
```

```
J = imcontour(I,5,'-');
```

```
Figure, imshow(J)
```

where J and I stands for the final and the initial image respectively and the symbol ('-') stands for the solid line drawing. An example of the initial image, the contour with  $n=15$  and  $n=5$  respectively, follows. (Peter F, 2009).

### **2.3.4 Image interpolation:**

Interpolation is a topic that has been widely used in image processing. It constitutes of the most common procedure in order to resample an image, to generate a new image based on the pattern of an existing one. Moreover, re-sampling is usually required in medical image processing in order to enhance the image quality or to retrieve lost information after compression of an image (Lehmann et al., 1999). Interpreting the interpolation process, the user is provided with several options. These options include the resizing of an image according to a defined scaling factor, the choice of the interpolation type and the choice of low-pass filter. The general command that performs image resizing is `imresize`. However, the way that the whole function has to be written depends heavily on the characteristics of the new image. The size of the image can be defined as a scaling factor of the existing image or by exact number of pixels in rows and columns. Concerning the interpolation types usually used in nuclear medicine, these are the following: a) nearest-neighbour interpolation ('nearest'), where the output pixel obtains the value of the pixel that the point falls within, without considering other pixels, b) bilinear interpolation ('bilinear'), where the output pixel obtains a weighted average value of the nearest 2x2 pixels, c) cubic interpolation ('bicubic'), where the output pixel obtains a weighted average value of the nearest 4x4 pixels (Lehmann et al., 1999). When an image has to resize in a new one, with specified scaling factor and method, then the function Implementing that, is the following:

```
New Image = imresize(Image, scale, method);
```

For example, for a given image I, the new image J shrunk twice of the initial one, using the bilinear interpolation method, the function will be:

```
J = imresize(I, 0.5, 'bilinear');
```

This way of image resizing contributes to the conversion of image information during any such process, a fact that is valuable in the precision of a measurement. Bilinear interpolation is often used to zoom into a 2D image or for rendering, for display purposes. Apart from the previous methods, the cubic convolution method can be applied to 3D images.

### **2.3.5 Image filtering:**

The factors that degrade the quality of nuclear medicine images result in blurred and noisy images with poor resolution. One of the most important factors that greatly affect the quality of clinical nuclear medicine images is image filtering. Image filtering is a mathematical processing for noise removal and resolution recovery. The goal of the filtering is to compensate for loss of detail in an image while reducing noise. Filters suppressed noise as well as deblurred and sharpened the image. In this way, filters can greatly improve the image resolution and limit the degradation of the image. An image can be filtered either in the frequency or in the spatial domain. In the first case the initial data is Fourier transformed, multiplied with the appropriate filter and then taking the inverse Fourier transform, re-transformed into the spatial domain. The basics steps of filtering in the frequency domain are illustrated in Fig. 3.

Fig. 3. Basics steps of frequency domain filtering. The filtering in the spatial domain demands a filter mask (it is also referred as kernel or convolution filter). The filter



mask is a matrix of odd usually size which is applied directly on the original data of the image. The mask is centred on each pixel of the initial image.

For each position of the mask the pixel values of the image is multiplied by the corresponding values of the mask. The products of these multiplications are then added and the value of the central pixel of the original image is replaced by the sum. This must be repeated for every pixel in the image. If the filter, by which the new pixel value was calculated, is a linear function of the entire pixel values in the filter mask (e.g. the sum of products), then the filter is called linear. If the output pixel is not a linear weighted combination of the input pixel of the image then the filtered is called non-linear. According to the range of frequencies they allow to pass through filters can be classified as low pass or high pass. Low pass filters allow the low frequencies to be retained unaltered and block the high frequencies. Low pass filtering removes noise and smooth the image but at the same time blur the image as it does not preserve the edges. High pass filters sharpness the edges of the image (areas in an image where the signal changes rapidly) and enhance object edge information. A severe disadvantage of high pass filtering is the amplification of statistical noise present in the measured counts. The next section is referred to three of the most common filters used by MatLab: the mean, median and Gaussian filter.

### **2.3.6 Preprocessing:**

The preprocessing algorithm, techniques and operators are use to perform initial processing that makes the primary data reduction and analysis task easier. They include operations related to:

- Extracting regions of interest.
- Performing basic algebraic operation on image.

- Enhancing specific image features.
- Reducing data in resolution and brightness.

Preprocessing is a stage where the requirements are typically obvious and simple, such as removal of artifacts from images or eliminating of image information that is not required for the application. For example, in one application we needed to borders from the images that have been digitized from film. Another example of preprocessing step involves a robotics gripper that needs to pick and place an object; for this we reduce a gray-level image to binary (two-valued) image that contains all the information necessary to discern the object 's outlines .

## **Chapter Three**

### **Materials and Methods**

#### **3.1. Materials:**

##### **3.1.1. Equipments:**

- Personal Computer (PC)
- MatLab program version R2009a (9.0.2.1.0)
- Gamma camera

#### **3.2. Study Duration:**

This study proposed to be carried between November2014 to March 2015.

##### **3.2.1.Study Place:**

The proposed study was conducted in Elnileen Center for Nuclear Medicine, College of Medical Radiological Science, Sudan University of Science and Technology

##### **3.2.2. Study Sample :**

The totals of number of patients in this study were 5 patients.

#### **3.3. Methods of data collection:**

## Image enhancement technique:

### 1. Contrast-limited adaptive histogram equalization (CLAHE):

This technique used to enhance the contrast of the grayscale image (bone scan) by transforming the values using contrast-limited adaptive histogram equalization (*CLAHE*). *CLAHE* operated on small regions in the image, called *tiles*, rather than the entire image, each tile's contrast was enhanced, so that the histogram of the output region approximately matches the histogram specified by the ['Distribution'](#) parameter. The neighboring tiles were then combined using bilinear interpolation to eliminate artificially induced boundaries. The contrast, especially in homogeneous areas, could be limited to avoid amplifying any noise that might be present in the image.

$J = \text{adapthisteq}(I)$  enhanced the contrast of the grayscale image  $I$  by transforming the values using contrast-limited adaptive histogram equalization (*CLAHE*).

### 2. Enhance contrast using histogram equalization:

This programming code (*histeq*) enhanced the contrast of images (bone scan) by transforming the values in an intensity image, or the values in the *colormap* of an indexed image, so that the histogram of the output image approximately matched a specified histogram.  $J = \text{histeq}(I, hgram)$  transformed the intensity image  $I$  so that the histogram of the output intensity image  $J$  with  $\text{length}(hgram)$  bins approximately matches *hgram*. The vector *hgram* should contain integer counts for equally spaced bins with intensity values in the appropriate range:  $[0, 1]$  for images of class *double*,  $[0, 255]$  for images of class *uint8*, and  $[0, 65535]$  for images of class *uint16*. *histeq* automatically scales *hgram* so that  $\text{sum}(hgram) = \text{prod}(\text{size}(I))$ . The histogram of  $J$

was better match *hgram* when *length(hgram)* is much smaller than the number of discrete levels in *I*.

$J = \text{histeq}(I, n)$  transforms the intensity image *I*, returning in *J* an intensity image with *n* discrete gray levels. A roughly equal number of pixels were mapped to each of the *n* levels in *J*, so that the histogram of *J* is approximately flat. (The histogram of *J* was flatter when *n* was much smaller than the number of discrete levels in *I*) The default value for *n* is 64.  $[J, T] = \text{histeq}(I, \dots)$  returns the grayscale transformation that maps gray levels in the image *I* to gray levels in *J*.

$\text{newmap} = \text{histeq}(X, \text{map}, \text{hgram})$  transforms the *colormap* associated with the indexed image *X* so that the histogram of the gray component of the indexed image (*X*,*newmap*) approximately matches *hgram*. The *histeq* function returns the transformed *colormap* in *newmap*. *length(hgram)* must be the same as *size(map,1)*.  $\text{newmap} = \text{histeq}(X, \text{map})$  transforms the values in the *colormap* so that the histogram of the gray component of the indexed image *X* is approximately flat. It returns the transformed *colormap* in *newmap*.  $[\text{newmap}, T] = \text{histeq}(X, \dots)$  returns the grayscale transformation *T* that maps the gray component of *map* to the gray component of *newmap*.

### 3.Adjust image intensity values or colormap:

This programming code (*imadjust*) used to enhance the images of bone scan by increased of contract of the image ,  $J = \text{imadjust}(I)$  mapped the intensity values in grayscale image *I* to new values in *J* such that 1% of data is saturated at low and high intensities of *I*. This increased the contrast of the output image *J*. This syntax was equivalent to  $\text{imadjust}(I, \text{stretchlim}(I))$ .  $J = \text{imadjust}(I, [\text{low\_in}; \text{high\_in}], [\text{low\_out};$

*high\_out*]) maps the values in *I* to new values in *J* such that values between *low\_in* and *high\_in* map to values between *low\_out* and *high\_out*. Values below *low\_in* and above *high\_in* were clipped; that is, values below *low\_in* map to *low\_out*, and those above *high\_in* map to *high\_out*. Could use an empty matrix ([]) for [*low\_in high\_in*] or for [*low\_out high\_out*] to specify the default of [0 1].

$J = \text{imadjust}(I, [\text{low\_in}; \text{high\_in}], [\text{low\_out}; \text{high\_out}], \text{gamma})$  maps the values in *I* to new values in *J*, where *gamma* specifies the shape of the curve describing the relationship between the values in *I* and *J*. If *gamma* was less than 1, the mapping was weighted toward higher (brighter) output values. If *gamma* was greater than 1, the mapping was weighted toward lower (darker) output values. If omitted the argument, *gamma* defaults to 1 (linear mapping).  $\text{newmap} = \text{imadjust}(\text{map}, [\text{low\_in}; \text{high\_in}], [\text{low\_out}; \text{high\_out}], \text{gamma})$  transforms the *colormap* associated with an indexed image. If *low\_in*, *high\_in*, *low\_out*, *high\_out*, and *gamma* were scalars, then the same mapping applies to red, green, and blue components. Unique mappings for each color component are possible when *low\_in* and *high\_in* are both 1-by-3 vectors. *low\_out* and *high\_out* are both 1-by-3 vectors, or *gamma* was a 1-by-3 vector. The rescaled *colormap newmap* was the same size as *map*.

$\text{RGB2} = \text{imadjust}(\text{RGB1}, \dots)$  performs the adjustment on each image plane (red, green, and blue) of the *RGB* image *RGB1*. As with the *colormap adjustment*, could apply unique mappings to each plane.

#### **4.2-D median filtering:**

Median filtering was a nonlinear operation often used in image processing to reduce "*salt and pepper*" noise from the images (bone scan). A median filter was more

effective than convolution when the goal was to simultaneously reduce noise and preserve edges.

**$B = \text{medfilt2}(A, [m\ n])$**  performs median filtering of the matrix  **$A$**  in two dimensions. Each output pixel contains the median value in the  **$m$** -by- **$n$**  neighborhood around the corresponding pixel in the input image.  **$\text{medfilt2}$**  pads the image with  **$0$ s** on the edges, so the median values for the points within  **$[m\ n]/2$**  of the edges might appear distorted.  **$B = \text{medfilt2}(A)$**  performs median filtering of the matrix  **$A$**  using the default 3-by-3 neighborhood.  **$B = \text{medfilt2}(A, 'indexed', ...)$**  processes  **$A$**  as an indexed image, padding with  **$0$ s** if the class of  **$A$**  is  **$uint8$** , or  **$1$ s** if the class of  **$A$**  is  **$double$** .  **$B = \text{medfilt2}(..., \text{padopt})$**  controls how the matrix boundaries were padded.  **$\text{padopt}$**  may be  **$'zeros'$**  (the default),  **$'symmetric'$** , or  **$'indexed'$** . If  **$\text{padopt}$**  is  **$'symmetric'$** ,  **$A$**  was symmetrically extended at the boundaries. If  **$\text{padopt}$**  is  **$'indexed'$** ,  **$A$**  was padded with ones if it was  **$double$** ; otherwise it was padded with  **$zeros$** .

## 5.contrast stretch image:

This programming code used to enhance of the images (bone scan),  **$LOW\_HIGH = \text{stretchlim}(I)$**  returns  **$LOW\_HIGH$** , a two-element vector of pixel values that specify lower and upper limits that can be used for contrast stretching image  **$I$** . By default, values in  **$LOW\_HIGH$**  specify the bottom 1% and the top 1% of all pixel values. The gray values returned can be used by the  **$\text{imadjust}$**  function to increase the contrast of an image.

**$LOW\_HIGH = \text{stretchlim}(I, TOL)$**  where  **$TOL$**  is a two-element vector  **$[LOW\_FRACT\ HIGH\_FRACT]$**  that specifies the fraction of the image to saturate at low and high pixel values. If  **$TOL$**  is a scalar,  **$LOW\_FRACT = TOL$** , and  **$HIGH\_FRACT = 1 -$**

*LOW\_FRACT*, which saturates equal fractions at low and high pixel values. If you omit the argument, *TOL* defaults to  $[0.01 \ 0.99]$ , saturating 2%. If  $TOL = 0$ ,  $LOW\_HIGH = [\min(I(:)); \max(I(:))]$ .  $LOW\_HIGH = stretchlim(RGB, TOL)$  returns a 2-by-3 matrix of intensity pairs to saturate each plane of the *RGB* image. *TOL* specifies the same fractions of saturation for each plane.

## Chapter Four

### The Results

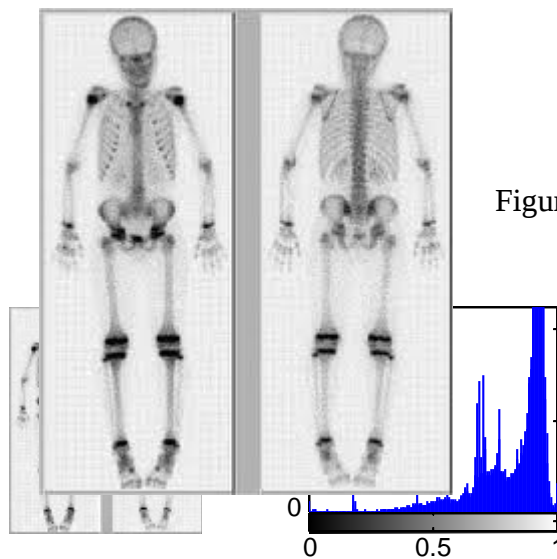


Figure 4-1. The original bone scan image test (A1)

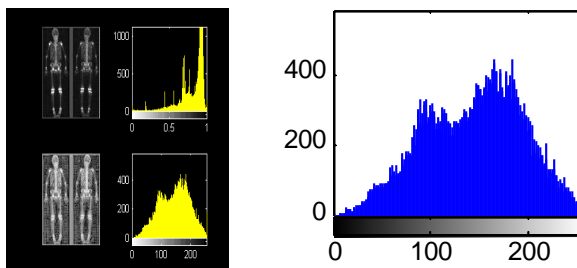


Figure 4-2. CLAHE technique; at upper side the original image with histogram and at lower side the enhanced image using CLAHE technique from test (A1)



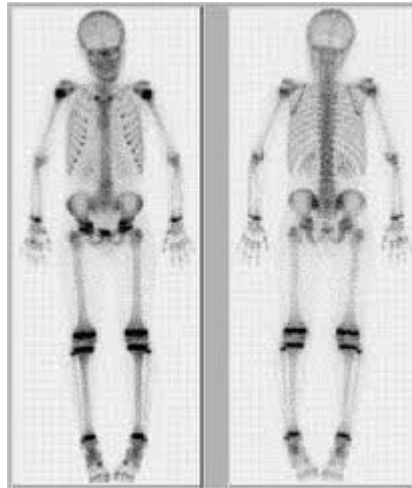


Figure 4-3. The original bone scan image test (A2)

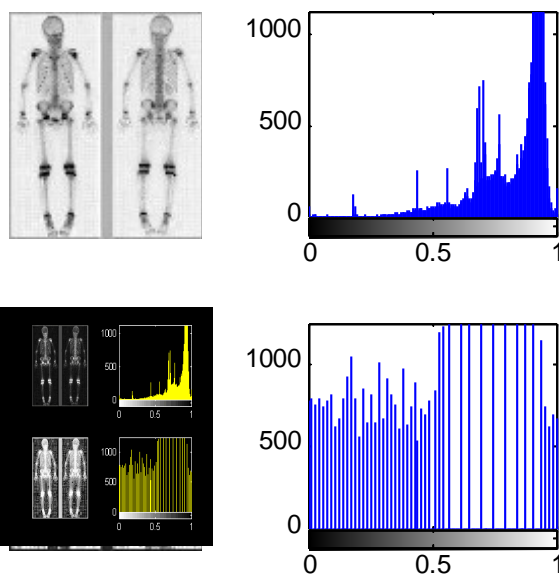


Figure 4-4. histogram equalization technique; at upper side the original image with histogram and at lower side the enhanced image using histogram equalization technique from test (A2)

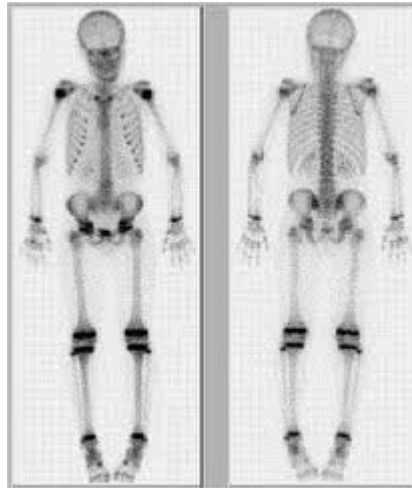


Figure 4-5. The original bone scan image test(A3)

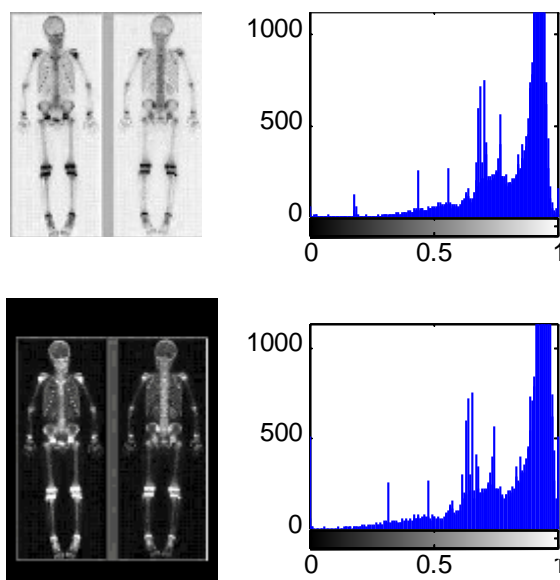


Figure 4-6. Adjust technique; at upper side the original image with histogram and at lower side the enhanced image using Adjust technique from test (A3)

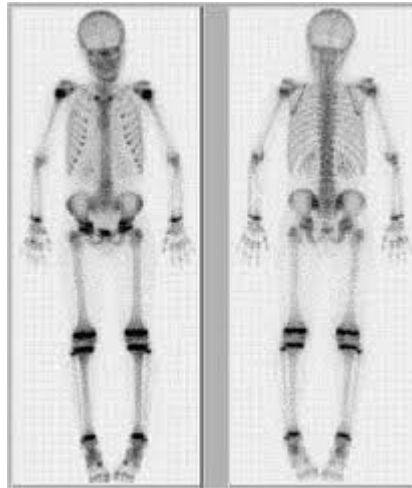


Figure 4-7. The original bone scan image test (A4)

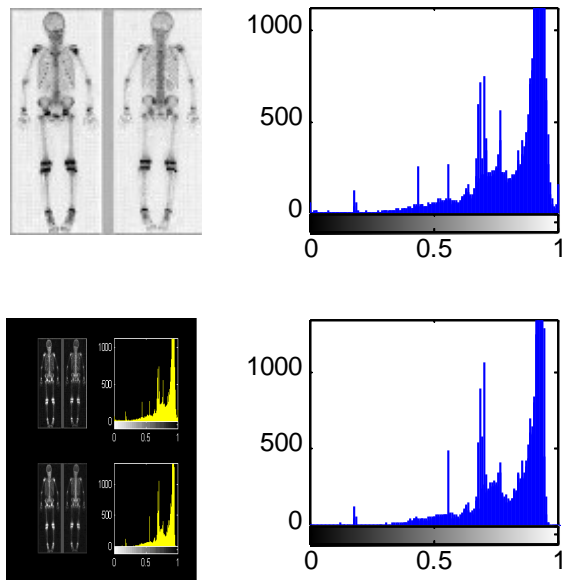


Figure 4-8. 2-D median filtering technique; at upper side the blurring image with histogram and at lower side the enhanced image using 2-D median filtering technique from test (A4)

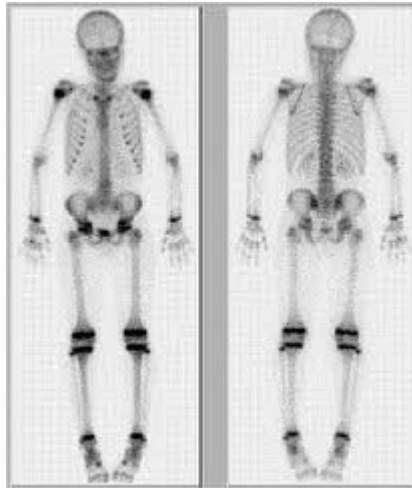


Figure 4-9. The original bone scan image test (A5)

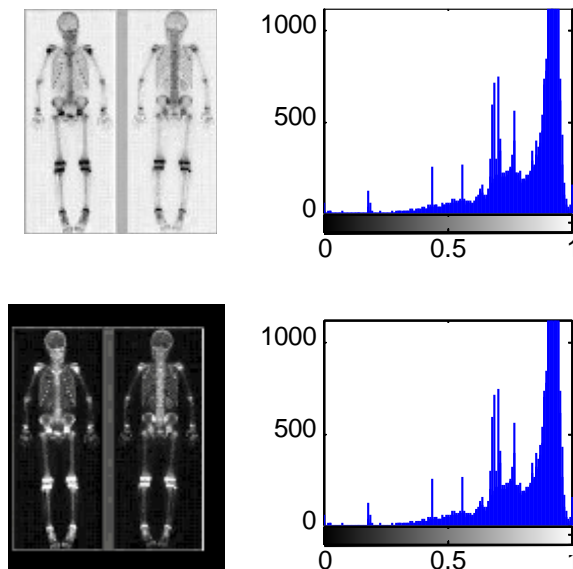


Figure 4-10. Contrast stretch image technique; at upper side the original image with histogram and at lower side the enhanced image using contrast stretch image technique from test (A5)

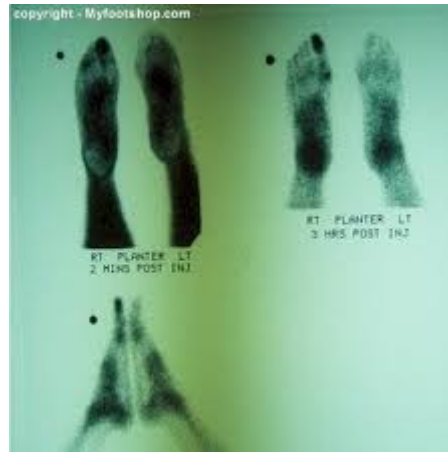


Figure 4-11. The original bone scan image test (B1)

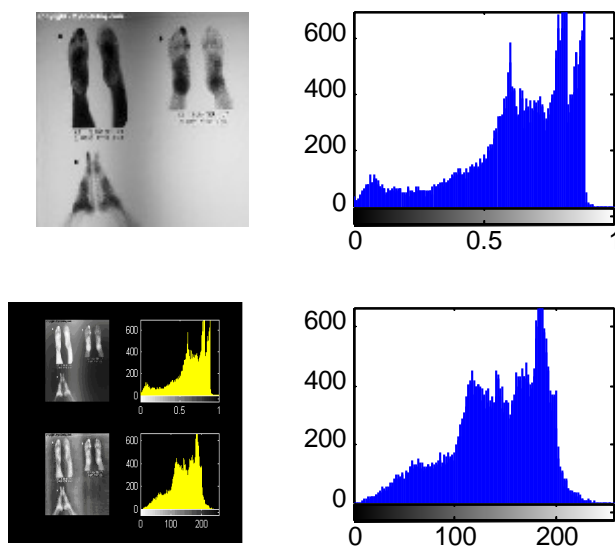


Figure 4-12. CLAHE technique; at upper side the gray scale image with histogram and at lower side the enhanced image using CLAHE technique from test (B1)

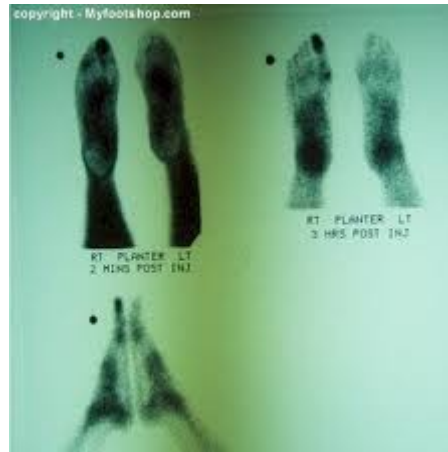


Figure 4-13. The original bone scan image test (B2)

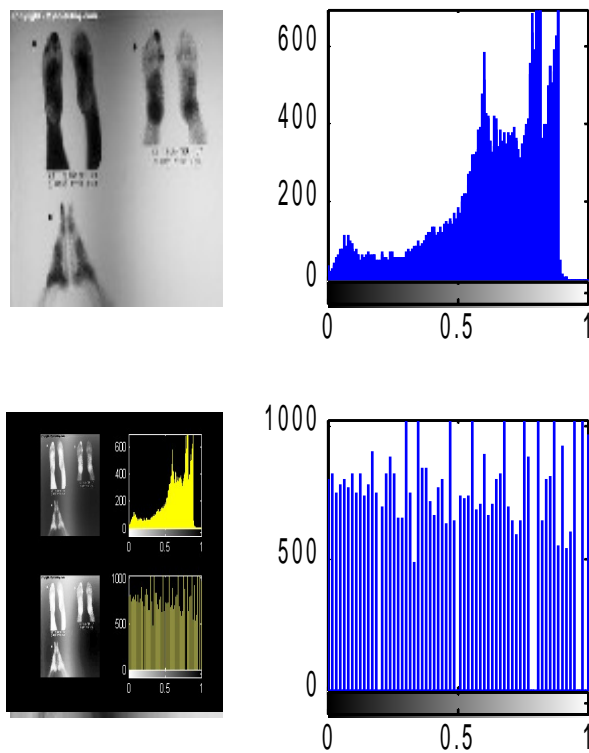


Figure 4-14. histogram equalization technique; at upper side the gray scale image with histogram and at lower side the enhanced image using histogram equalization technique from test (B2)

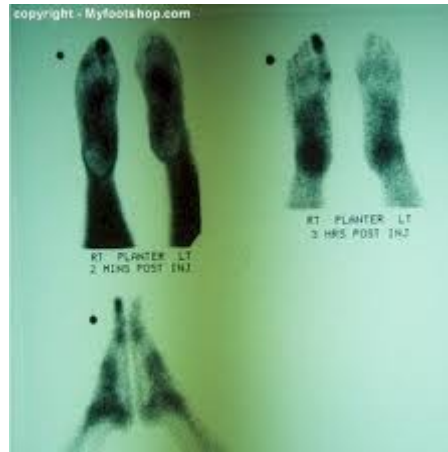


Figure 4-15. The original bone scan image test (B3)

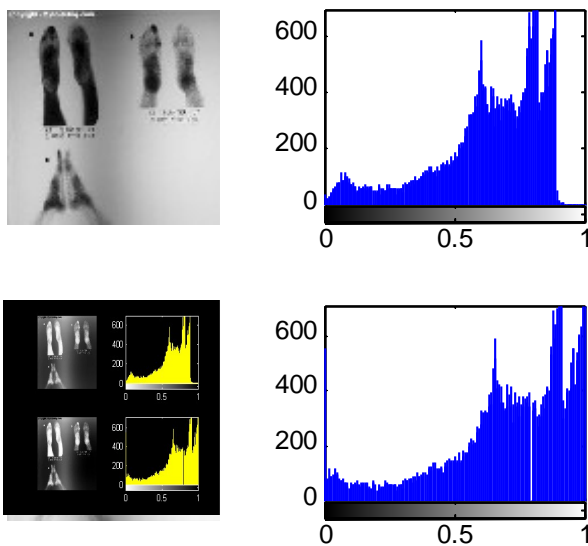


Figure 4-16. Adjust technique; at upper side the gray scale image with histogram and at lower side the enhanced image using Adjust technique from test (B3)

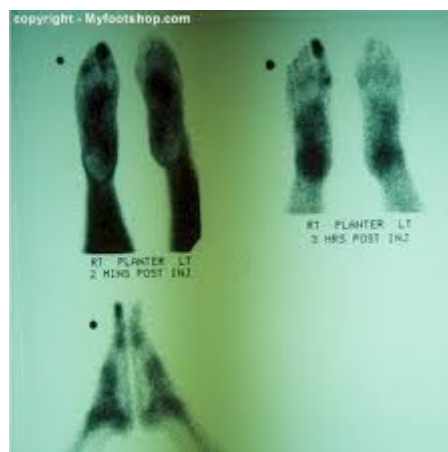


Figure 4-17. The original bone scan image test (B4)

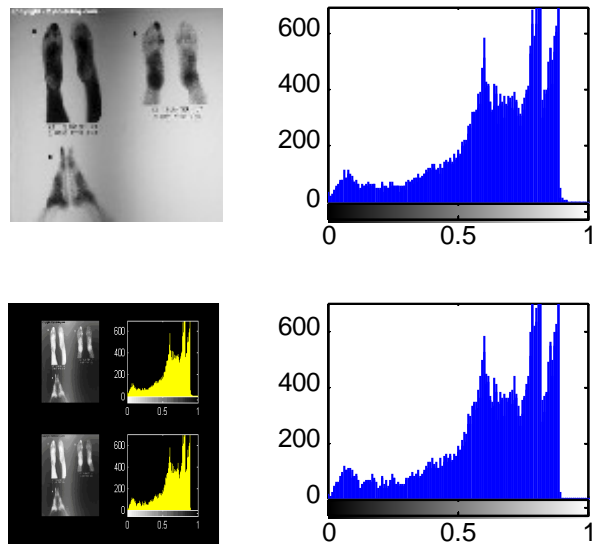


Figure 4-18. 2-D median filtering technique; at upper side the blurring image with histogram and at lower side the enhanced image using 2-D median filtering technique from test (B4)

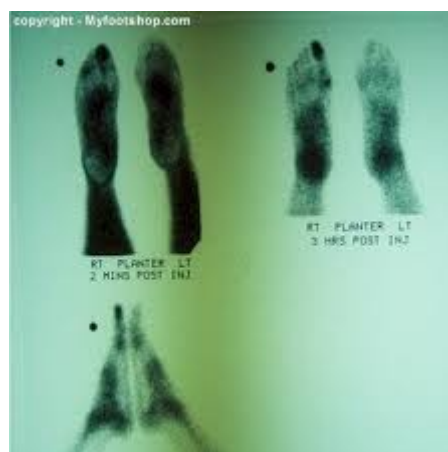


Figure 4-19. The original bone scan image test (B5)



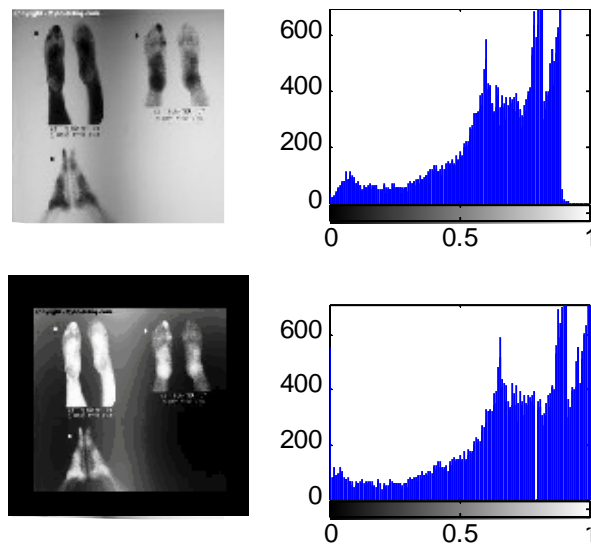


Figure 4-20. Contrast stretch image technique; at upper side the original image with histogram and at lower side the enhanced image using contrast stretch image technique from test (B5)Used four types of techniques

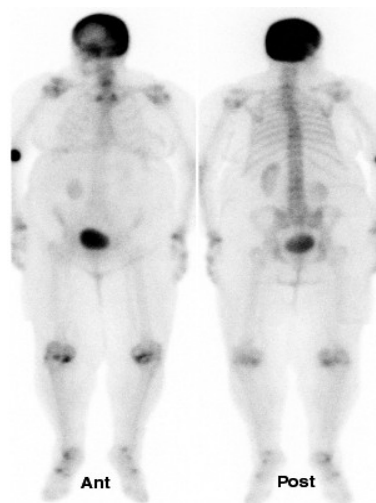


Figure 4-21. The original bone scan image test (C1)

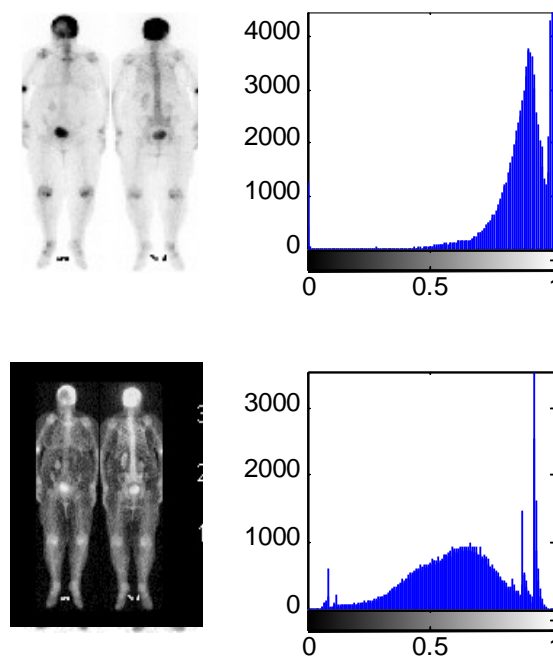
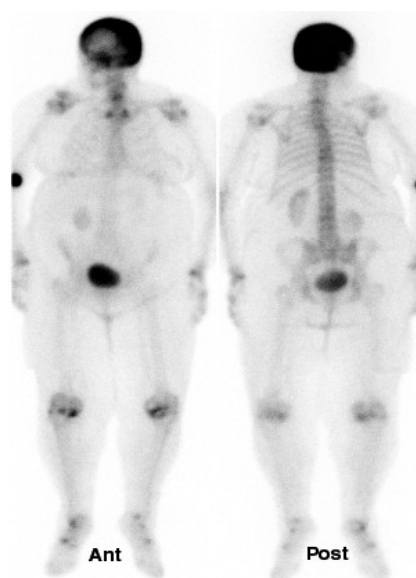


Figure 4-22. CLAHE technique; at upper side the gray scale image with histogram and at lower side the enhanced image using CLAHE technique from test (C1)



**Figure 4-23. The original bone scan image test (C2)**

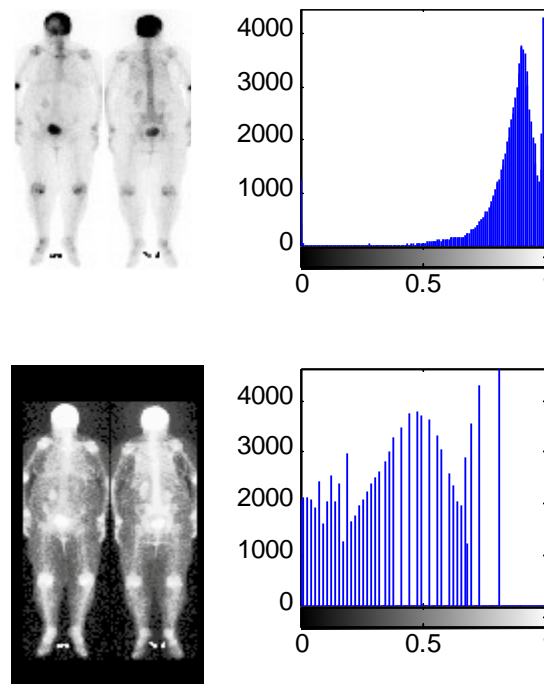


Figure 4-24. histogram equalization technique; at upper side the gray scale image with histogram and at lower side the enhanced image using histogram equalization technique from test (C2)

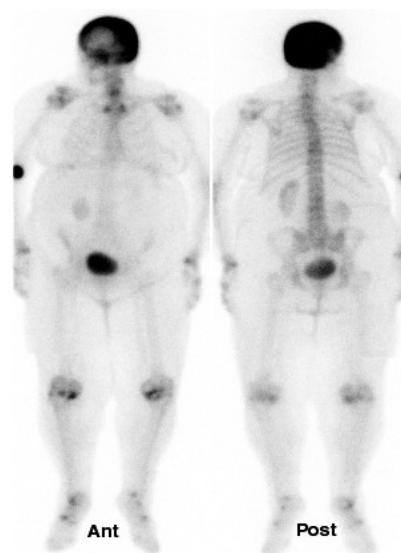


Figure 4-25. The original bone scan image test (C3)

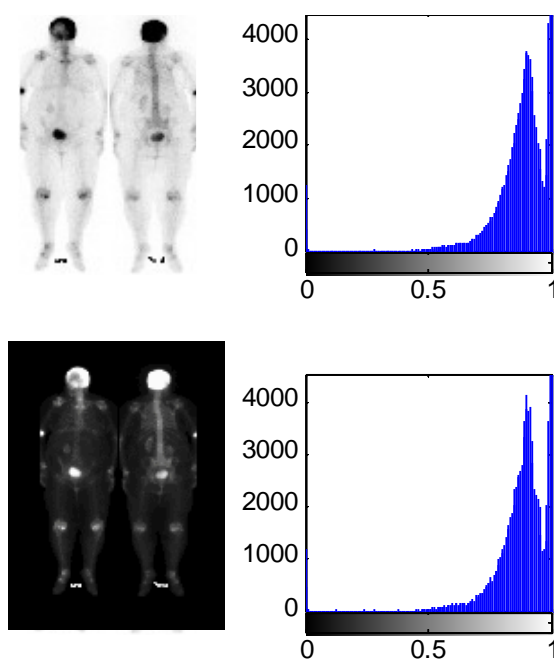


Figure 4-26. 2-D median filtering technique; at upper side the blurring image with histogram and at lower side the enhanced image using 2-D median filtering technique from test (C3)

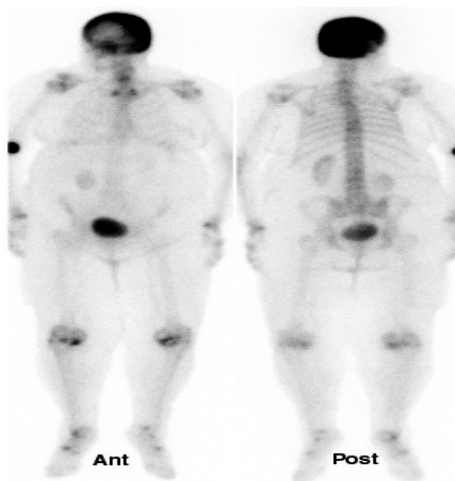
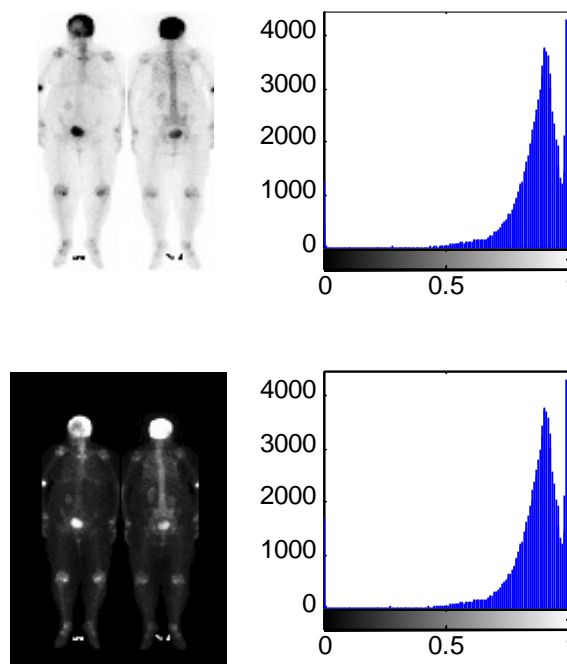


Figure 4-27. The original bone scan image test (C4)



*Figure 4-28. Contrast stretch image technique; at upper side the gray scale image with histogram and at lower side the enhanced image using contrast stretch image technique from test (C4)*

More cases studies for patient 4 and 5 were displayed in appendix from page 52-55

## Chapter Five

### Discussion, Conclusion and Recommendations

#### 5.1.Discussion:

Nuclear medicine images show characteristic information about the physiological properties of the structures-organs. In order to have high quality medical images for reliable diagnosis, the processing of image is necessary. The scope of image processing and analysis applied to medical applications is to improve the quality of

the acquired image and extract quantitative information from medical image data in an efficient and accurate way. In Nuclear Medicine, there are two main methods of patient imaging, the imaging with Planar Imaging, Dynamic Imaging or SPECT and the PET. In this study data analyzed by using MatLab program to enhance the contrast within the bones, the gray levels in both enhanced and unenhanced images and noise variance. The technique used for this study were Contrast-limited adaptive histogram equalization (CLAHE) which operated on small regions in the image, called tiles, rather than the entire image, each tile's contrast was enhanced, so that the histogram of the output region approximately matches the histogram specified by the ['Distribution'](#) parameter., Enhance contrast using so that the histogram of the output region approximately matches the histogram specified by the ['Distribution'](#) parameter. The neighboring tiles were then combined using bilinear interpolation to eliminate artificially induced boundaries. The contrast, especially in homogeneous areas, could be limited to avoid amplifying any noise that might be present in the image as shown in figure 4-2 and figure 4-12. The histogram equalization enhanced the contrast of images (bone scan) by transforming the values in an intensity image, or the values in the colormap of an indexed image, so that the histogram of the output image approximately matched a specified histogram as shown in figure 4-4 and figure 4-14. Adjust image intensity values or colormap used to enhance the images of bone scan by increased of contract of the image, it mapped the intensity values in grayscale image as shown in original image (figure 4-1). 2-D median filtering used as nonlinear operation which often used in image processing to reduce "salt and pepper" noise in figure 4-7 and figure 4-12 from the images (bone scan). Contrast stretch image used to enhance of the images (bone scan) within which a two-element vector of pixel values that specify lower and upper limits that can be used for contrast stretching

image as in figure 4-10 and figure 4-20. The results of this technique agreed the results of Robiul et al, (2011), Nasrul et al, (2012), Gupta et al, (2012) and Smriti et al, (2012) who used non-linear filtering based methods to enhance the nuclear medicine images. The another technique was Convolution kernel filter. Filtering is a technique for modifying or enhancing an image. For example, researchers can filter an image to emphasize certain features or remove other features. Image processing operations implemented with filtering include smoothing, sharpening, and edge enhancement. Filtering is a neighborhood operation, in which the value of any given pixel in the output image is determined by applying some algorithm to the values of the pixels in the neighborhood of the corresponding input pixel. A pixel's neighborhood is some set of pixels, defined by their locations relative to that pixel. Linear filtering is filtering in which the value of an output pixel is a linear combination of the values of the pixels in the input pixel's neighborhood.

## **5.2.Conclusion:**

- The very small details of bone scan could be clearly appear at used this techniques.
- The histogram is appear the significant different of enhance the images.
- The quality of images is increased by using Matlab techniques.

- Implemented the image processing operations with filtering to smoothing, sharpening and edge enhancement.
- Used non-linear filtering based methods to enhance the nuclear medicine images by reduce of "salt and pepper" noise.

### **5.3.Recommendations:**

- The special, noise removal and Wiener filtering are recommended to study x-rays images in order to reduce the speckle without fully eliminating the image edges.
- Using both hat-top and blind Deconvolution algorithm is recommended as a new approach of noise estimation using image processing technique (MATLAB).

### **5.4. References:**

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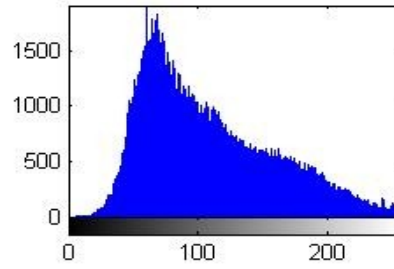
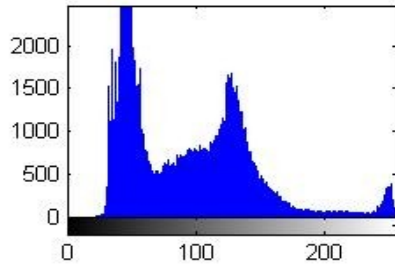
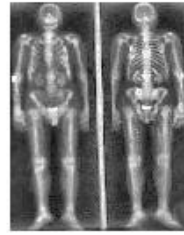
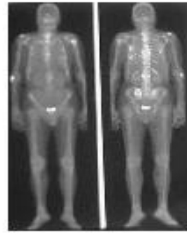
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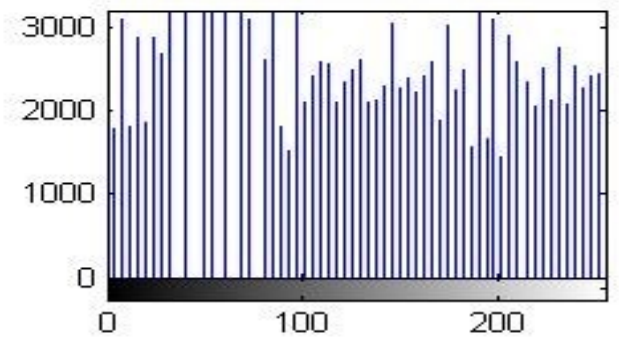
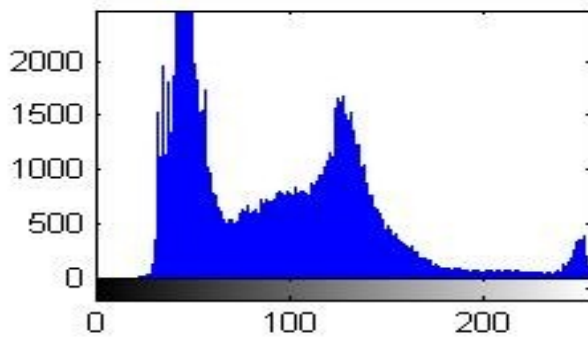
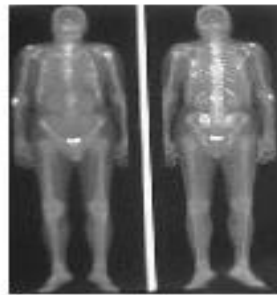
## **Appendix**

### **Patient 4:**

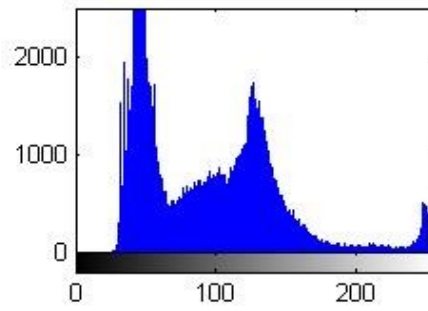
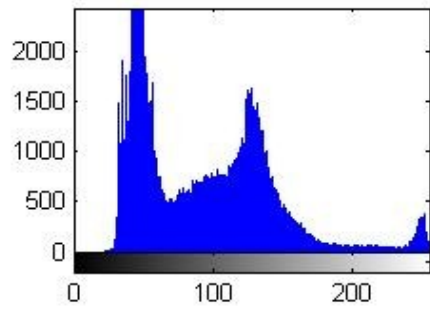
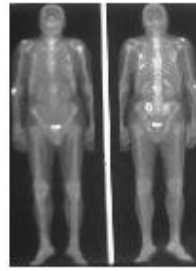
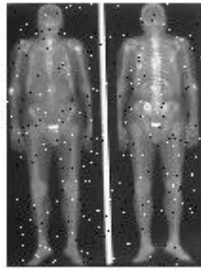
#### **(CLAHE) code**



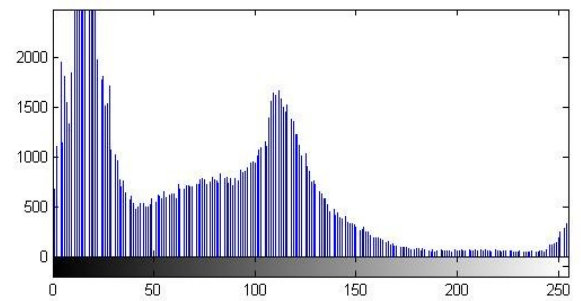
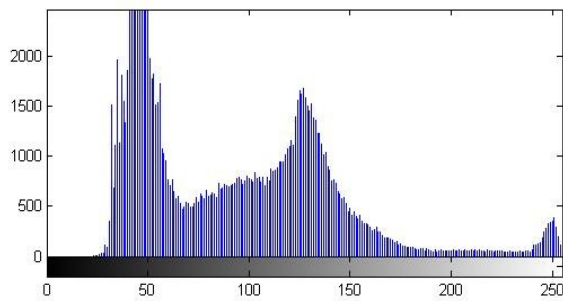
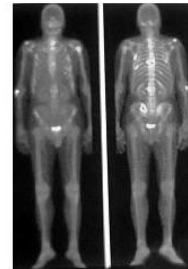
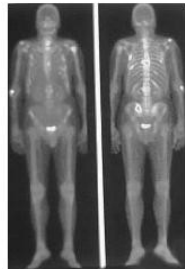
### Histogram code



### 2-D median filtering

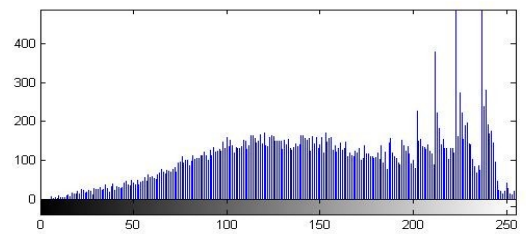
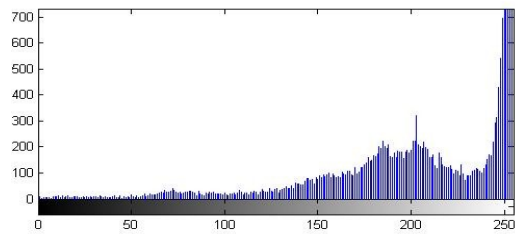
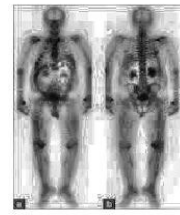
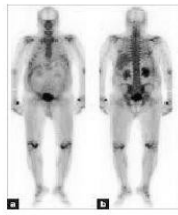


**Stretch image**

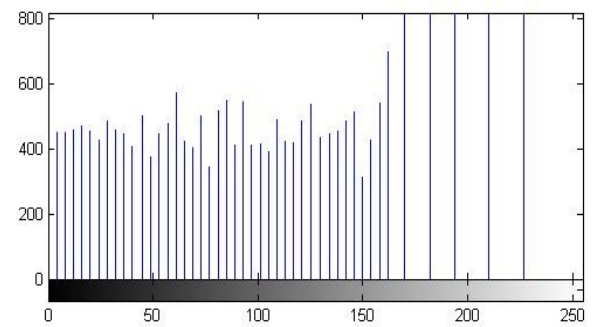
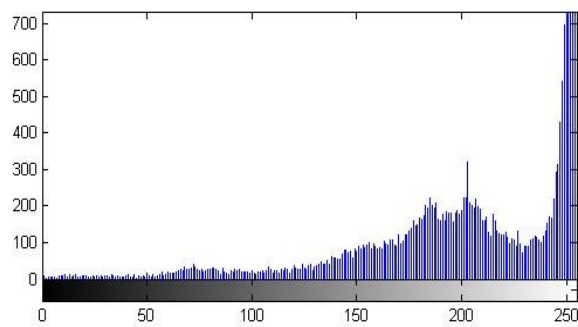
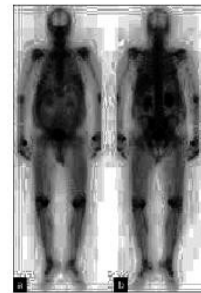


**Patient 5:**

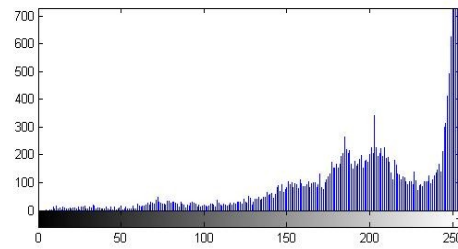
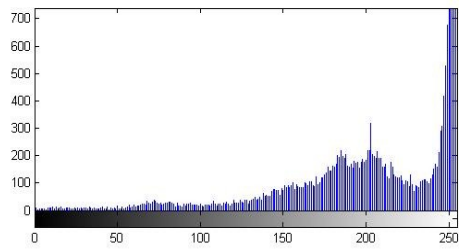
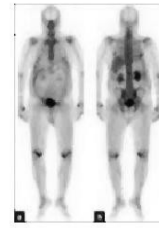
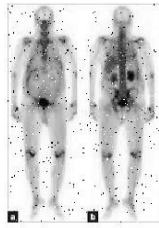
**(CLAHE) code**



## Histogram code



## 2-D median filtering



## Stretch image

