

1.1 Introduction

Nuclear medicine provides noninvasive imaging tools to detect a variety of human diseases. The two major components of these imaging procedures are radiopharmaceuticals and an imaging system (Magdy M. K., 2011).

Radionuclide imaging is the most important application of radioactivity in nuclear medicine. The purpose of radionuclide imaging is to obtain a picture of the distribution of a radioactively labeled substance within the body after it has been administered to a patient. This is accomplished by recording the emissions from the radioactivity with external radiation detectors placed at different locations outside the patient. The preferred emissions for this application are gamma rays in the approximate energy range of 80 to 500 KeV (or annihilation photons, 511 KeV). Gamma rays of these energies are sufficiently penetrating in body tissues to be detected from deep-lying organs, can be stopped efficiently by dense scintillators, and are shielded adequately with reasonable thicknesses of lead (Simon R. C., et al, 2012).

1.1.1 The Gamma Camera

The gamma or scintillation camera is an imaging device that is most commonly used in nuclear medicine. It is also called the Anger camera in honor of Hal O. Anger, who invented it in the late 1950s. Gamma cameras detect radiation from the entire field of view simultaneously and therefore are capable of recording dynamic as well as static images of the area of interest in the patient. Various designs of gamma cameras have been proposed and made available, but the Anger camera with a single crystal is by far the most widely used. Although many sophisticated improvements have been made on the gamma cameras over the years, the basic principles of the operation have essentially remained the same (Gopal B. S., 2012).

The gamma camera usually consists of several components: a detector, a collimator, PM tubes, a preamplifier, an amplifier, a pulsed-height analyzer (PHA), an X-, Y-positioning circuit, and a display or recording device. A schematic diagram of a gamma camera is illustrated in Figure (1.1). The detector, PM tubes, and amplifiers are housed in a unit called the *detector head*, which is mounted on a stand. The head can be moved up or down and rotated with electrical switches to position it in the field of view on the patient. The X-, Y-positioning circuits, PHA, and some recording devices are mounted on a console. In the past, the cameras were operated by switches and dials on the console. Currently, much of the operation of the camera is performed by a computer built in it. The computer is run by appropriate software in conjunction with a keyboard, a mouse, and a video monitor. High voltage, window, and photopeaks are all set by the operator's choice of parameters (Gopal B. S., 2012).

Acquisition of the data and processing of the data are carried out by the computer. Whereas stationary cameras are permanently installed at desired locations, portable gamma cameras are mounted on wheels for use in situations requiring movement of the camera from room to room, such as to the patient's bedside. Mobile cameras are installed in wheeled vans such that they can be moved to places where gamma cameras are not available for nuclear medicine studies (Gopal B. S., 2012).

The operational principles of a gamma camera are identical to those of solid scintillation counters. Basically, gamma-rays from a source interact with the NaI(Tl) detector, and light photons are emitted. The latter strike the photocathode of PM tubes, and a pulse is generated, which is then amplified by an amplifier and sorted out by a PHA. Finally, the pulse is positioned by an X-, Y-positioning circuit on the recording device or stored in the computer, corresponding to the location of gamma-ray interaction in the detector (Gopal B. S., 2012).

Most modern gamma cameras are completely digital, in the sense that the output of each PM tube is directly digitized by an analog to digital converter (ADC). The calculation of X-Y position and pulse-height are performed in software based on the digitized PM tube signals, and errors in energy and positioning caused by noise and pulse distortions caused by the analog positioning circuitry are eliminated (Simon R. C., et al, 2012).

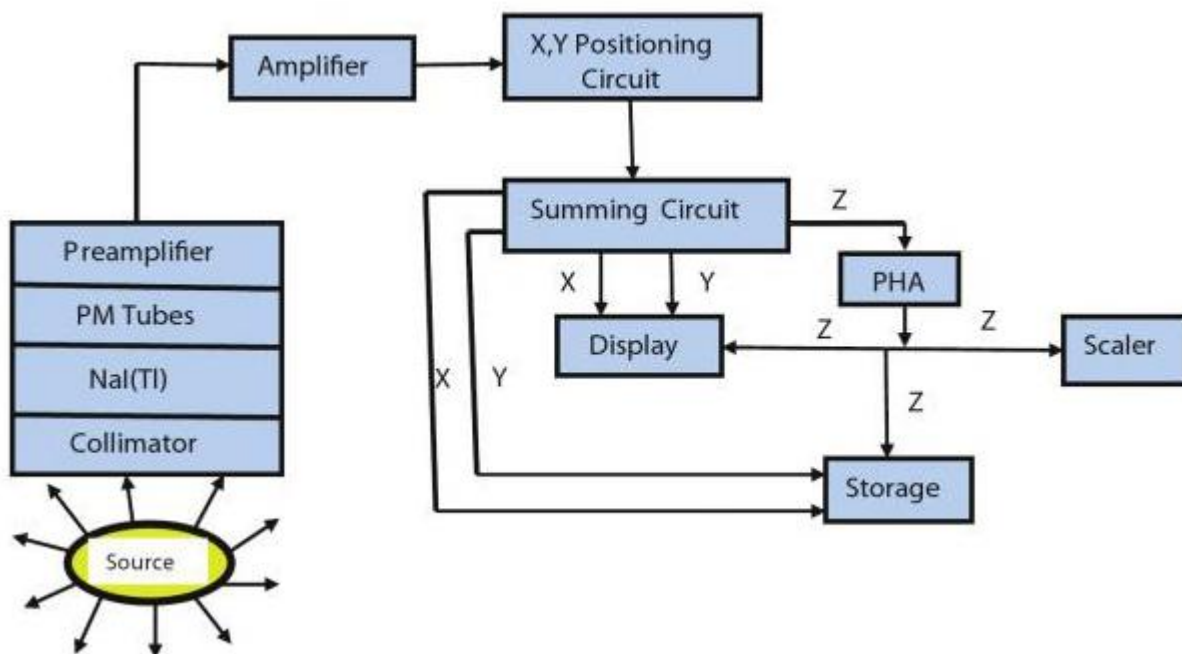


Figure 1.1: A schematic diagram of a gamma camera.

1.1.2 Gamma Camera Performance & Quality Control

Quality control (QC), a critical component of analytical procedures and instrumentation in general and medical devices in particular, is an established set of ongoing measurements and analyses designed to ensure that the performance of

a procedure or instrument is within a pre-defined acceptable range. An extensive series of parameters have been developed over the years to characterize gamma camera, SPECT scanner, and PET scanner performance, and detailed data acquisition and analysis protocols have been promulgated by the National Electrical Manufacturers Association (NEMA), the American Association of Physicists in Medicine (AAPM), and others for this purpose. In practice, however, less extensive and less rigorous procedures have proven adequate for day-to-day QC (Hans, et al, 2007).

The performance of a gamma camera is characterized by six parameters, spatial resolution, nonuniformity, spatial distortion, sensitivity, count rate characteristic, and energy resolution (Peter F. S., et al, 2005).

A typical quality assurance program might involve daily measures of flood-field uniformity, weekly checks of spatial resolution and spatial linearity, and semiannual checks of other performance parameters. It is important that all measurements be taken under the same conditions (pulse-height window width, correction algorithms, and correction circuitry on or off) as are used for routine clinical studies (Simon R. C., et al, 2012).

1.1.3 Uniformity

The most sensitive parameter to changes in system performance is system uniformity. Changes in photopeak location, photomultiplier tube (PMT) performance, energy and linearity correction, etc all affect image uniformity. Hence this is probably the single most important QC test that can be performed on a gamma camera system and one that should be performed on a daily basis (Michael K. O'Connor).

Intrinsic uniformity is determined from floodfield images acquired without a collimator. A ^{99m}Tc source is placed at a distance of approximately $5 \times$ the UFOV from the front face of the gamma camera. The source activity is such that the counting rate on the gamma camera is less than approximately 30,000 cps. Flood-field images are acquired so that there are a minimum of 4000 counts in each pixel of the image and then smoothed with a 9-point (3×3) smoothing filter with the following weighting:

$$1 \quad 2 \quad 1$$

$$2 \quad 4 \quad 2$$

$$1 \quad 2 \quad 1$$

Integral uniformity is based on the maximum and minimum pixel counts in the image and is defined as

$$\text{Integral uniformity (\%)} = (C_{\max} - C_{\min}) / (C_{\max} + C_{\min}) \times 100$$

This is calculated for the UFOV and CFOV.

Differential uniformity is based on the change in counts of five consecutive pixels across all rows and columns of the image. It is defined as

$$\text{Differential uniformity (\%)} = (\text{high} - \text{low}) / (\text{high} + \text{low}) \times 100$$

Where “high” refers to the maximum count difference for any five consecutive pixels (row or column) in the image and “low” refers to the minimum count difference for any five consecutive pixels. This usually is reported for the UFOV.

For convenience, uniformity measurements often are made with the collimator in place (extrinsic uniformity). A thin flood-field source of ^{99m}Tc or a disk source of

^{75}Co that covers the active area of the gamma camera is placed on top of the collimator to provide uniform irradiation. This protocol is more practical for routine quality assurance because the measurement can be done without removing the collimator. Extrinsic uniformity measurements also have the advantage that they reveal any defects or problems caused by the collimator itself (Simon R. C., et al, 2012).

1.2 Problem of Study

Flood-field uniformity evaluation is an essential element in the assessment of the gamma camera performance. It serves as the central element of a NM quality control (QC) program, regularly performed on a daily basis. Uniformity assessment usually uses quantitative pixel value-based metrics. However, these metrics sometime fail to reveal structural patterns in the image requiring additional subjective visual examination which can be difficult in a busy department.

1.3 Objectives

The general objective of this research is to Study and apply a new quantitative uniformity analysis technique to the gamma camera uniformity test.

Specific objectives

- To calculate the integral and differential uniformity using an algorithm
- To enhance the uniformity image using shade surface (3D) for better visual perception.
- To generate a correction map and apply it to non uniform image

1.4 Thesis outlines

This thesis will be consisting of five chapters. Chapter one will deal with the introduction, objective, and theses outline. Chapter two will deals with the gamma camera, and quality control, previous study. Chapter three will show the material and method. Chapter four will show the results. Chapter five will show the discussion, recommendations, and references.

CHAPTER TWO

2.1 The Gamma Camera System

2.1.1 Introduction

The gamma camera is the principal instrument for imaging in nuclear medicine and is shown in Figure (2.1). As can be seen, it consists of a large detector in front of which the patient is positioned. Gamma cameras with more than one detector are now common, allowing a higher throughput of patients by acquiring two or more views simultaneously. Every aspect of the modern gamma camera is under computer control, allowing the operator to select the study acquisition time, or the numbers of counts to be acquired, to set the pulse height analyzers to reject scattered radiation, control the detector and patient bed positions for SPECT and whole body procedures, and display the image (Peter F. S., et al, 2005).



Figure (2.1) The gamma camera. The patient is lying between the detectors of this double-headed system. To the right of the camera is a rack containing extra

collimators. The technician is seated at the computer controlling data acquisition and image display.

2.2 Mode of Operation of the Gamma Camera

The basic principles of how a camera works are shown in Figure (2.2). The image of the distribution of the gamma-ray-emitting radiopharmaceutical is produced in the scintillation crystal by a collimator. The gamma rays, which are not visible to the eye, are converted into flashes of light by the scintillation crystal. This light is, in turn, transformed into electronic signals by an array of photomultiplier tubes (PMT) viewing the rear face of the crystal. After processing, the outputs from the PMTs are converted into three signals, two of which (X and Y) give the spatial location of the scintillation while the third (Z) represents the energy deposited in the crystal by the gamma ray. To improve their quality these signals then pass through correction circuits. The Z signal goes to a pulse height analyzer (PHA), which tests whether the energy of the gamma ray is within the range of values expected for the particular radionuclide being imaged. If the Z signal has an acceptable value, then a signal is sent instructing the display to record that there has been a gamma ray detected, the position being determined by the X and Y signals (Sharp, et al, 2005).

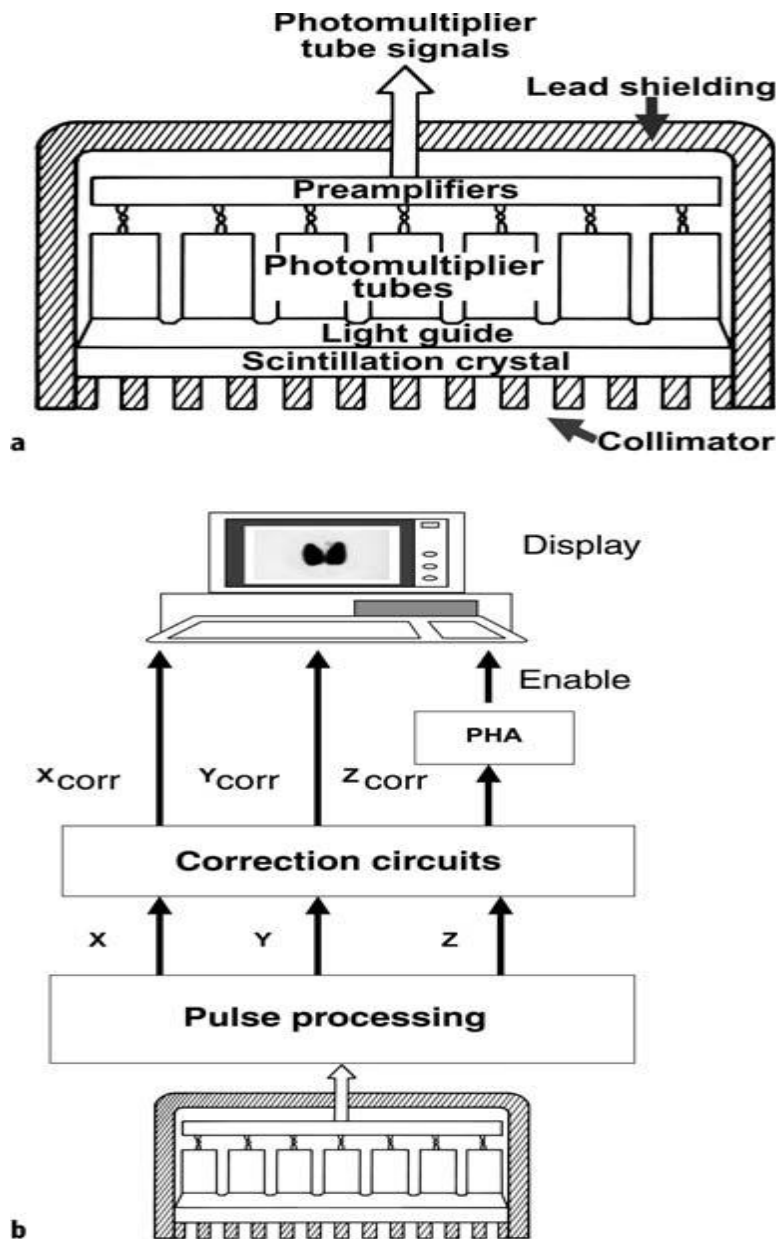


Figure (2.2) a Cross-section through the detector head of a gamma camera. Gamma rays emitted from the patient pass through the collimator to form an image in the scintillation crystal. The light from this image is converted into electronic signals by the PMTs. **b** The signals from the camera head are processed to give the X and Y position signals and the Z energy signal. The Z signal goes to a pulse height analyzer and, if it falls within the predetermined range of acceptable energy values, generates a signal which instructs the display system to record a gamma ray as having been correctly detected at the X and Y location

2.3 Collimators

The collimator of a scintillation camera forms the projection image by permitting x- or gamma-ray photons approaching the camera from certain directions to reach the crystal while absorbing most of the other photons. Collimators are made of high atomic number, high-density materials, usually lead. The most commonly used collimator is the *parallel-hole collimator*, which contains thousands of parallel holes (Figure 2.3). The holes may be round, square, or triangular; however, most state-of-the-art collimators have hexagonal holes and are usually made from lead foil, although some are cast. The partitions between the holes are called septa. The septa must be thick enough to absorb most of the photons incident upon them. For this reason, collimators designed for use with radionuclides that emit higher-energy photons have thicker septa. There is an inherent compromise between the spatial resolution and efficiency (sensitivity) of collimators. Modifying a collimator to improve its spatial resolution (e.g., by reducing the size of the holes or lengthening the collimator) reduces its efficiency. Most scintillation cameras are provided with a selection of parallel-hole collimators. These may include "low-energy, high-sensitivity," "low-energy, all-purpose" (LEAP), "low-energy, high-resolution", "medium-energy" (suitable for Ga-67 and In-111), "high-energy" (for I-131), and "ultra-high-energy" (for F-18) collimators. The size of the image produced by a parallel-hole collimator is not affected by the distance of the object from the collimator. However, its spatial resolution degrades rapidly with increasing collimator-to-object distance (Jerrold T. B., et al, 2002).

A pinhole collimator (Fig. 2.3) is commonly used to produce magnified views of small objects, such as the thyroid or a hip joint. It consists of a small (typically 3- to 5-mm diameter) hole in a piece of lead or tungsten mounted at the apex of a leaded cone. Its function is identical to the pinhole in a pinhole photographic

camera. As shown in the figure, the pinhole collimator produces a magnified image whose orientation is reversed. The magnification of the pinhole collimator decreases as an object is moved away from the pinhole. If an object is as far from the pinhole as the pinhole is from the crystal of the camera, the object is not magnified and, if the object is moved yet farther from the pinhole, it is minified. (Clinical imaging is not performed at these distances.) There are pitfalls in the use of pinhole collimators due to the decreasing magnification with distance (Jerrold T. B., et al, 2002).

A converging collimator (Fig. 2.3) has many holes, all aimed at a focal point in front of the camera. As shown in (Fig. 2.3), the converging collimator magnifies the image. The magnification increases as the object is moved away from the collimator. A diverging collimator (Fig. 2.3) has many holes aimed at a focal point behind the camera. It produces a minified image in which the amount of minification increases as the object is moved away from the camera. A diverging collimator may be used to image a large portion of a patient on a small (25-cm diameter) or standard (30-cm diameter) field-of-view (FOV) camera. If a diverging collimator is reversed on a camera, it becomes a converging collimator. The diverging collimator is seldom used today, as it has inferior imaging characteristics to the parallel-hole collimator and the large crystal sizes of most modern cameras render it unnecessary (Jerrold T. B., et al, 2002).

The converging collimator is also seldom used; its imaging characteristics are superior, in theory, to the parallel-hole collimator, but its decreasing FOV with distance and varying magnification with distance have discouraged its use. However, a hybrid of the parallel-hole and converging collimator, called a fan-beam collimator, is used in single photon emission computed tomography

(SPECT) to take advantage of the favorable imaging properties of the converging collimator (Jerrold T. B., et al, 2002)

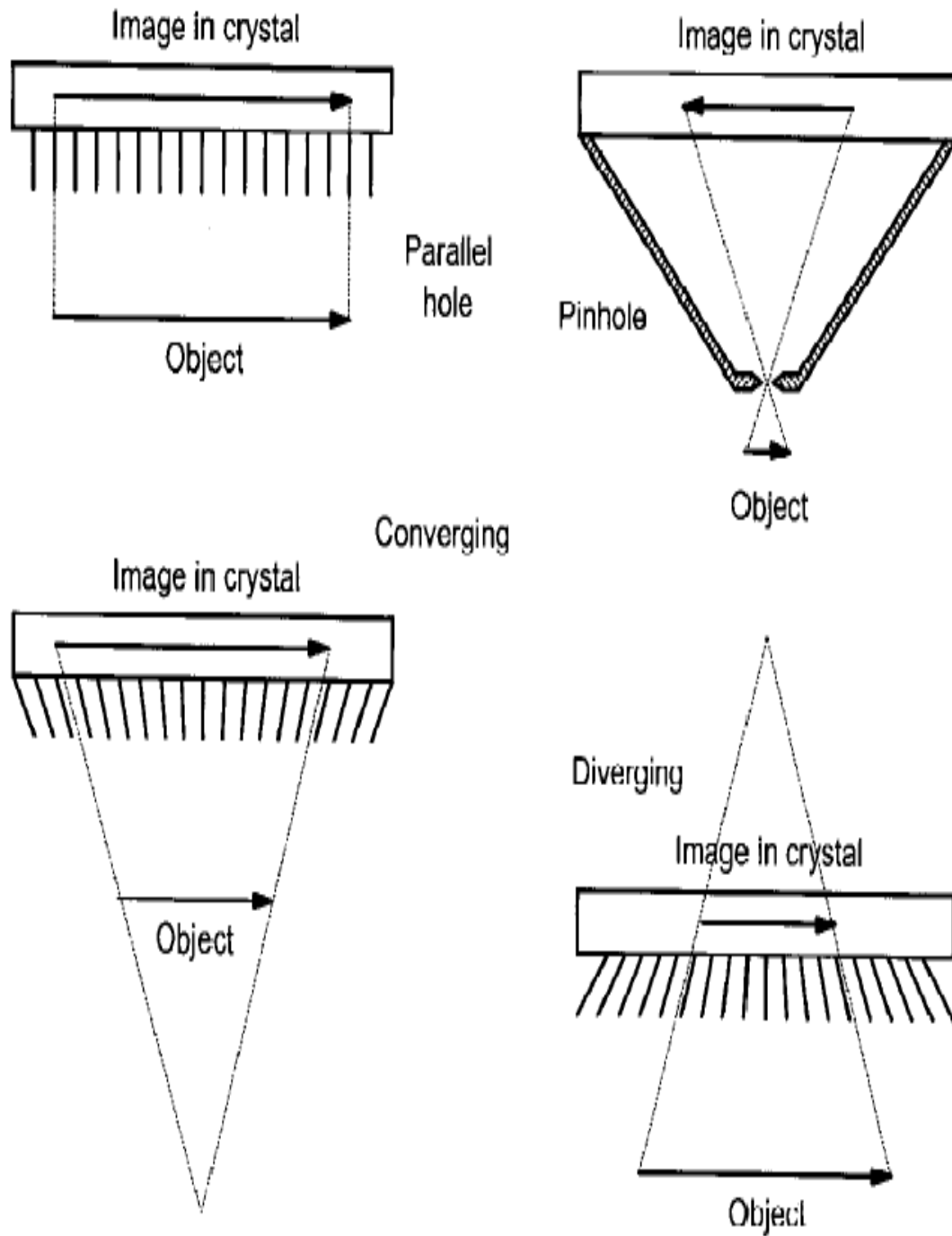


Figure (2.3) collimators

2.4 Detectors

While the collimator modifies the gamma ray flux so as to create an image, it is the function of the detector assembly to convert the gamma rays into a form that will, eventually, allow a visible image to be produced. This process takes place in two stages. The first step is the conversion of the gamma rays into visible light by means of a scintillation crystal, while in the second these scintillations are turned into electrical signals by the PMTs. The scintillation crystal used in gamma cameras is made from sodium iodide with trace quantities of thallium added, NaI(Tl). Its effectiveness at stopping the gammas depends not only on its density but also on the thickness of crystal used (Peter F. S., et al, 2005).

Unfortunately, only a small fraction of the energy lost by a gamma ray is converted into light, typically 10%, producing about 3000 light photons at a wavelength of 410 nm for each 100 keV of gamma ray energy absorbed (Peter F. S., et al, 2005).

The desirable properties of the scintillation crystal are:

- High efficiency for stopping gamma rays
- Stopping should be without scatter
- High conversion of gamma ray energy into light
- Wavelength of light should match response of the PMTs
- Crystal should be transparent to emitted light
- Crystal should be mechanically robust
- Length of scintillation should be short

2.5 Signal Processing

Three types of signal processing are to be found in use at present. Analogue circuitry was used in cameras exclusively until about the mid-1990s. Even though so-called digital cameras have been marketed since the mid-1980s, in practice they used analogue circuits to produce the X, Y, and Z signals and digitized them prior to them entering the signal correction and display modules. Only since the mid-1990s has truly digital signal processing been used in commercial cameras, with the signal from the individual PMTs being digitized (Peter F. S., et al, 2005).

2.5.1 Analogue Systems

The signals from the PMTs are processed to give the three signals required, X and Y providing spatial information and Z the energy. The energy signal is produced simply by summing the outputs from all of the tubes, so measuring the total light produced by the scintillation. The spatial information is more difficult to produce. What is required is that the processed signal should be proportional to the X or Y location of the scintillation. This is achieved by weighting the signals from each tube by passing the output from the PMTs through resistors or capacitors:

$$X = \sum w_i o_i / o_i$$

where w_i is the weighting factor for the i th PMT and o_i is the output signal. The value of the weighting factor is proportional to the spatial coordinate of the PMT, separate factors being used for the X and Y signals. The divisor, which is equal to the energy of the gamma, is necessary to prevent the X value being dependent upon the energy as well as spatial location of the scintillation (Peter F. S., et al, 2005).

2.5.2 Digital Systems

Modern camera systems now rely heavily on digital technology, both for correction of the spatial, energy, and temporal information and for the analysis of image data. Since this requires the analogue signal to be digitized, there is a strong case for having a completely digital camera. Two general approaches are currently taken; either to digitize the X, Y, and Z signals immediately after these have been computed by analogue circuitry or, in the latest generation of cameras, to completely replace the analogue circuits, the signals from the PMTs being digitized before the X, Y, and Z signals are computed. In the latter case each PMT has an analogue to digital converter located after the preamplifier, and the signal position is calculated from the centroid of the signals from a group of PMTs, usually chosen to be those having the strongest and hence least noisy signals. The Z signal is calculated by summing the outputs from the group of tubes (Peter F. S., et al, 2005).

2.5.3 Accuracy of Signal Processing

In order to reproduce the image that has been formed in the scintillation crystal the X and Y signals must be proportional to the coordinates of the scintillation. In other words the system must demonstrate good spatial linearity. The detector assembly must be constructed so that small changes in the position of the scintillation alter significantly the relative strengths of the outputs from the PMTs (Peter F. S., et al, 2005).

2.6 Pulse Height Analysis

The collimator is responsible for creating an image out of the flux of gamma rays incident on it, yet not all of these gamma rays will carry useful image information.

In particular, those rays that have been Compton scattered in the patient, and so appear to come from another location, simply reduce image contrast. Such rays can be identified by the fact that in being scattered they also lose energy, the amount being dependent on the angle through which they were deviated. As the scintillation detector allows the energy deposited by the gamma ray in the crystal to be measured, this information can be used to exclude scattered gamma rays. However, as has already been mentioned, the gamma camera is limited in the accuracy with which it can measure energy, with the result that even in the absence of scatter the gamma rays appear to have a range of energies. The width of this so-called photopeak spectrum, measured at half the maximum height, is about 10% of the true energy. The pulse height analyzer allows the operator to select only the signals from those gammas in which the height of the Z signal, that is, gamma ray energy, has a certain value or range of values. If many useful gammas are not to be excluded from the image, a range of energies must be allowed through the PHA, and typically a window equal to 20% of the peak energy value is used; i.e. for ^{99m}Tc with a gamma ray of 140 keV those signals with energies between 126 and 154 keV are judged to be acceptable (Peter F. S., et al, 2005).

While the scattered gamma rays are distinguishable by their lower energy, the spreading out of the photopeak spectrum may mean that the energy of unscattered gammas overlaps that of scattered ones. This overlap of the Compton and photopeak spectrum means that a choice must be made between using a narrow window and so excluding a large proportion of unscattered rays or accepting the presence of some scattered radiation in the image. Usually the latter is chosen, and with a 20% window about 30% of the gamma rays in the image will have been scattered. When using radionuclides that emit gamma rays at different energies, multiple-window-analyzers need to be employed. Typically a maximum of three

sets of windows is available. In this instance, it is important to remember that scattered radiation from the higher-energy gammas may overlap into the lower-energy photopeaks and this may influence which gamma ray energies should be selected (Peter F. S., et al, 2005).

2.7 Correction Circuits

To improve image quality, real-time compensation is provided for some of the defects in camera performance. It is, however, only possible to correct where the cause of the distortion is not random. So, while non-linearity can be corrected, it is not possible to improve the intrinsic spatial resolution of the camera (Peter F. S., et al, 2005).

2.7.1 Spatial Linearity

Spatial linearity correction is carried out by presenting the camera with an image consisting of a series of parallel straight lines aligned with either the X or Y axis of the camera. The deviation between the true position of each point on the line, as calculated from a best-fit straight line, and the image of the line, is recorded and stored as a correction factor to be applied to subsequently acquired clinical images. To achieve the spatial resolution required to define the straight lines it is necessary for the test pattern to be imaged without the collimator, i.e. placed against the crystal and illuminated with a flood source of gamma rays, usually from a point source of activity placed a long way from the camera. As this requires a phantom manufactured to high tolerances, the generation of new correction factors is usually done by the manufacturer's service engineer (Peter F. S., et al, 2005).

2.7.2 Energy

The energy signal may also require correction, as the calculated value is found to vary with the position of the gamma ray across the detector. Shifts in the position of the locally computed energy signal can be measured by comparing the number of gamma rays whose energy falls inside two narrow windows, one set to each side of the average photopeak value. Once again the amount of correction required to ensure that the local signal matches the average value, i.e. that there are equal numbers of gamma rays within each window, is stored in a correction matrix to be applied to the subsequent energy signals. This correction is acquired simply by imaging a flood source of radioactivity and can be performed regularly by the user. Typically the average correction should yield a variation in the energy signal of less than 0.2% across the detector (Peter F. S., et al, 2005).

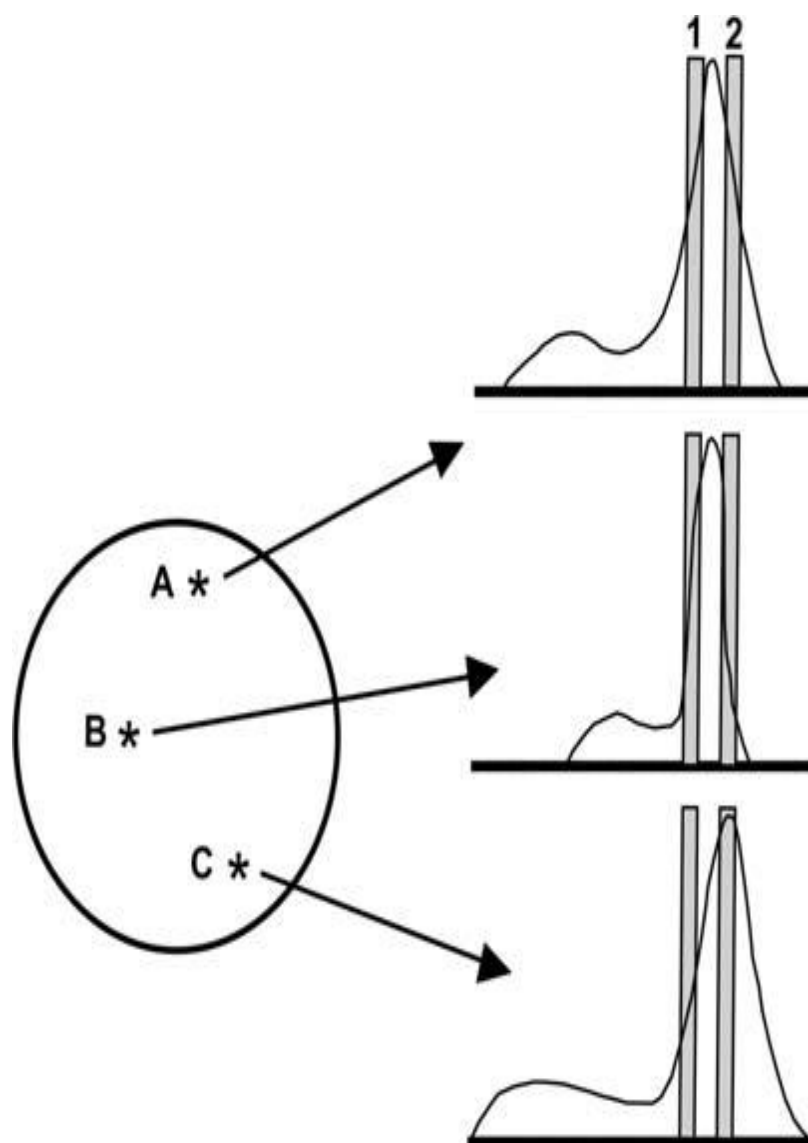


Figure 2.4 Variation of the energy signal with the position of the scintillation in the crystal. At A, the two windows are situated symmetrically around the photopeak. At B, the calculated values for the energy signals are lower than at A and so more events are accumulated in window1 than 2, while at C the energy signal is higher than at A and there are more events in window 2. Correction factors for the energy signals can be generated by comparing the number of gamma rays in the two windows.

2.7.3 Scatter Correction

In many clinical studies a significant proportion of the gamma rays in the image will be scattered. The effect is to reduce the contrast since they will, approximately, form a uniform background, and to defeat any attempts to quantify the amount of radioactivity in an organ. Since the amount of scatter will vary within the image depending upon the amount of tissue through which photons need to pass before reaching the detector, any correction will need to be made on a pixel-by-pixel basis. Hence, the use of the PHA alone is insufficient to remove the scatter (Peter F. S., et al, 2005).

2.7.4. Temporal Correction

The typical nuclear medicine study requires the camera to process only a few tens of thousands of counts per second. There are some studies, however, cardiac first pass being the main example, where the count rate can be very high. The probability that successive pulses passing through the camera's electronic circuits will overlap each other becomes unacceptably high. Such pulse pile-up results in incorrect X, Y, and Z signals being produced. Not only will this distort the image, but the recorded energy of these events will be higher than that of the individual pulses leading to them being rejected by the PHA; the detected count rate will no longer increase linearly, or indeed monotonically, with the amount of activity in the camera's field of view (Peter F. S., et al, 2005).

The effectiveness of correction circuits varies considerably between manufacturers. An image of a flood source often provides a simple test. The prospective purchaser of a camera should check that the correction facilities are usable for all operations with the camera and that the correction works for different radionuclides and for window settings of different energies (Peter F. S., et al, 2005).

2.8 Image Display

Modern camera systems employ digital display systems. If the Z signal corresponding to a particular detected gamma ray falls within the window that has been set on the PHA, then an enable signal is sent and the X and Y signals are recorded (Peter F. S., et al, 2005).

The most common form of image acquisition is called matrix or frame mode. The camera's field of view is divided into a regular matrix of picture elements or pixels. Each pixel is assigned a unique memory location in the computer. The value stored in this location is the number of gamma ray events that have been detected in the corresponding location on the camera face. The number, and hence the size, of the pixels used is of practical importance and depends on (i) the available computer memory (unlikely to be a problem with modern systems), (ii) the total number of images to be acquired, (iii) the number of counts contained in each image, and (iv) the required temporal and spatial resolution (Peter F. S., et al, 2005).

As modern systems can acquire and display static images with array sizes of up to 2048×2048 pixels, image quality is comparable with that of analogue images. Each pixel is typically stored as a 16 bit unsigned integer (allowing count values to range from 0 to 65 535); more bits per pixel would allow higher counts to be recorded, but at the expense of storage space, image transfer rate, and possibly processing time (Peter F. Sharp, 2005).

The image data are mapped from the array of numbers, representing the number of gamma rays acquired at a specific spatial location, into a viewable image by a look-up table which links the number of gamma rays to a specific value of displayed image intensity. The first question to be addressed is how best to

represent image intensity, both gray shades and colors having been used. The human eye can distinguish thousands of different colors, compared with only several tens of gray shades. The use of color to display intensity information, known as pseudo color, should therefore permit a much wider dynamic range to be reproduced. For instance, the so-called “hot body” scale, first proposed for coding ultrasound images, uses a gradual variation in hue and intensity to increase the dynamic range of the display; the scale changes from black, through shades of red, to white. Color scales can also make identification of specific ranges of pixel values easier, and are particularly useful for parametric (Peter F. S., et al, 2005).

However, there are several serious problems associated with color scales. Firstly, if the scale includes distinct perceptual color steps it can introduce false edges and contours in the image, greatly exaggerating small, and possibly insignificant, changes. Secondly, rather than enhancing the dynamic range, poorly chosen color scales can degrade it and reduce image contrast. Finally, there is the issue as to whether the display device (whether a screen or a hard-copy device) is actually capable of reproducing all the colors in the color scale. Failing to do so can result in large portions of the color scale being reproduced as the same color (Peter F. S., et al, 2005).

The second question is how to map count density to the chosen gray shades or colors. The simplest case is a linear relationship between the number of counts and the displayed intensity. Other transformations include compression of parts of the scale to increase contrast over a specific count range, and non-linear mappings, such as logarithmic, power law, square root or exponential. Histogram equalization is a non-linear intensity transform which has been applied to nuclear medicine images. Count density is mapped such that each display intensity, gray shade or color, occurs approximately (due to digitization) the same number of times in the

image. Although equalization theoretically maximizes image contrast, in practice it is rarely useful. A variation on this, adaptive histogram equalization, involves the application of histogram equalization in small sub-areas of the image, rather than globally to the whole image. However, it also tends to amplify image noise, an undesirable property for nuclear medicine images (Peter F. S., et al, 2005).

One of the simplest ways of manipulating the displayed data is by thresholding (sometimes known as windowing). For example, in bone imaging the radiopharmaceutical is excreted into the bladder, giving an area of very high count density. In the absence of thresholding most of the display levels will be used to display the bladder activity rather than the bones. By applying an upper threshold all areas having counts above a threshold value will be given the same maximum gray shade. This leaves the remaining gray shades or colors to be assigned to those counts found in the bones. Similarly a lower threshold may be applied such that all low count density areas are given the same low gray shade (Peter F. S., et al, 2005).

2.9 TYPES OF GAMMA CAMERAS AND THEIR CLINICAL USES

The most common type of gamma camera is the single-headed system. It consists of a gamma camera detector mounted on a gantry that allows the camera head to be positioned in a flexible way over different regions of the patient's body. Often, a moving bed is incorporated to permit imaging studies of the whole body. The gamma camera head often is mounted on a rotating gantry, allowing it to take multiple views around the patient. This feature also is necessary for producing *tomographic* images, or cross-sectional images through the body.



FIGURE (2.5) Single-headed gamma camera mounted on a rotating gantry. The camera is operated from the computer (left). The flexible positioning of the camera head and the bed (right) allows the system to obtain images of many different parts of the body.

Dual-headed gamma cameras are becoming increasingly popular. In these systems, two gamma camera heads are mounted onto the gantry as shown in Figure 2-3. Usually, the two heads can be positioned at a variety of locations on the circular gantry. An obvious advantage of a dual-headed camera is that two different views of the patient can be acquired at the same time. For example, in whole-body imaging, the two detector heads can be placed at 180 degrees to each other to provide anterior and posterior views simultaneously. Triple-headed systems also exist, primarily for tomographic studies.

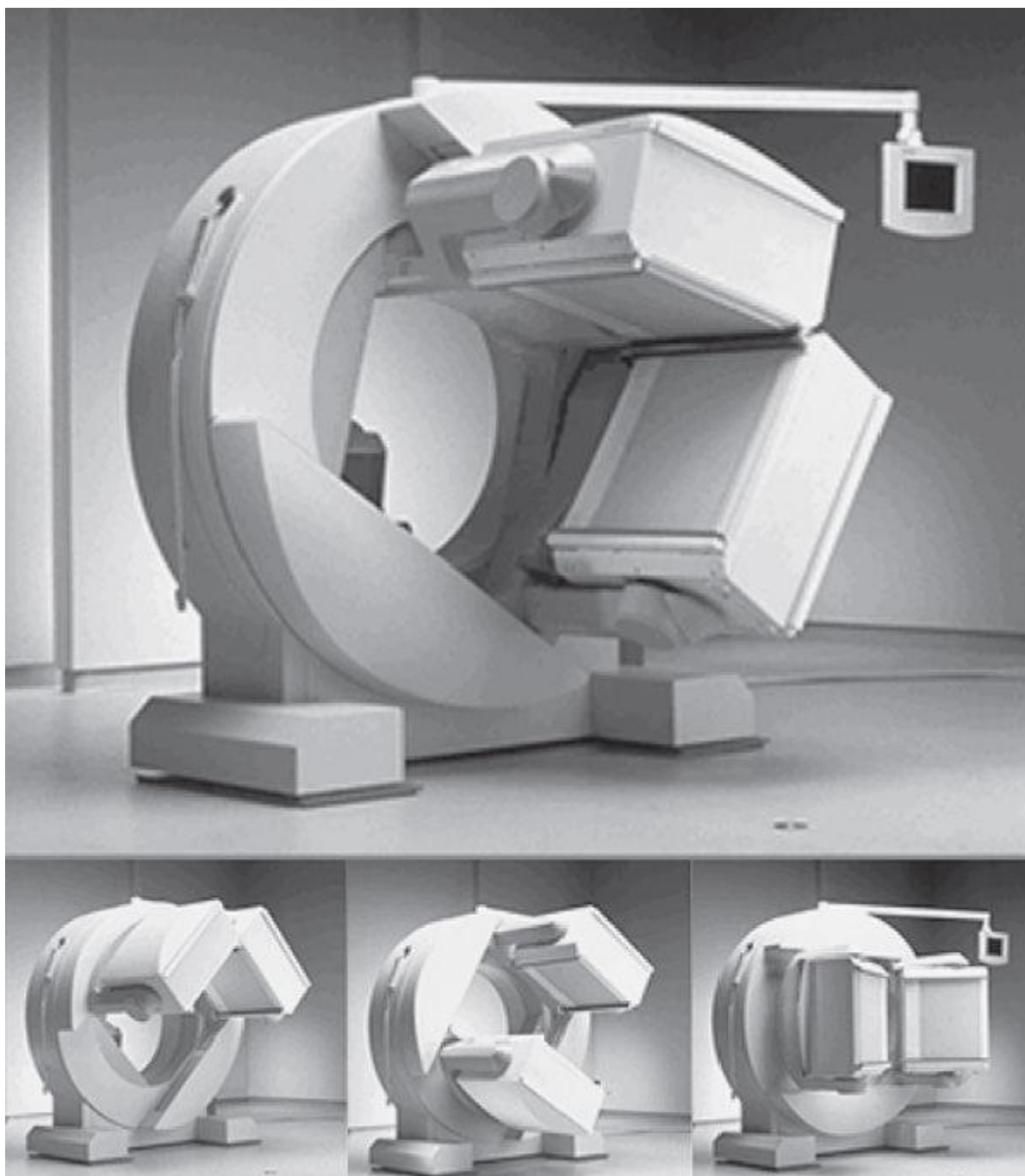


FIGURE (2.6) A dual-headed gamma camera system (top). Note that the camera heads can be placed in different orientations to provide two simultaneous views of an organ or the body (bottom).

2.10 Gamma camera performance and quality control

The performance of a gamma camera is characterized by six parameters included spatial resolution, uniformity, linearity, sensitivity, count rate characteristic and energy resolution. To understand gamma camera performance and routine QC, several terms like intrinsic versus extrinsic (or system) performance and useful versus the central field of view must be illustrated. Intrinsic performance refers to gamma camera performance without a collimator in place while extrinsic performance refers to gamma camera performance with collimator. The useful field of view (UFOV) of a gamma camera is essentially the entire detector (crystal) area while the central field of view (CFOV) corresponds to the inner, or central, $\frac{3}{4}$ of the crystal area, the CFOV is the portion of the detector actually used in clinical imaging [(Peter F. S., et al, 2005), (Hans, et al, 2007)].

2.10.1 Spatial Resolution

The spatial resolution is related to the smallest separation between two point sources which will permit them to be distinguished as two distinct sources and is measured by imaging a point or line source it is more usually, the line source is placed parallel to the collimator face and at a given distance from it. The measured quantities are the full width at half maximum (FWHM) is the width of the curve at half the peak count and full width at

tenth maximum (FWTM) is the width of the curve at one tenth of the maximum counts, of the resulting curve. The detector system itself has an intrinsic spatial resolution (R_i) and the collimator has a geometric resolution (R_c), the system resolution (R_s) is given by:

$$(R_s)^2 = (R_i)^2 + (R_c)^2$$

in practice performance by using resolution phantom (four-quadrant bar-phantom), this should be done at least weekly [(Peter F. S., et al, 2005), (Hans, et al, 2007)].

2.10.2 Sensitivity

Sensitivity regarded to the detection efficiency of the camera. It is defined as the counting rate obtained per unit activity in a standard source geometry. It is affected primarily by the choice of the collimator and deteriorates as resolution improves for a given camera (Peter F. S., et al, 2005).

2.10.3 Linearity

Linearity of the gamma camera image is tested by examining the image of the bar phantom obtained with a high resolution collimator in place. The lines in the image should be straight and unbroken, note that linearity here

refers to the appearance of the bars as lines. Integral linearity is assessed by calculating the variation in distances between adjacent pairs of line source images, while differential linearity measures the maximum in adjacent segments of any one line image [(Peter F . S., et al, 2005), (Hans, et al, 2007)].

2.10.4 Energy Resolution

Energy resolution it is defined as the ability of gamma camera to distinguish between photons of different energies. The ability of the analyzer to rejected scatter photons is dependent upon the width of the photopeak in the energy spectrum, usually expressed in term of the FWHM. The pulse height analyzer require adjustment to properly center the window of photon energies accepted, the procedure for checking the location of the energy window is to place a vial or syringe containing a small quantity of isotope without collimator and the computer displayed a plot of counts veers energy. The user can then adjust the location and width of the window, for example a standard setting for technetium-99m is a potopeak of 140 KeV and the window of 20% [(Peter F . S., et al, 2005), (Hans, et al, 2007)].

2.10.5 Count Rate Characteristic

A gamma camera has a pulse processing time associated with each event. The region is reached in which the response of the camera (observed count

rate) is no longer linear with increasing source radioactivity (true count rate). The theoretical linear response is obtained by extrapolation of the data points corresponding to count rates below 10000 counts per second, it being assumed that no data loss occurs below this rate (Peter F. S., et al, 2005).

2.10.6 Uniformity

Ideally, a scintillation camera should produce a uniform image of a uniform source, this ideal is not met due to imperfections in the collimators, variations in crystal response, difference among PMT response and minor fluctuations in the electrical circuitry.

The uniformity of the camera's response can be checked by imaging a flood source. A solid plastic disk manufactured with 5 to 20 mCi of ^{57}Co uniformly distributed throughout its extend or a fluid filled sheet source containing a dilute solution of radioactivity is placed directly on the collimator (extrinsic flood field), and other used a point source suspended several feet at a distance three to four times the diameter of the crystal directly above the surface of the crystal (without collimator or intrinsic flood field), 5 to 30 million count image of the flood is then collected according to the manufacturer's directions [Rachel A. P., et al, 2006].

It is usual to carry out the calculations for both the useful field of view (UFOV) it is whole usable field of view and the central field of view (CFOV) is defined as an area centered on the UFOV, having linear dimensions of the UFOV scaled by factor of 0.75 (Peter F . S., et al, 2005).

Integral uniformity is defined as the largest variation (maximum-minimum) in counts over the useful field of view, while differential uniformity is a measurement of the worst case rate of change of uniformity over a limited distance (~ 5 pixels). Modern gamma camera systems typically have integral and differential uniformities of between 4-7% (Michael K. O'Connor).

2.10.7 Center of Rotation

It is assumed that the camera heads will rotate in a near perfect circle and that heads will remain almost precisely aligned in their opposing positions. It is also assumed that the predicted or electronic center of the path of rotation will match the mechanical or actual center of the camera head rotation. Deviation from either expectation will degrade image resolution and can be seen as a displacement of the center of rotation (COR). Probably the most common cause of apparent displacement of COR is a result of errors not leveling the camera head or bumping the table during data collection.

The most common cause of true shift of the COR is electronic malfunction [Rachel A. P., et al, 2006].

2.11 National Electrical Manufacturers Association (NEMA)

The National Electrical Manufacturers Association is the trade body of gamma camera manufacturers in the USA. They have define arrange of standard tests for gamma camera performance in their publication (performance measurements of scintillation cameras-NEMA,2001). The purpose of that publication is to define a common set of criteria for measuring camera performance that can be used by all manufactures defines two types of standard; performance standards and class standards [NEMA, 2001].

Table (2.1): Quality control procedure according to IAEA [International Atomic Energy Agency, 1991].

Test	Acceptance	Daily	Weekly	Half a year	Monthly	Quarterly
Energy spectrum	✓	✓				
Intrinsic uniformity	✓	✓				
Extrinsic uniformity	✓	✓		✓		
Intrinsic energy resolution	✓					✓
Extrinsic energy resolution	✓					
Intrinsic spatial resolution	✓		✓			
Sensitivity	✓			✓		
Center of rotation	✓		✓			

2.12 Image Quality

Image quality is a general concept that applies to all types of images. It applies to medical images, photography, television images, and satellite reconnaissance images. "Quality" is a subjective notion and is dependent on the function of the image. In radiology, the outcome measure of the quality of a radiologic image is its usefulness in determining an accurate diagnosis. The true test of an imaging system, and of the radiologist that uses it, is the reliable detection and accurate depiction of subtle abnormalities. As visual creatures, humans are quite skillful at visual cognition and description. Most people can look at an image and determine if it is "grainy" or "out of focus." One of the primary aims of this chapter is to introduce the reader to the terminology used to describe image quality, and to the parameters that are used to measure it. The principal components of image quality are *contrast*, *spatial resolution*, and *noise* (Jerrold T. B., et al, 2002).

2.12.1 Contrast

The contrast is mainly determined by the characteristics of the tracer and the amount of scatter. The specificity of a tracer for a particular metabolic process is usually not 100%. For example, for most tracers the blood concentration decreases rapidly but is typically not zero during the study. Consequently, the blood concentration produces a "background" tracer uptake, which decreases the contrast. Scattered photons also produce a background radiation that reduces the contrast (Paul S., 2009).

2.12.2 Spatial Resolution

The spatial resolution is related to the smallest separation between two point sources which will permit them to be distinguished as two distinct sources and is measured by imaging a point or, more usually, line source. The line spread function, $L(x)$, is the graph of count rate as a function of the x-coordinate when the line source is placed parallel to the y-axis in a plane parallel to the collimator face and at a given distance from it. The measured quantities are the full width at half maximum (FWHM) and full width at tenth maximum (FWTM) of the resulting curve (Figure). The detector system itself has an intrinsic spatial resolution (R_i), due to the electronics, and the collimator has a geometric resolution (R_c). The system (overall) resolution (R_s) is given, to a good approximation, by

$$(R_s)^2 = (R_i)^2 + (R_c)^2.$$

In practice performance by using resolution phantom (four-quadrant bar-phantom), this should be done at least weekly [(Peter F. S., et al, 2005), (Hans, et al, 2007)].

2.12.3 Noise

Poisson noise contributes significantly to the measurements. The ML-EM reconstruction algorithm takes this into account and inherently limits the influence of this noise by terminating the procedure after a few tens of iterations (Paul S., 2009).

Another noise factor is due to Compton scatter. It produces a secondary photon that is deflected from the original trajectory into a new direction. Some of the scattered photons reach the detector via this broken line. Such a contribution to the measurement is undesired, and a good camera suppresses this as much as possible.

The system can reject scattered photons based on their energy. As compared to primary photons, the scattered photons have a lower energy, which is measured by the detector electronics. However, the energy resolution is finite (10% for Na(Tl)), and some of the scatter is unavoidably accepted. The remaining scatter has a negative effect on the image contrast and the accuracy of quantitative measurements (Paul S., 2009).

2.13 Previous studies

Uniformity image analysis with traditional pixel value-based methods is not reliable because the tendency to miss subtle nonuniformities that may be visually apparent and clinically significant. This results in the need for additional meticulous and subjective visual assessment. A metric that we derived from the 2D NPS—the SNI—outperformed currently established pixel value-based methods in both identifying nonuniformities and correlating with expert visual analysis, thereby possibly reducing the need for additional visual assessment. This metric provided an accurate and objective uniformity evaluation and may be especially useful if integrated into daily quality control programs in which qualified medical physicists are unable to verify g-camera flood quality before performing patient studies (Jeffrey S. N., et al, 2014).

CHAPTER THREE

Material and method

3.1 Materials:

The images have been obtained from a gamma camera (SPECT) manufactured by Mediso (Hungary), with the serial number (DH-506069-V0) installed in the Radiation and Isotopes Center of Khartoum (RICK) Department of Nuclear Medicine. The image of the machine is shown in figure (3.1).



Figure (3.1): Gamma camera machine (SPECT) in hospital (RICK).

3.2 Methods:

3.2.1 Area of study

The images have been obtained from the nuclear medicine department in the Radiation and Isotopes Center of Khartoum (RICK). This study performed in period between June 2014 and February 2015.

3.2.2 Method of data collection

Intrinsic uniformity

It's the camera ability to detect a uniform source of the radioactivity distribution.

Procedure:

- Remove the collimator, use matrix size (256*256*16).
- Make the two head of collimator perpendicular to each other for the dual head.
- From main menu select patient study, uniformity test and press F3, the computer will calculate the integral and differential uniformity from the flood image.

Extrinsic uniformity

Procedure:

- Place the Co⁵⁷ flood source directly on the collimated camera.
- Make the two head of collimator perpendicular to each other for dual head.

- Use matrix size (256*256*16).
- From main menu select patient study, uniformity test and press F3, the computer will calculate the integral and differential uniformity from the flood image.

3.2.3 Data analysis method

The results in this study have been obtained using computerized method in which a home-built Matlab program has been used.

The images offered as a TIF format then using Matlab the images were accessed. After that, the region of interest were determined using crop function, then it was smoothed by 7*7 window to reduce image noise (to increase signal to noise ratio).

Then integral uniformity was calculated by finding the maximum and the minimum count, as follow:

$$\text{Integral uniformity (\%)} = (C_{\max} - C_{\min}) / (C_{\max} + C_{\min}) \times 100$$

For the differential uniformity the image were scanned by using two loops to find the different uniformity in rows and columns, where the different uniformity were calculated for every 5 pixel in case of columns and rows, were calculated using the difference in maximum and minimum count pixel in each, using the same equation above.

The regions of interest were displayed in a 3d for better visualization of the uniformity as an objective method rather than the subjective method (the visual perception).

CHAPTER FOUR

RESULTS

The results in this chapter will be presented by using tables and figures

Table (4 – 1): a table that shows the analysis result of (IU%, DU_{xy}%, Maximum intensity and the minimum intensity) of the flood uniformity images (image0 to image11) from the intrinsic test.

#	IU%	DU _x %	DU _y %	Max. Intensity	Min. Intensity
Image 0	1.86	0.60	0.60	5.76E+04	5.55E+04
Image 1	2.07	0.77	0.77	5.77E+04	5.54E+04
Image 2	51.92	13.06	13.06	5.87E+04	1.86E+04
Image 3	41.97	9.18	9.18	6.02E+04	2.46E+04
Image 4	48.81	13.95	13.95	6.07E+04	2.09E+04
Image 5	1.45	0.73	0.73	5.73E+04	5.56E+04
Image 6	1.73	0.68	0.68	5.78E+04	5.58E+04
Image 7	1.71	0.75	0.75	5.74E+04	5.55E+04
Image 8	1.75	0.60	0.60	5.76E+04	5.57E+04
Image 9	1.74	0.59	0.59	5.75E+04	5.55E+04
Image 10	1.48	0.57	0.57	5.75E+04	5.59E+04
Image 11	1.57	0.69	0.69	5.75E+04	5.58E+04

Table (4 – 2): a table that shows the analysis result of (IU%, DUxy%, Maximum intensity and the minimum intensity) of the flood uniformity images (image12 to image15) from the extrinsic test.

#	IU%	DUx%	DUY%	Max. Intensity	Min. Intensity
Image 12	5.40	1.60	1.60	5.20E+04	4.67E+04
Image 13	5.92	1.38	1.38	5.37E+04	4.77E+04
Image 14	5.40	1.60	1.60	5.20E+04	4.67E+04
Image 15	5.92	1.38	1.38	5.37E+04	4.77E+04

Figure (4-1): a plot that shows the integral uniformity as a function of the maximum intensity outlined from the intrinsic test result.

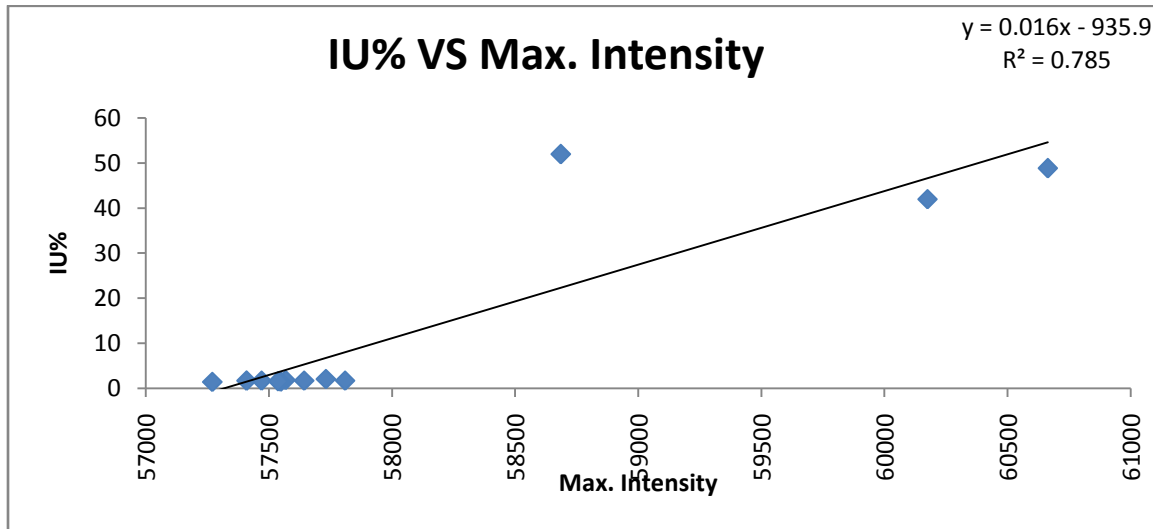


Figure (4-2): a plot that shows the integral uniformity as a function of the minimum intensity outlined from the intrinsic test result.

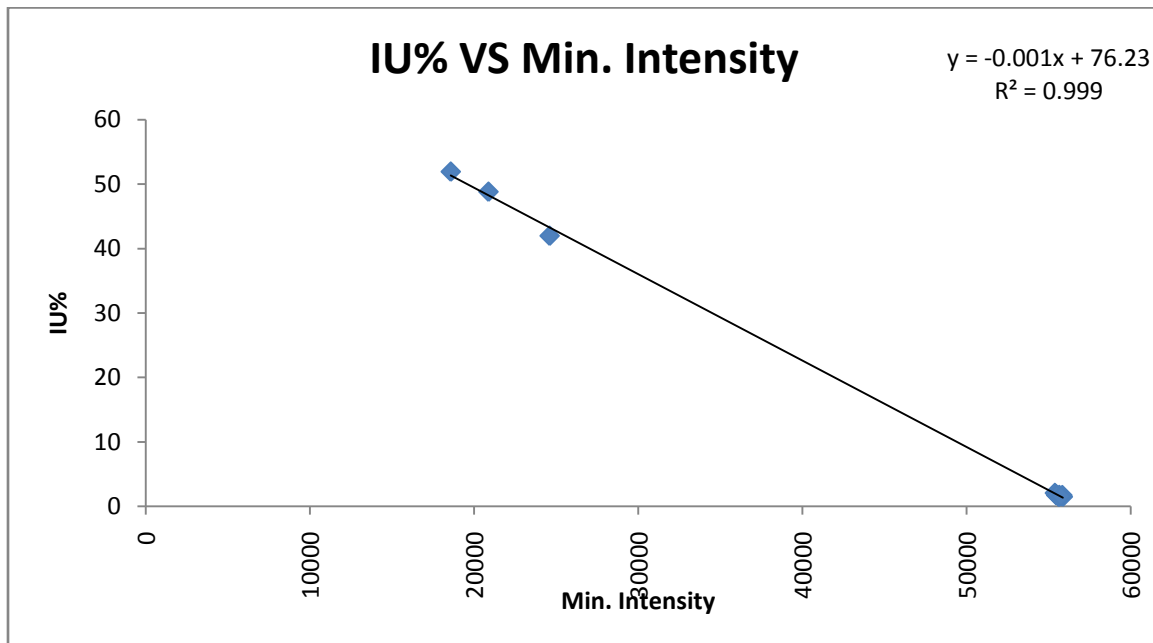


Figure (4-3): a plot that shows the differential uniformity as a function of the maximum intensity outlined from the intrinsic test result.

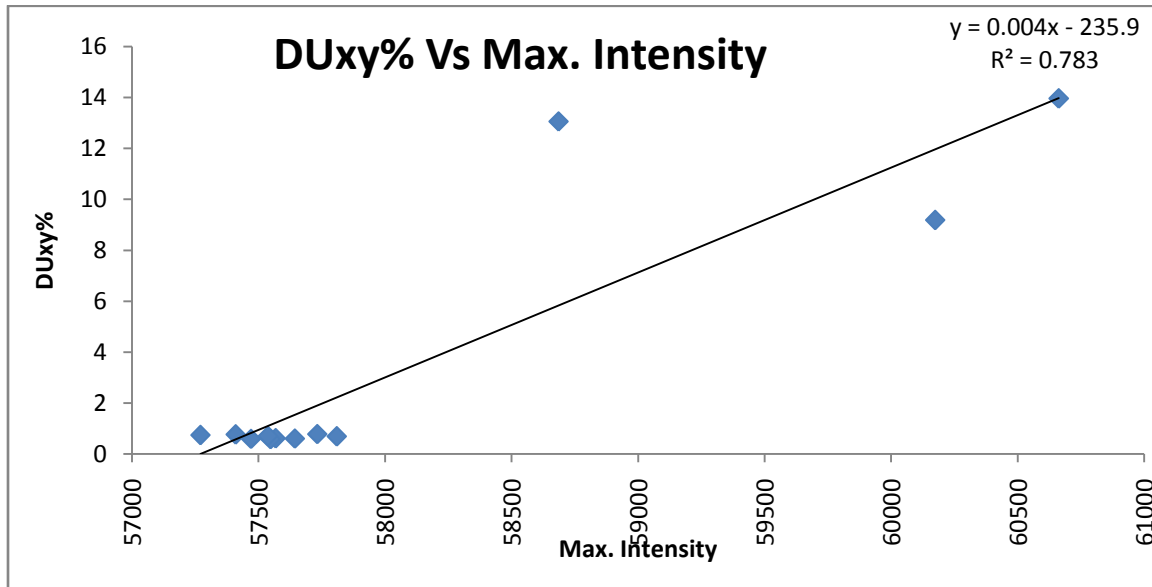


Figure (4-4): a plot that shows the differential uniformity as a function of the minimum intensity outlined from the intrinsic test result.

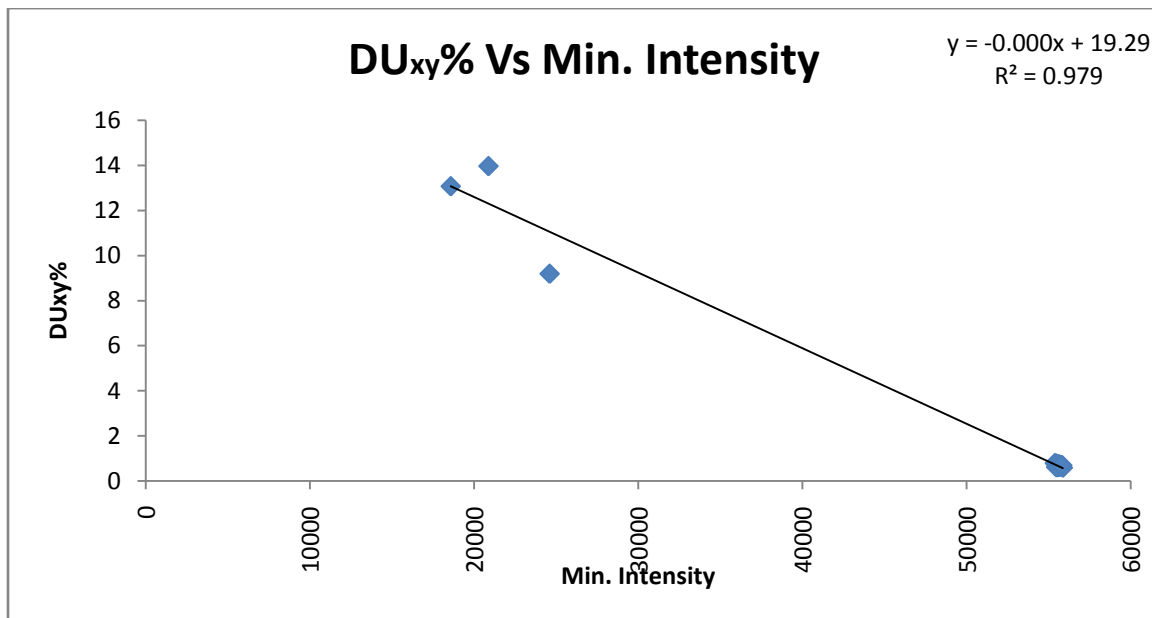


Figure (4-5): a plot that shows the integral uniformity as a function of the maximum intensity outlined from the extrinsic test result.

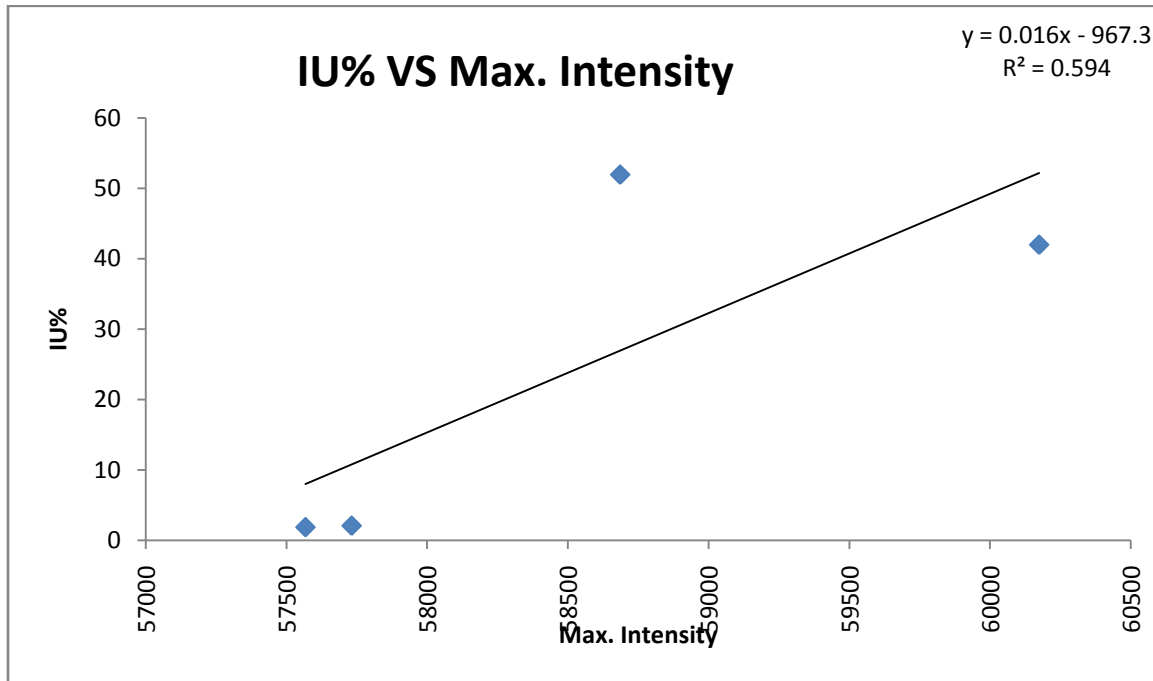


Figure (4-6): a plot that shows the integral uniformity as a function of the minimum intensity outlined from the extrinsic test result.

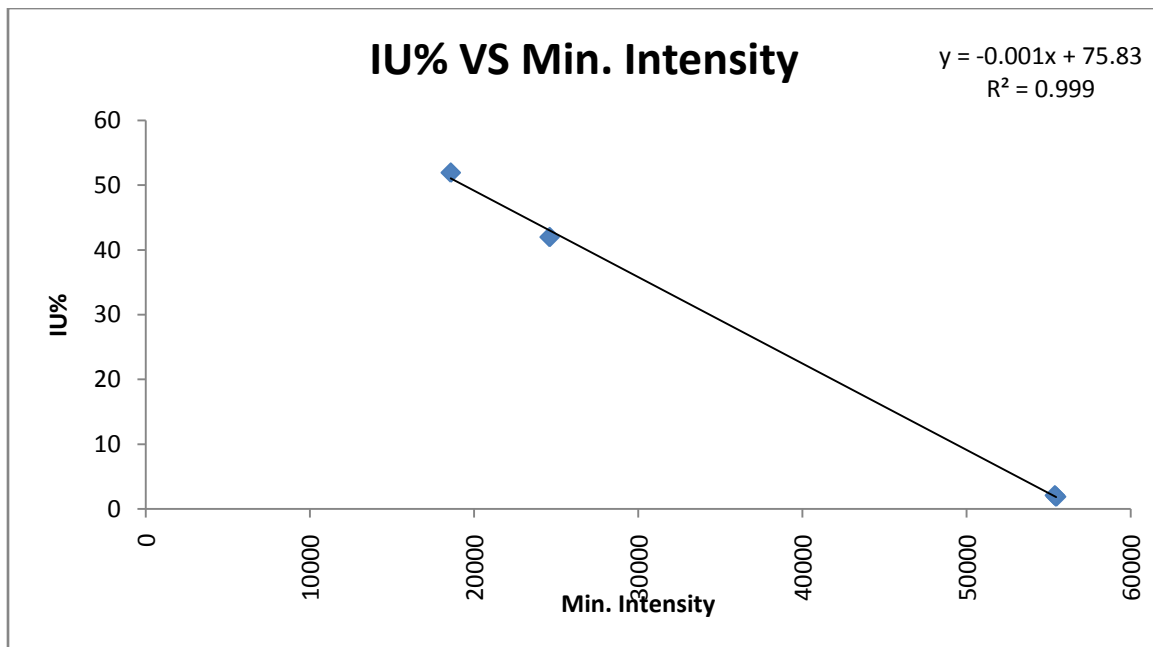


Figure (4-7): a plot that shows the differential uniformity as a function of the maximum intensity outlined from the extrinsic test result.

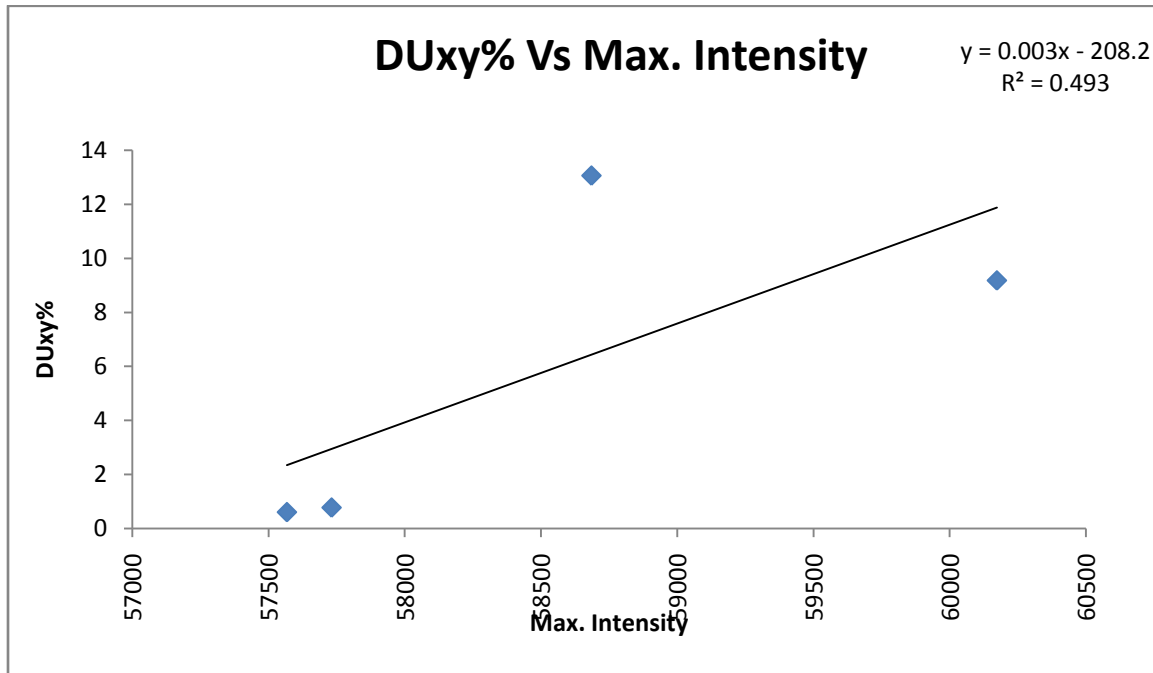


Figure (4-8): a plot that shows the differential uniformity as a function of the minimum intensity outlined from the extrinsic test result.

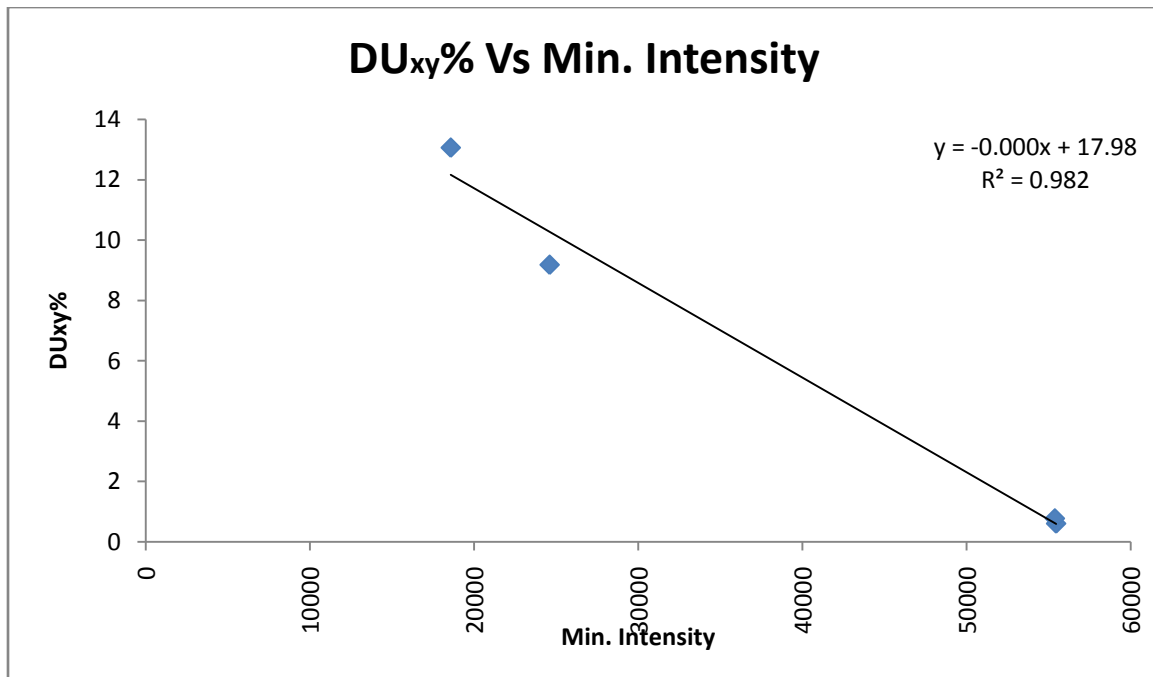


Figure (4 -9) that show image1. The integral uniformity had been calculated (2.07) and the differential uniformity (0.77) and we found these values are acceptable.

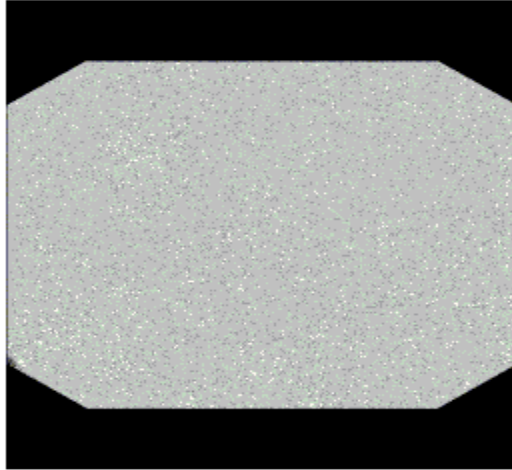


Figure (4-10) that show image3. The integral uniformity had been calculated (51.92) and the differential uniformity (13.06) and which have been found not acceptable.

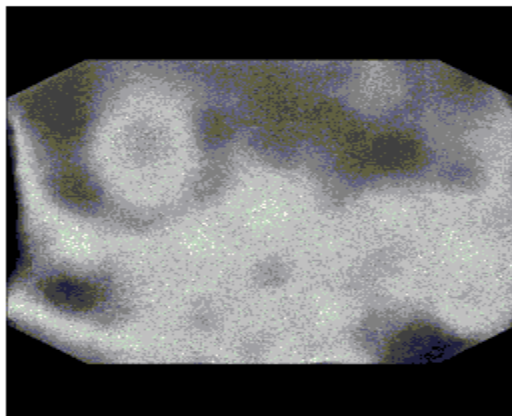


Figure (4-11): the figure depicts image1 in shade-surface illustration for better visualization.

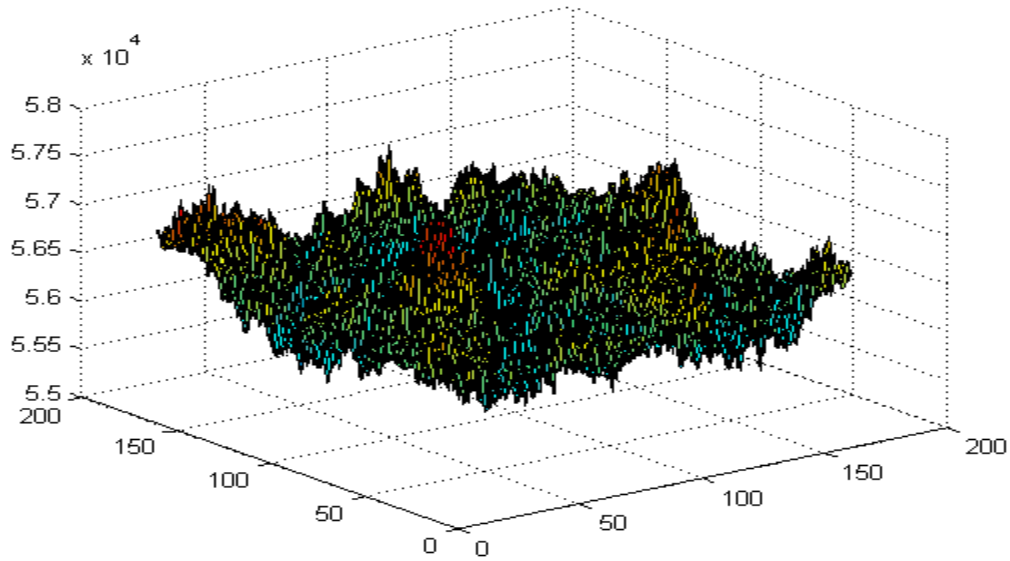


Figure (4-12): the figure depicts image3 in shade-surface illustration for better visualization.

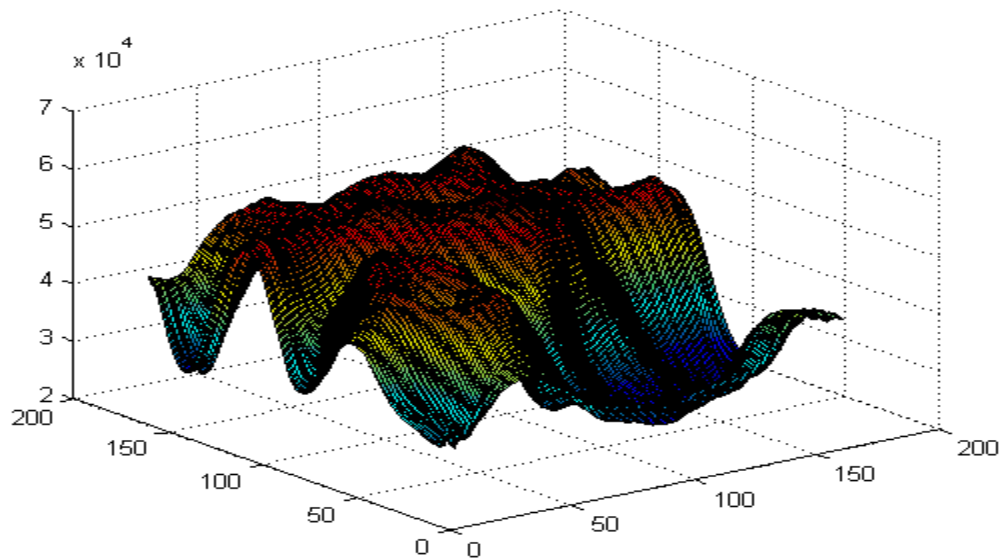


Figure (4-13): the figure represents the correction map in the x and y axis of image3 in shade-surface illustration.

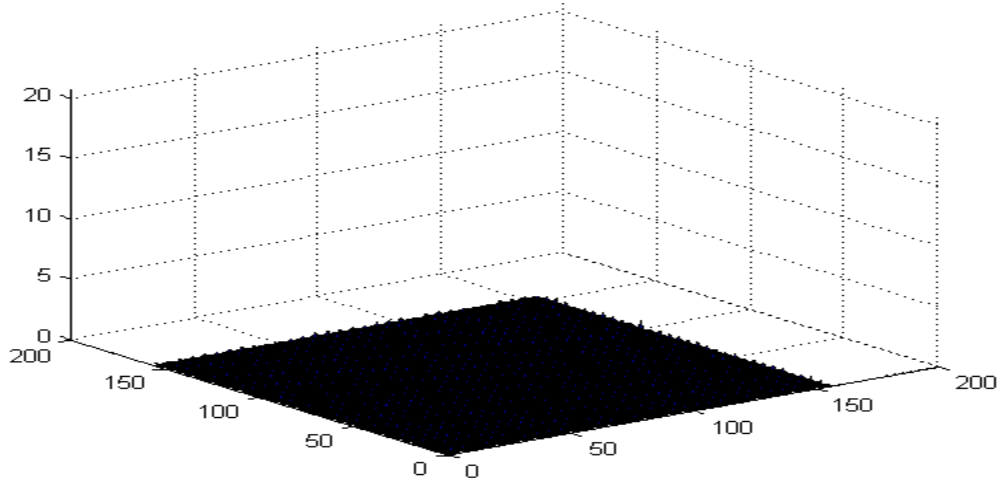
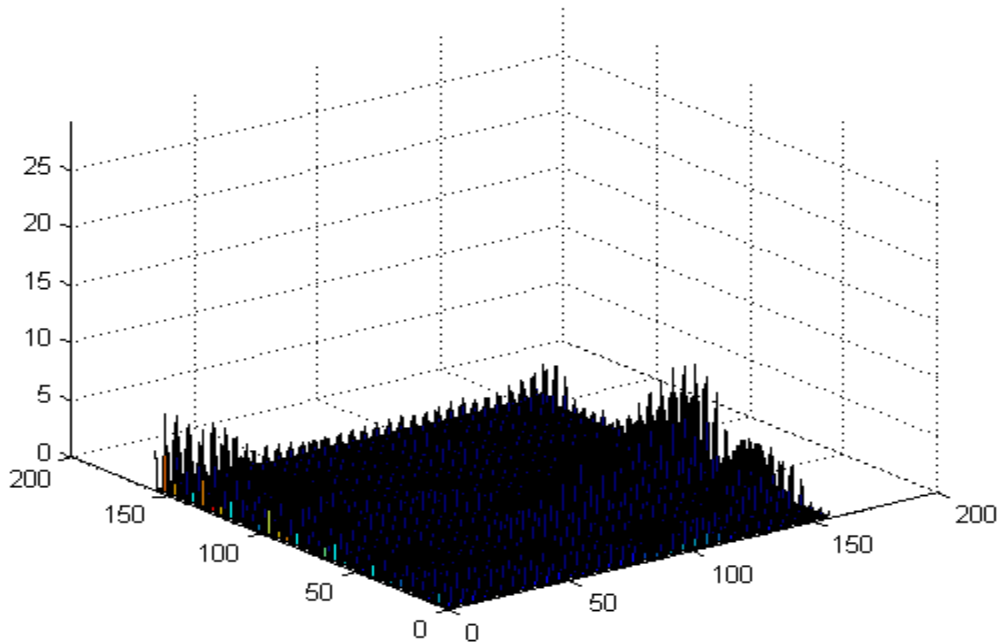


Figure (4-14): the figure represents the correction map in the x and y axis of image3 in shade-surface illustration.



Chapter Five

Discussion, Conclusion and Recommendations

5.1 Discussion

The images were obtained from the Radiation and Isotopes center of Khartoum (RISK) department of nuclear medicine diagnostic nuclear medicine in Sudan. The study includes 16 flood field uniformity test images of the intrinsic and extrinsic test. The data were collected in two phases: firstly, the images (in TIF) format was obtained from the archive. Secondly, the images were analyzed using Matlab program. The research came out with the results shown in tables and the figures tailed as appendix.

Table (4-1) shows the assessment result of Integral Uniformity, Differential Uniformity, Maximum intensity and the Minimum intensity of the flood images (image 0 to image11) from the intrinsic test. And it has been found that there is strong and significant correlation between the maximum intensity and the integral uniformity ($R^2=0.785$). The impact of the maximum intensity on the integral uniformity is increasing its value according to the equation ($y=0.016x - 935.9$).

Table (4-2) shows the assessment result of Integral Uniformity, Differential Uniformity, Maximum intensity and the Minimum intensity of the flood images (image 12 to image15) from the extrinsic test. And it has been found that there is strong and significant correlation between the minimum intensity and the integral uniformity ($R^2=0.999$). The impact of the minimum intensity on the integral uniformity is decreasing its value according to the equation ($y = -0.001x + 76.23$).

For the intrinsic test, in the figure (4-3), it has been found that there is strong and significant correlation between the maximum intensity and the differential

uniformity ($R^2=0.783$). The impact of the maximum intensity on the differential uniformity is increasing its value according to the equation ($y = 0.004x - 235.9$). Also, in the figure (4-4), it has been that there is strong and significant correlation between the minimum intensity and the differential uniformity ($R^2=0.979$). The impact of the minimum intensity on the differential uniformity is decreasing its value according to the equation ($y = -0.0001x + 19.29$).

For the extrinsic test, in the figure (4-5), it has been found that there is strong and significant correlation between the maximum intensity and the integral uniformity ($R^2= 0.594$). The impact of the maximum intensity on the integral uniformity is increasing its value according to the equation ($y = 0.016x - 967.3$). Also, in the figure (4-6), we found that there is strong and significant correlation between the minimum intensity and the integral uniformity ($R^2=0.999$). The impact of the minimum intensity on the integral uniformity is decreasing its value according to the equation ($y = -0.001x + 75.83$).

For the extrinsic test, in the figure (4-7), it has been found that there is strong and significant correlation between the maximum intensity and the differential uniformity ($R^2=0.493$). The impact of the maximum intensity on the differential uniformity is increasing its value according to the equation ($y = 0.003x - 208.2$).

For the extrinsic test, in the figure (4-8), it has been found that there is strong and significant correlation between the minimum intensity and the differential uniformity ($R^2= 0.982$). The impact of the minimum intensity on the differential uniformity is decreasing its value according to the equation ($y = -0.000x + 17.98$).

The intrinsic uniformity test results were accepted unless for some of the images that illustrates score abnormal values. For example, comparing between figures (4.9) and (4-10), figure (4-10) shows much abnormality (blurring).

For the better visualization inspection, the images have been described in a 3d surface, as shown in figures (4-11) and (4-12).

Figure (4-13): the figure represents the correction map in the x and y axis of image3 in shade-surface illustration. Figure (4-14): the figure represents the correction map in the x and y axis of image3 in shade-surface illustration.

5.2 Conclusion

The general objective of this Study was to study and apply a new quantitative uniformity analysis technique to the gamma camera uniformity flood images using an image processing technique.

The data of this study were collected from of gamma camera using 16 images represent intrinsic and extrinsic uniformity test. The author used his own algorithm to calculate the uniformity parameters.

In general, the result of this study showed that there is a strong correlation between the intensity (maximum & minimum) of the image and the uniformity parameters (integral uniformity & differential uniformity).

In conclusion, we can have our own program to quantify uniformity and hence we can have our own data base.

5.3 Recommendations

- Planar uniformity measurement is the most important QC test. It is performed daily and is designed to verify camera spatial uniformity and measure deviations from perfectly uniform count distributions.
- Uniformity testing help in identifying quite a lot of common performance issues, including spatial nonlinearities in camera response, drifting gains in photomultiplier tubes, corrupt correction files, and incorrectly positioned photo-peaks. Hence flood-field uniformity plays a vital role in determining the quality or clinical readiness of a system, how to effectively quantify this characteristic remains a major challenge and need more work in the future.
- The program that had been developed to be used in the research did not reach the NEMA standards software, thus, it may need more developing and enhancing to fully reach the specification needed.

5.4 References

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1011- 4289, printed by the IAEA in Austria May 1991

5.4 Appendices:

Images:

Figure (5.4.1): original image number 0

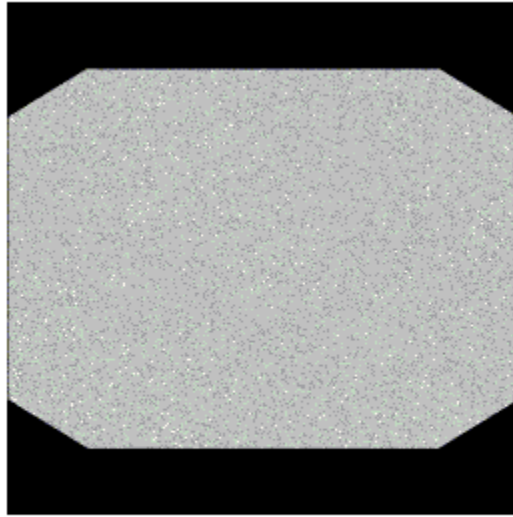


Figure (5.4.2): original image number 1

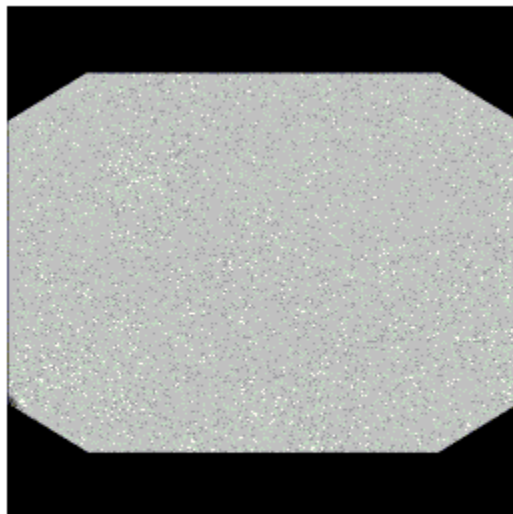


Figure (5.4.3): original image number 2

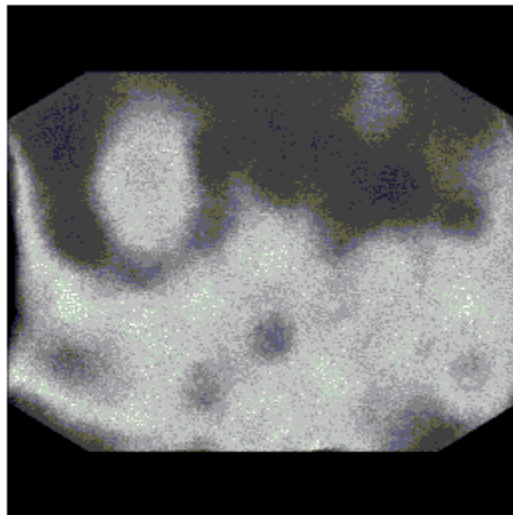


Figure (54.4): original image number 3

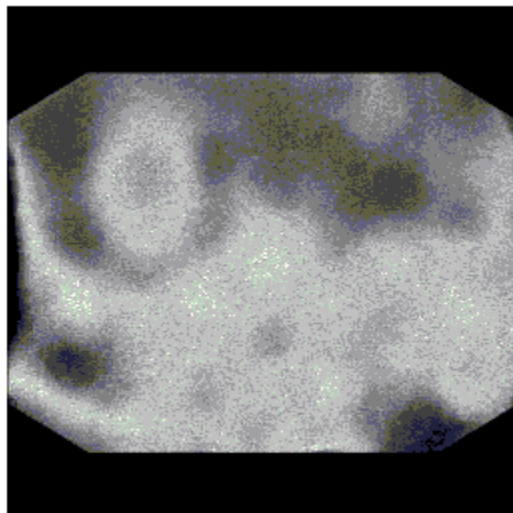


Figure (5.4.5): original image number 4

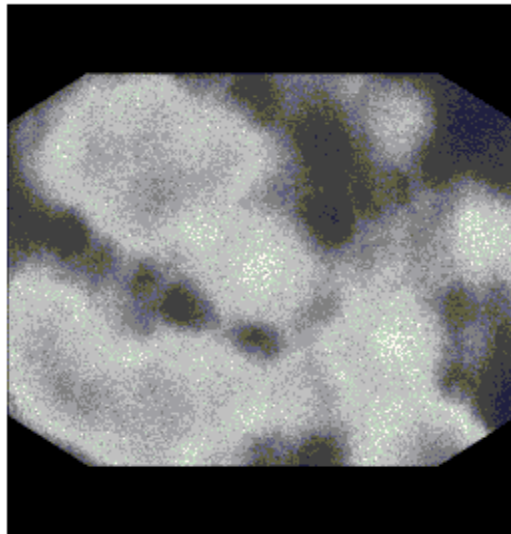


Figure (5.4.6): original image number 5

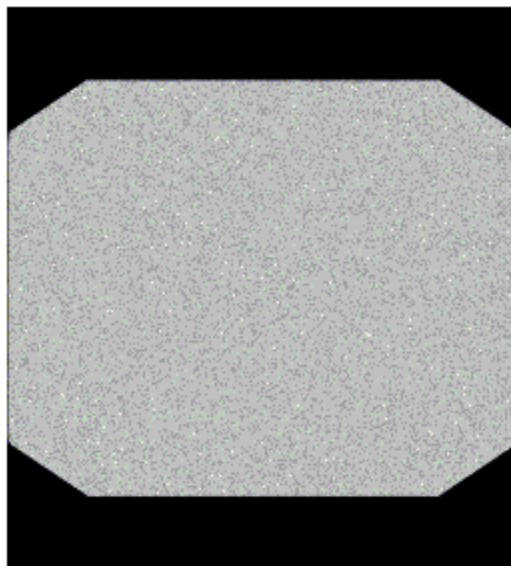


Figure (5.4.7): original image number 6

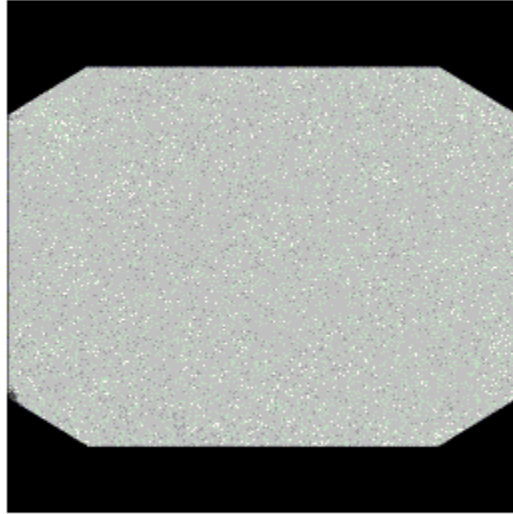


Figure (5.4.8): original image number 7

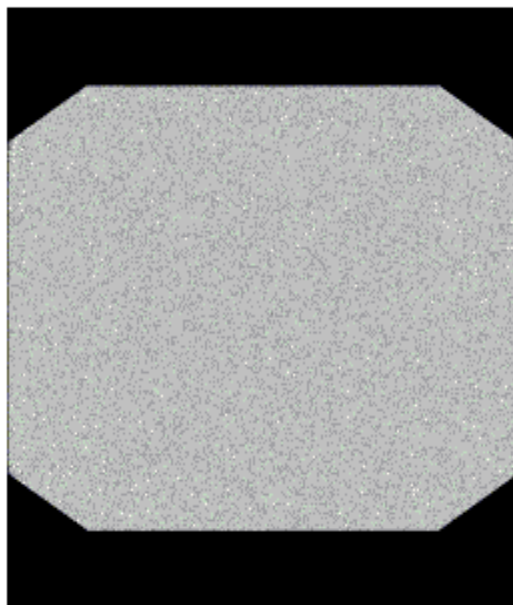


Figure (5 .4. 9): original image number 8

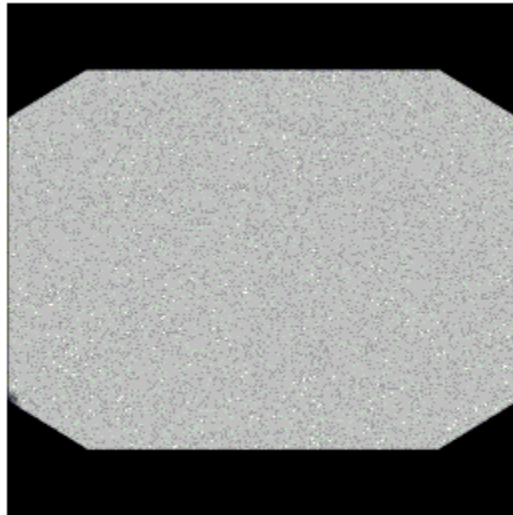


Figure (5.4.10): original image number 9

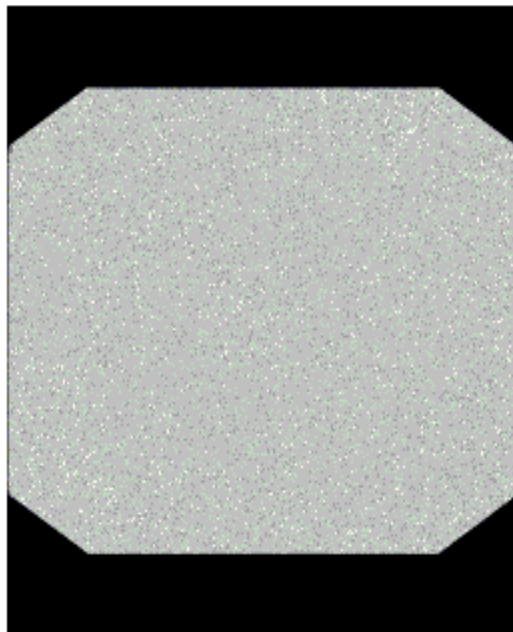


Figure (5.4.11): original image number 10

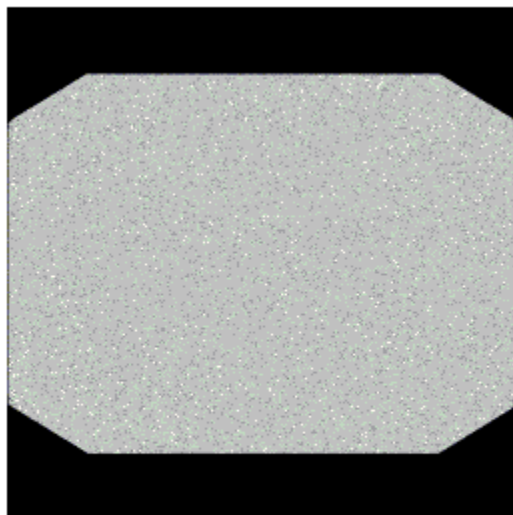


Figure (5.4.12): original image number 11

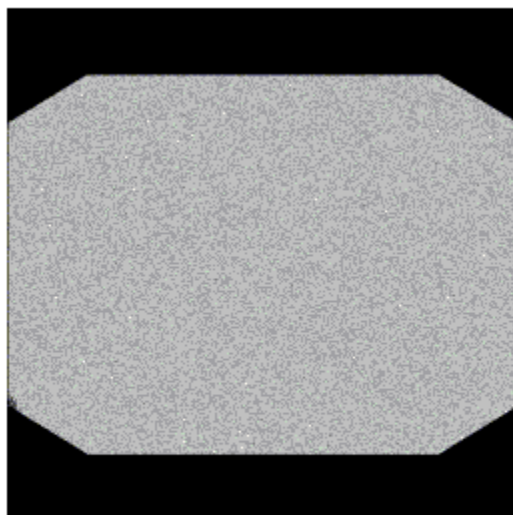


Figure (5.4.13): original image number 12

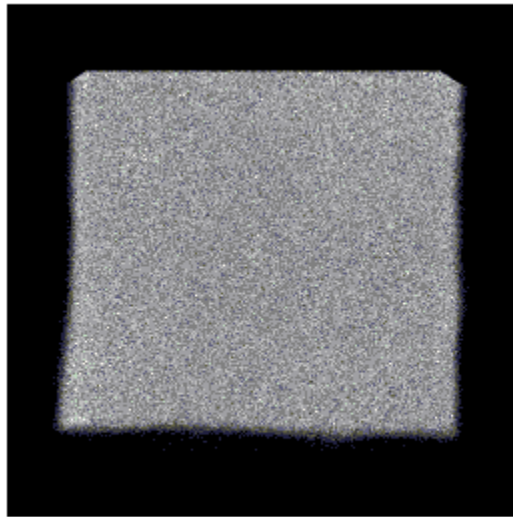


Figure (5.4.14): original image number 13

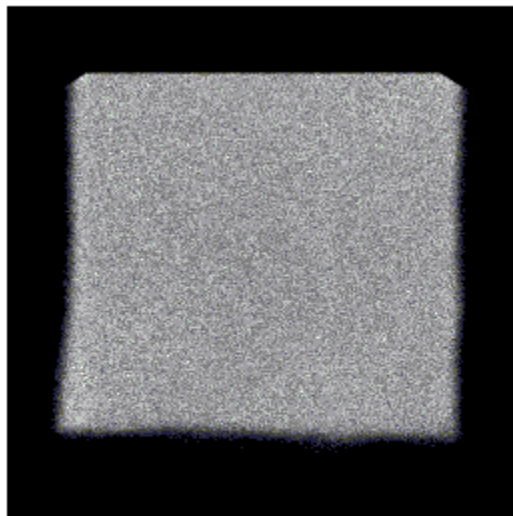


Figure (5.4.15): original image number 14

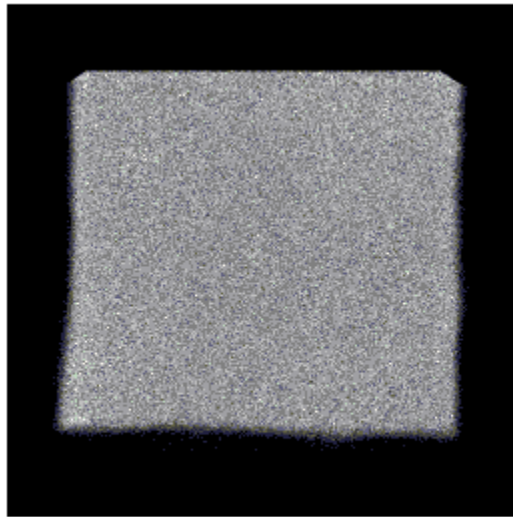


Figure (5.4.16): original image number 15

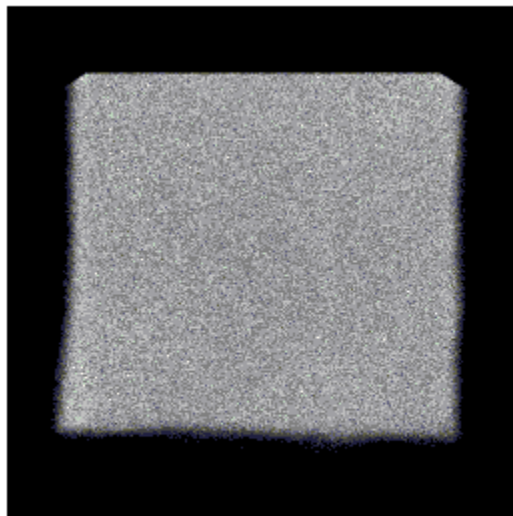


Figure (5.4.17): shad-surface plot for image number 0

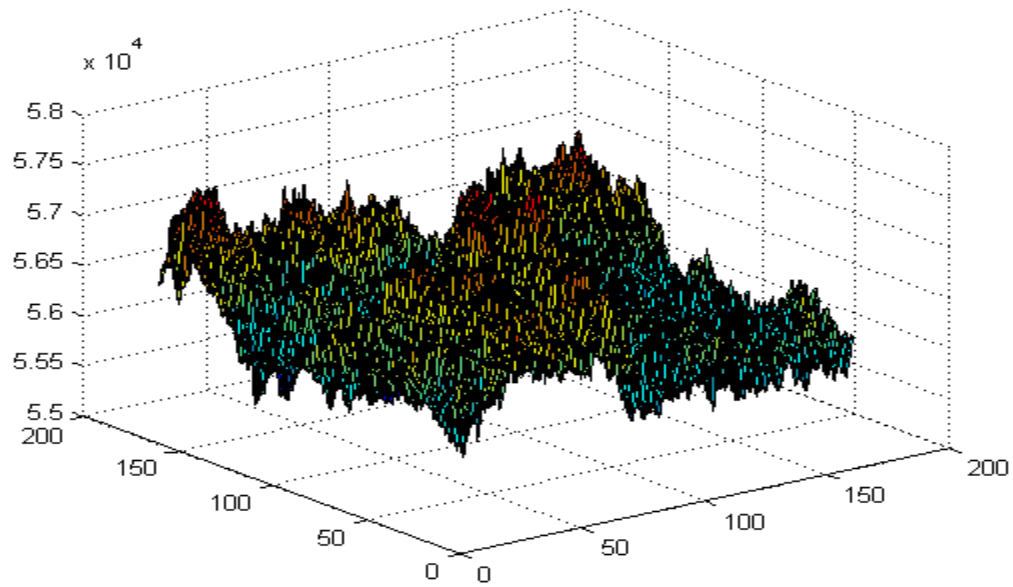


Figure (5.4.18): shad-surface plot for image number 1

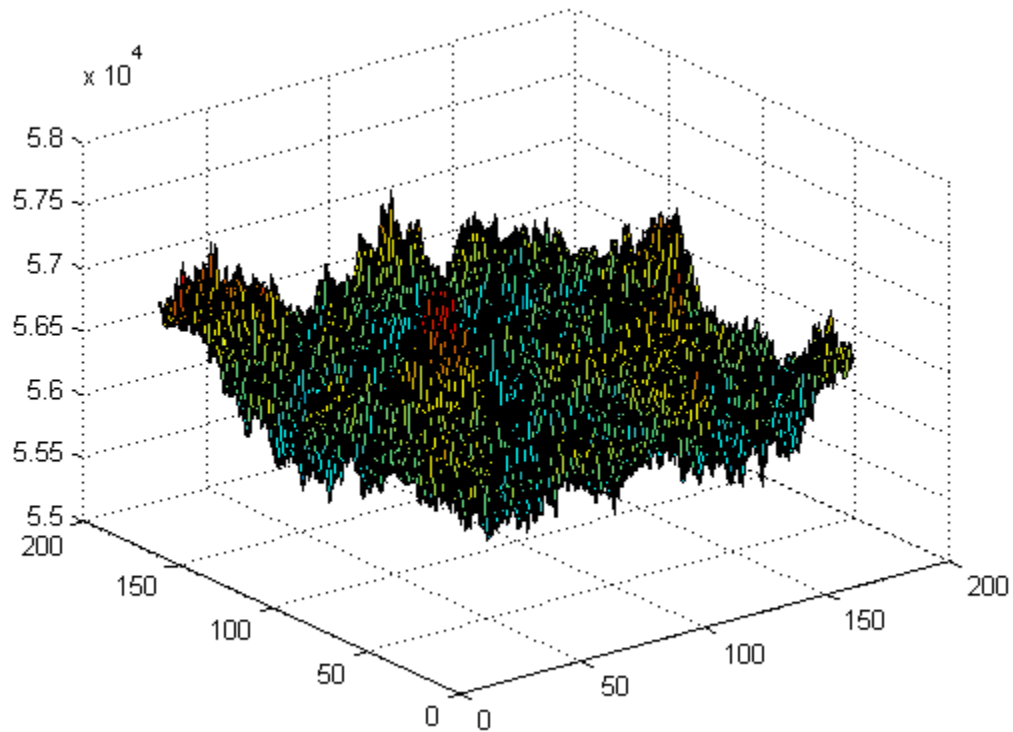


Figure (5.4.19): shad-surface plot for image number 2

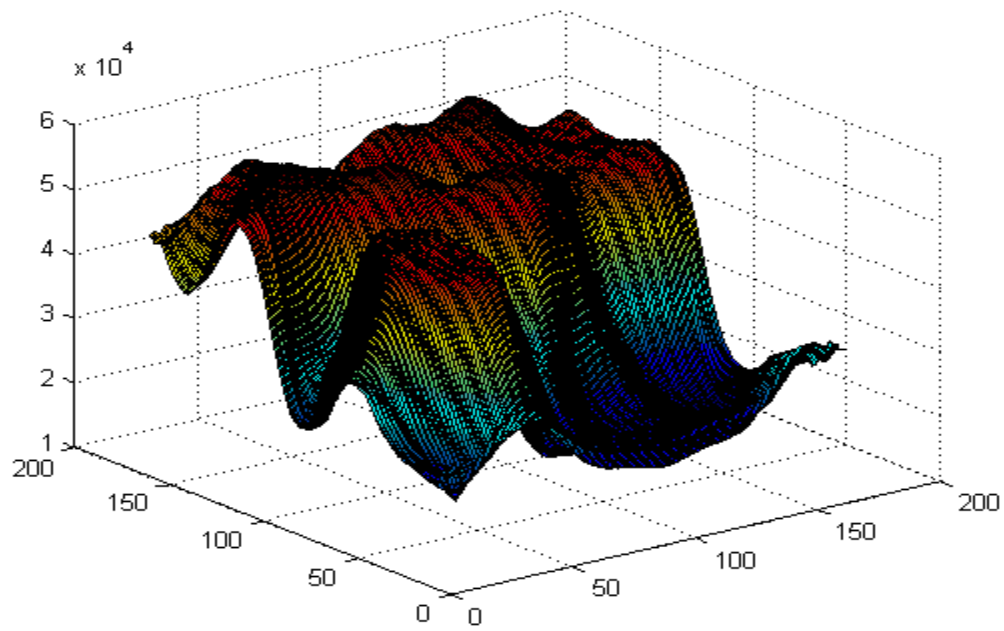


Figure (5,4.20): shad-surface plot for image number 3

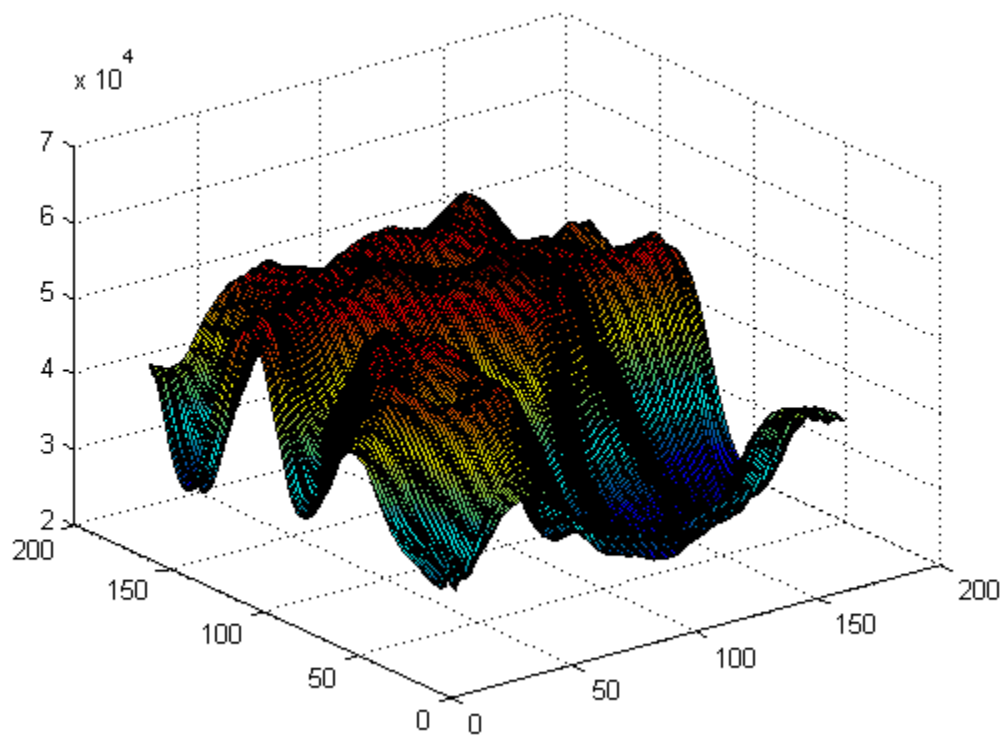


Figure (5.4.21): shad-surface plot for image number 4

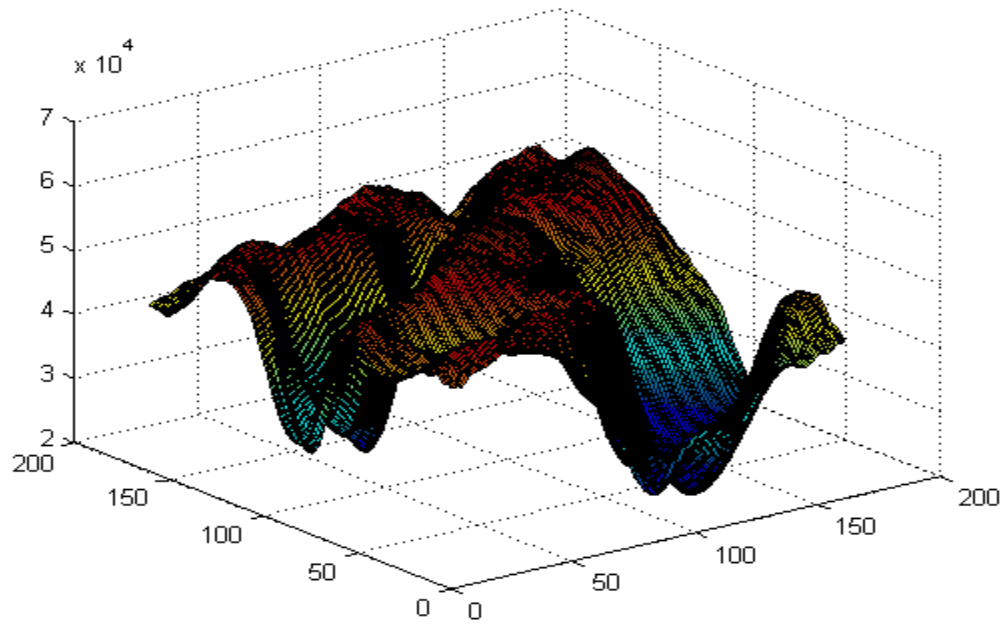


Figure (5.4.22): shad-surface plot for image number 5

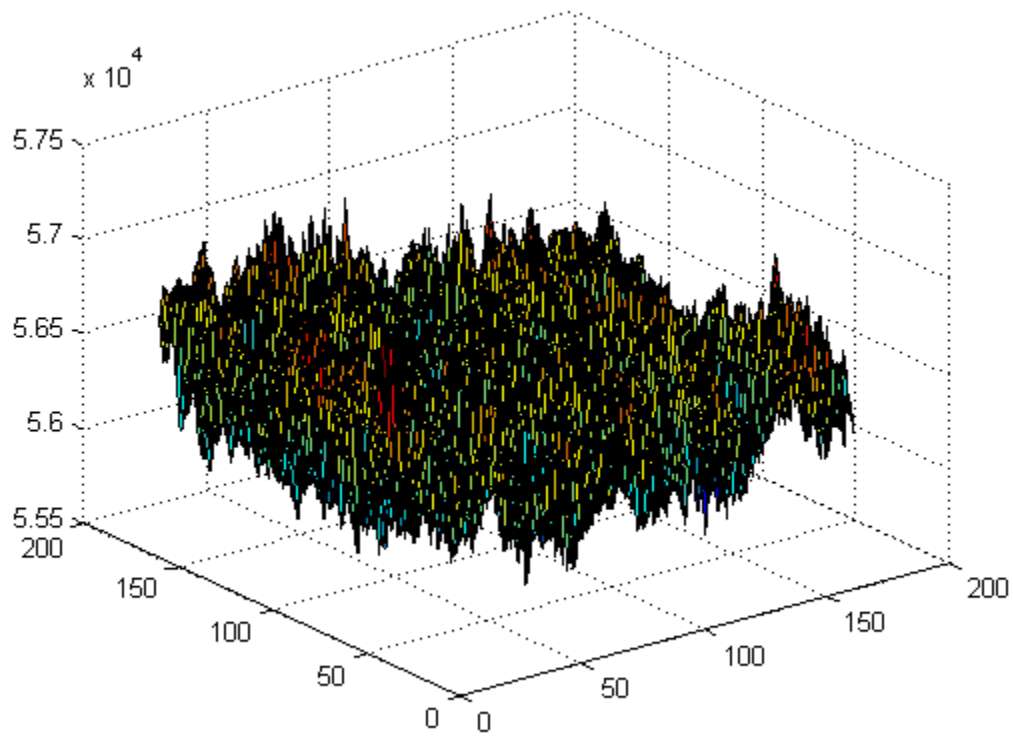


Figure (5.4.23): shad-surface plot for image number 6

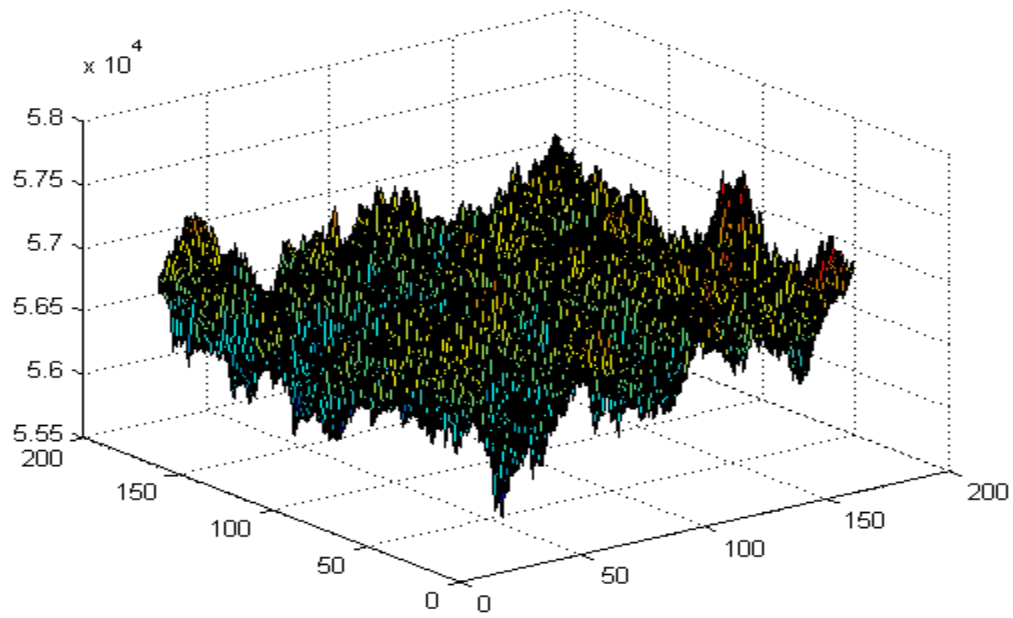


Figure (5.4.24): shad-surface plot for image number 7

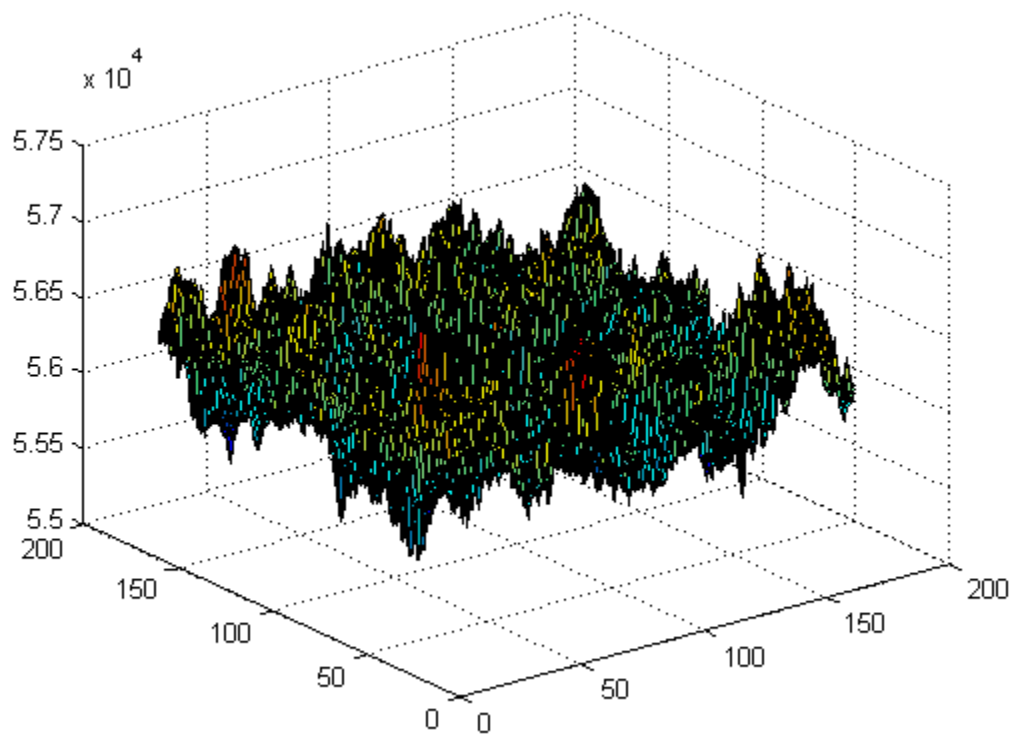


Figure (5.4.25): shad-surface plot for image number 8

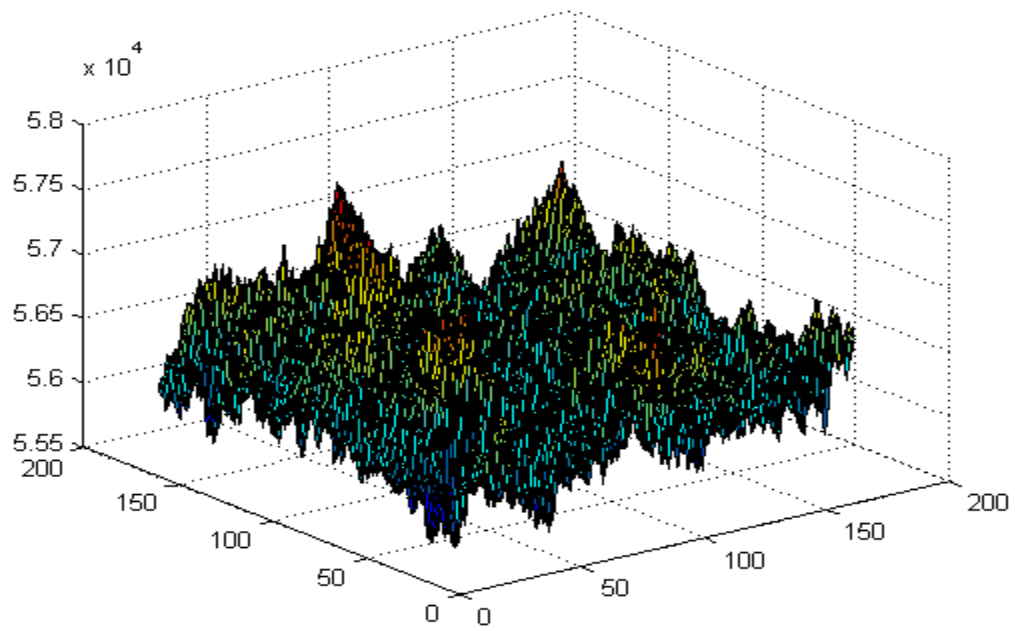


Figure (5.4.26): shad-surface plot for image number 9

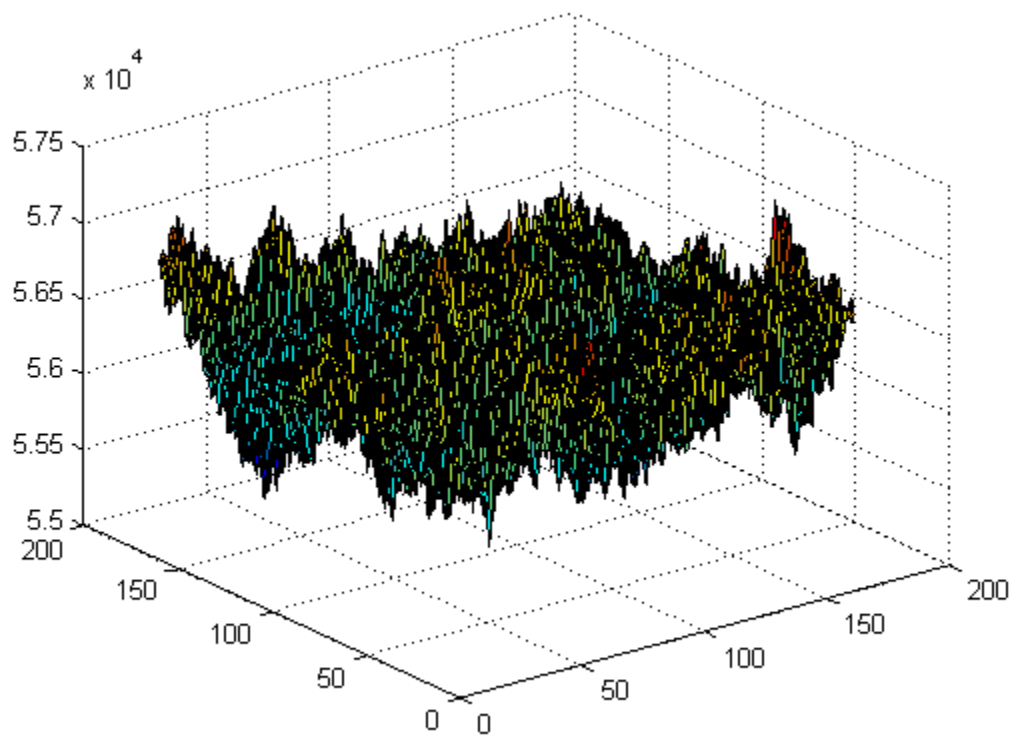


Figure (5.4.27): shad-surface plot for image number 10

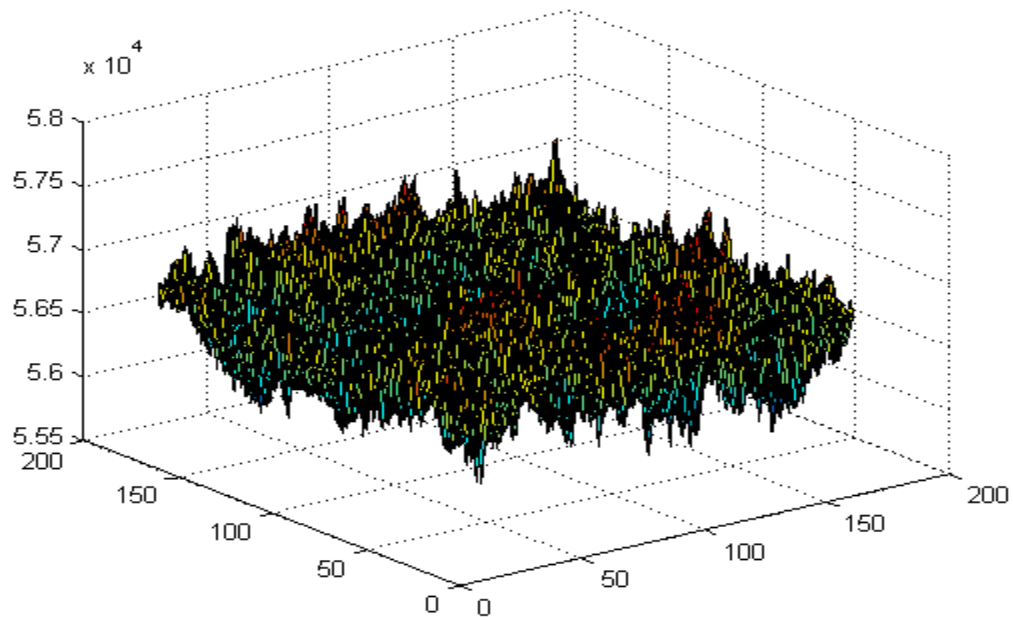


Figure (5.4.28): shad-surface plot for image number 11

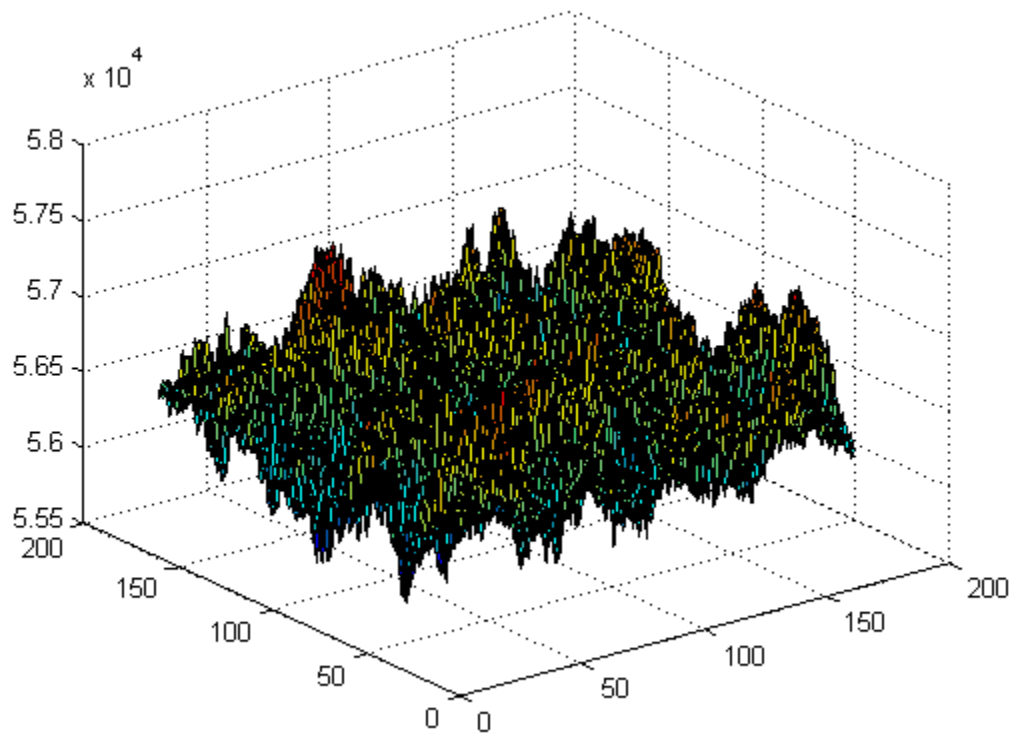


Figure (5.4.29): shad-surface plot for image number 12

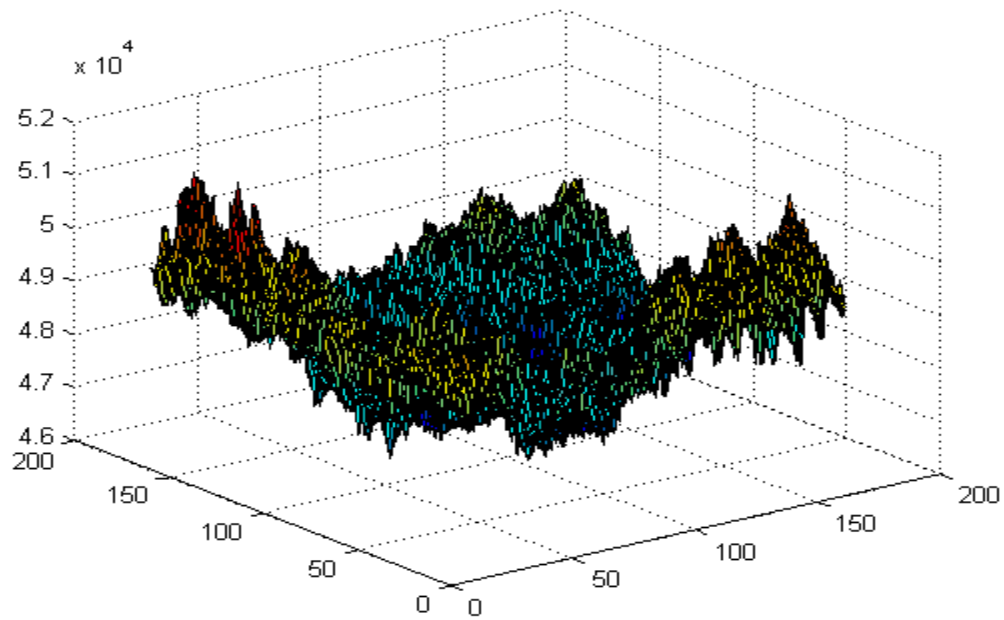


Figure (5.4.30): shad-surface plot for image number 13

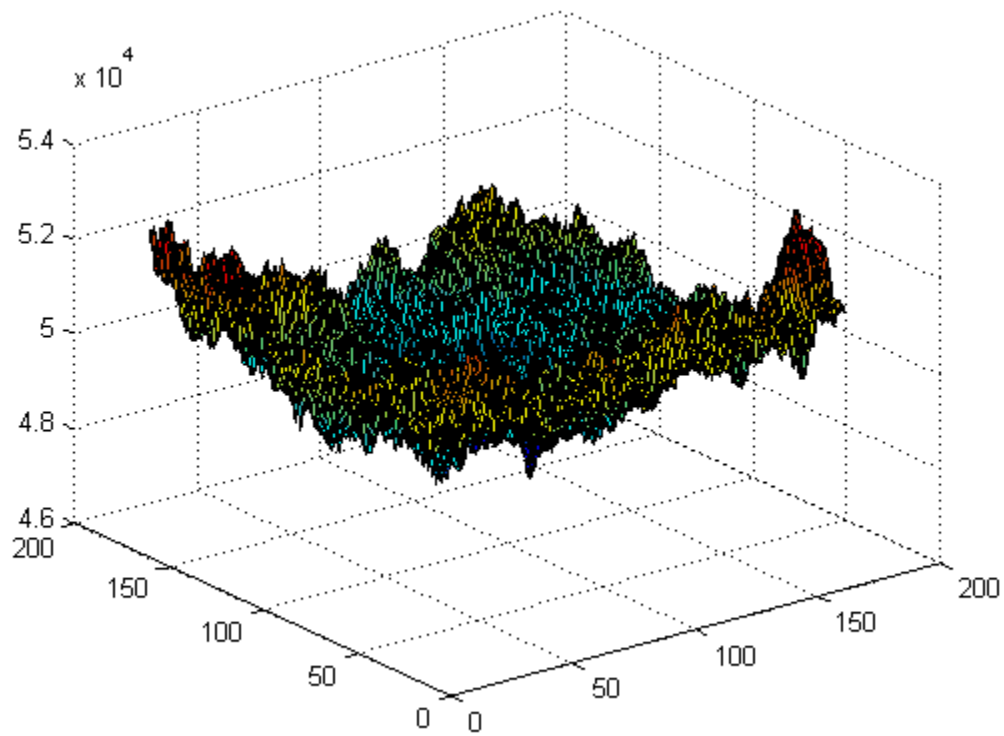


Figure (5.4.31): shad-surface plot for image number 14

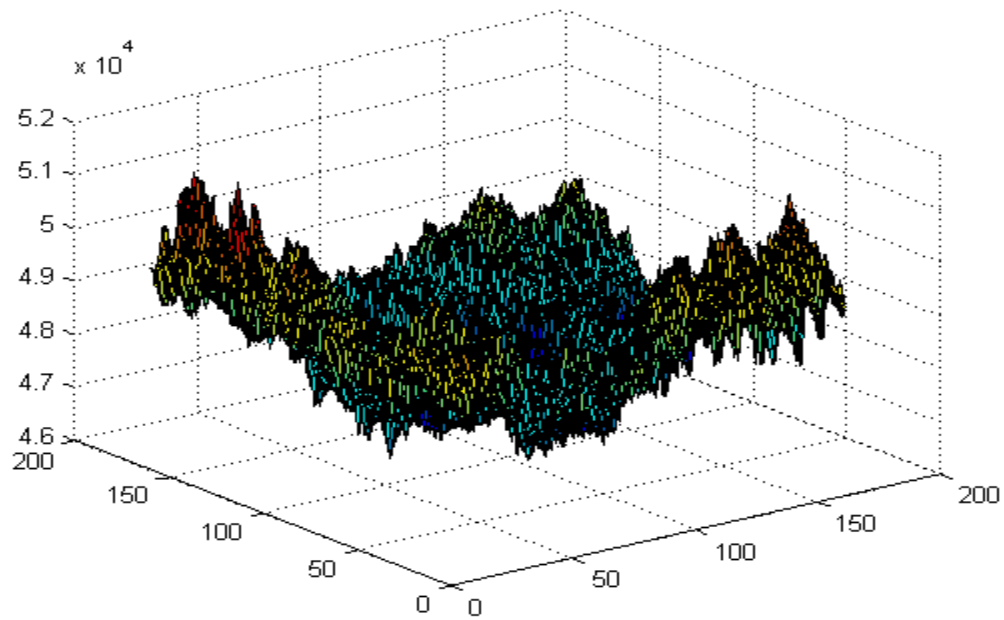


Figure (5.4.32): shad-surface plot for image number 15

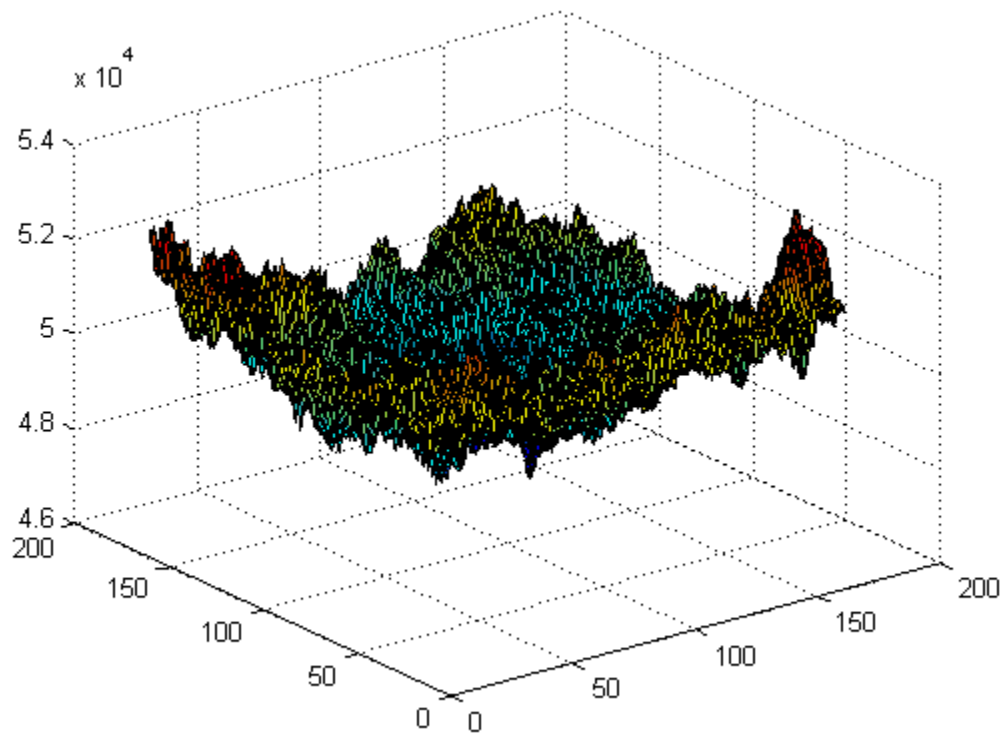


Figure (5.4.33): X-image surface and Y-image surface plot for image number 0

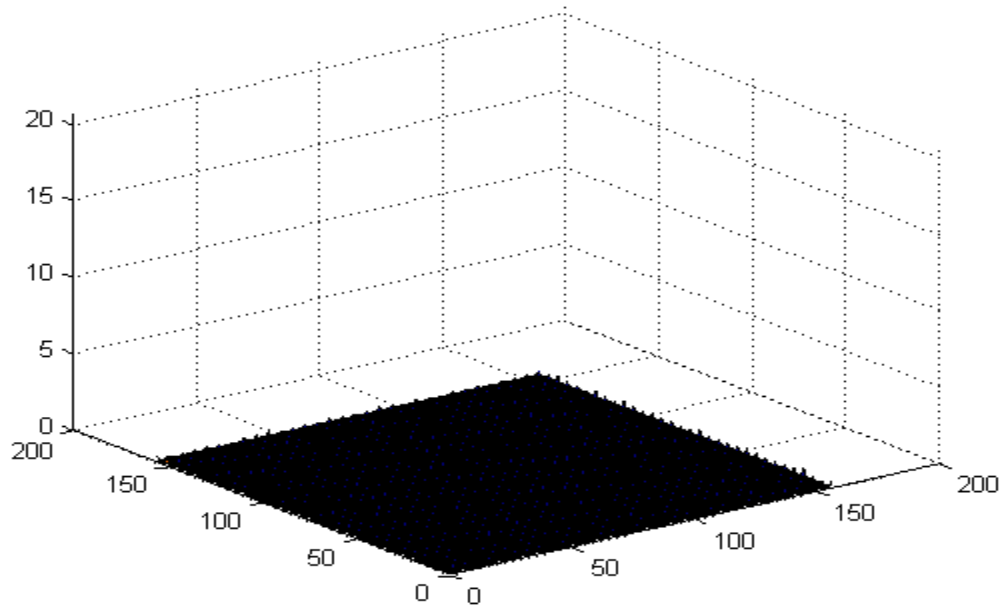


Figure (5.4.34): X-image surface and Y-image surface plot for image number 1

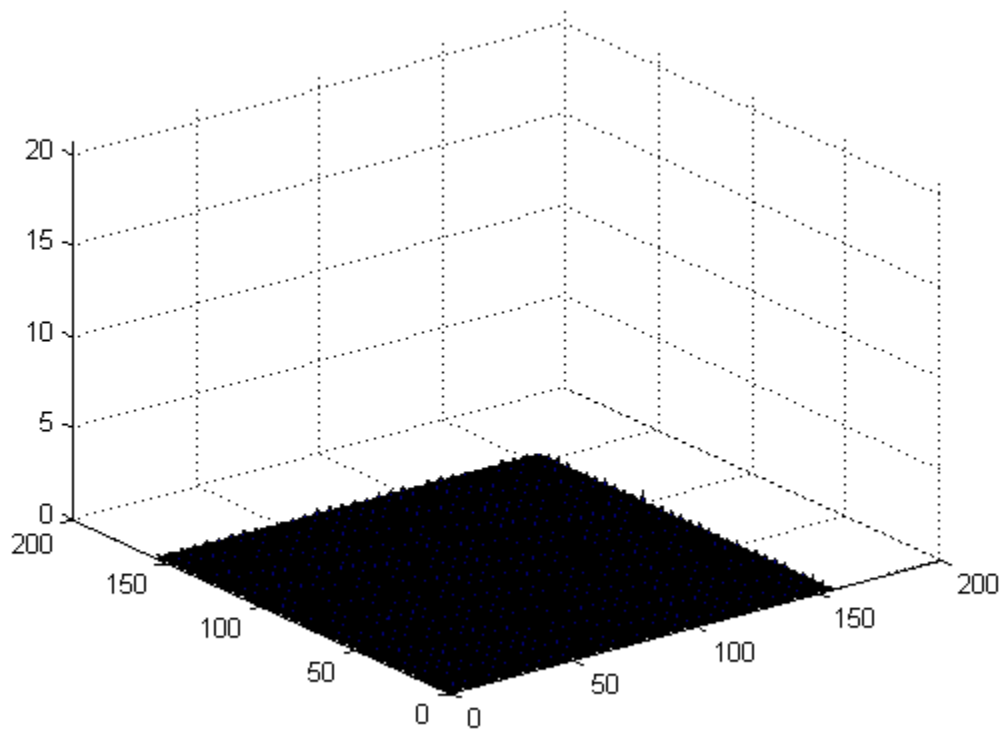


Figure (5.4.35): X-image surface and Y-image surface plot for image number 2

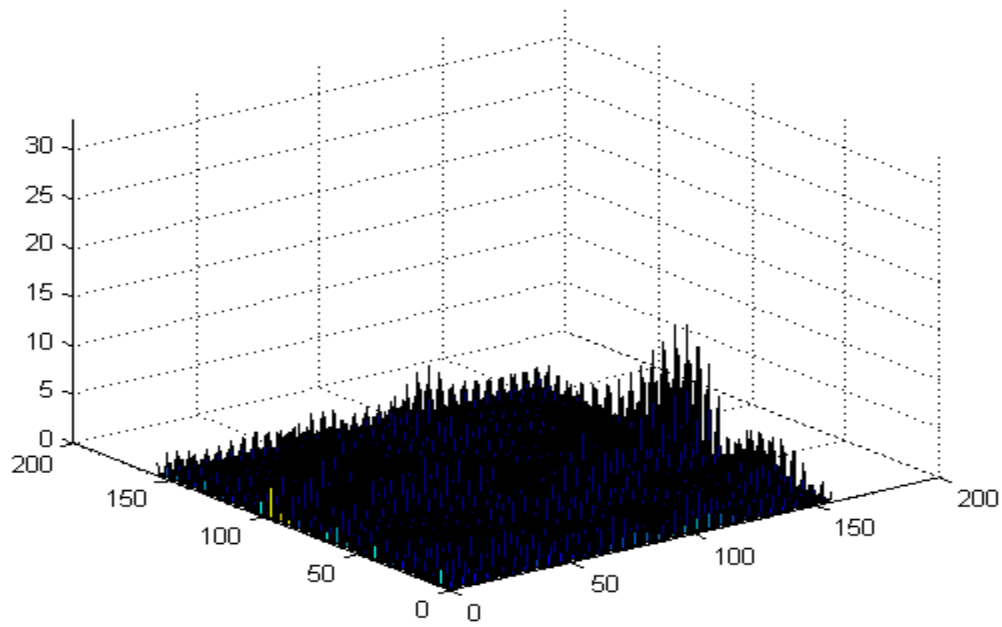


Figure (5.4.36): X-image surface and Y-image surface plot for image number 3

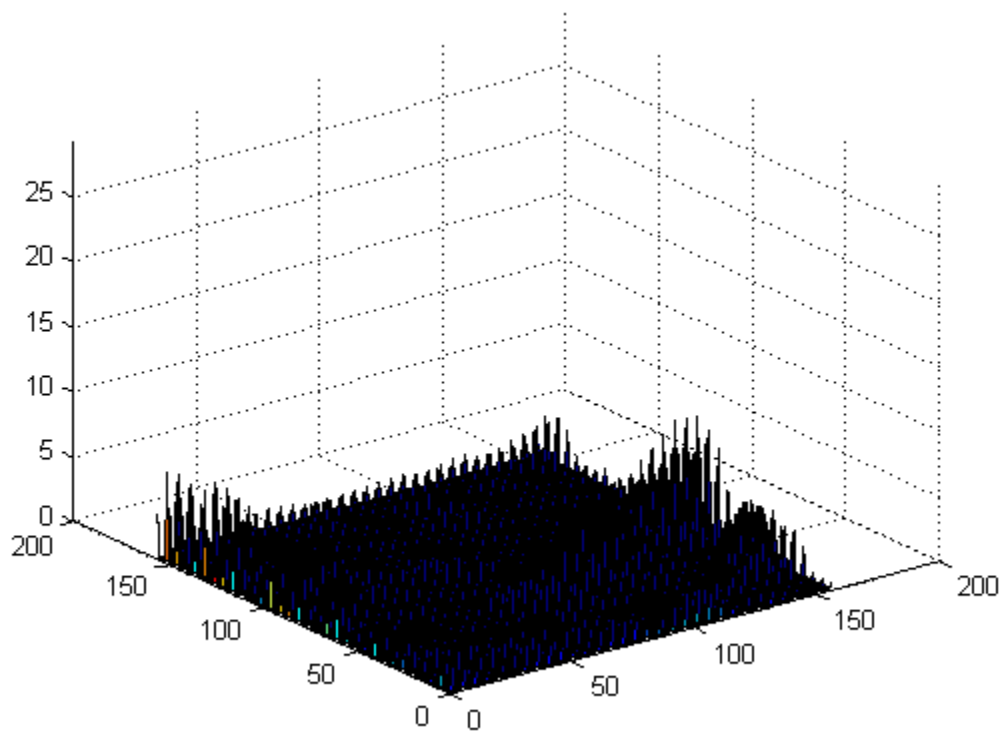


Figure (5.4.37): X-image surface and Y-image surface plot for image number 4

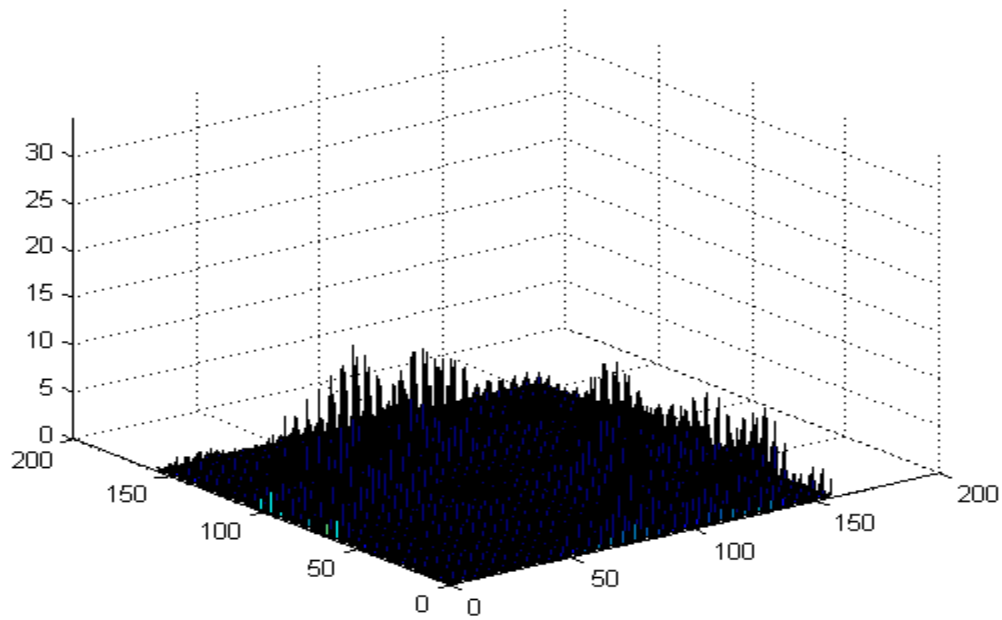


Figure (5.4.38): X-image surface and Y-image surface plot for image number 5

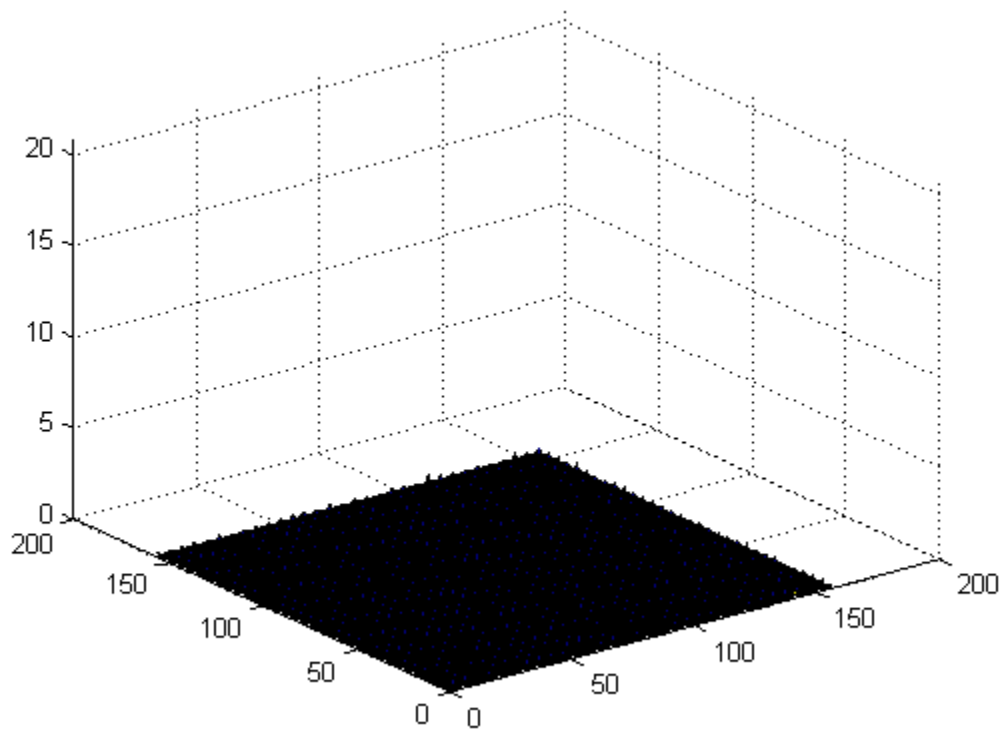


Figure (5.4.39): X-image surface and Y-image surface plot for image number 6

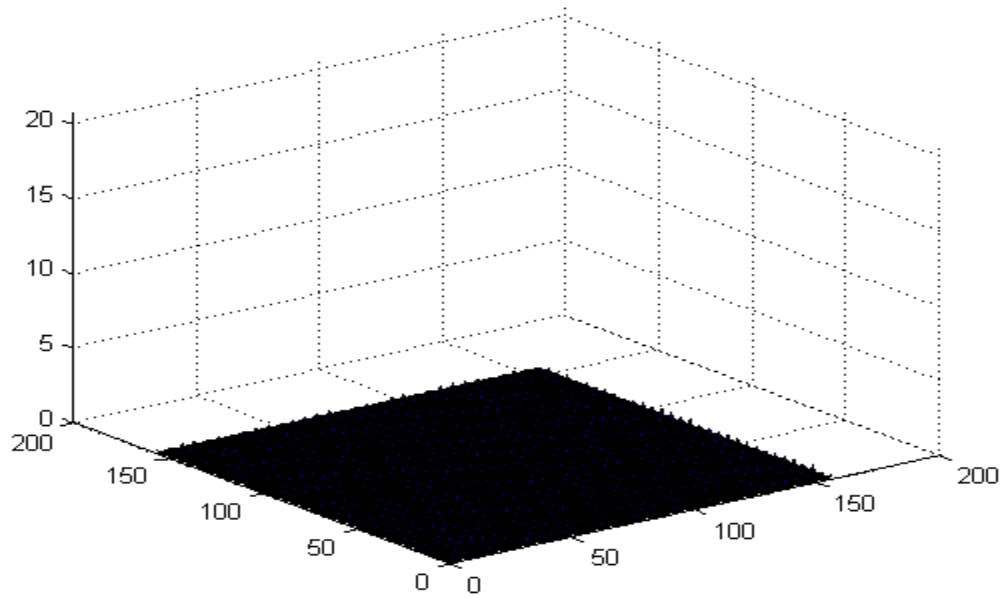


Figure (5.4.40): X-image surface and Y-image surface plot for image number 7

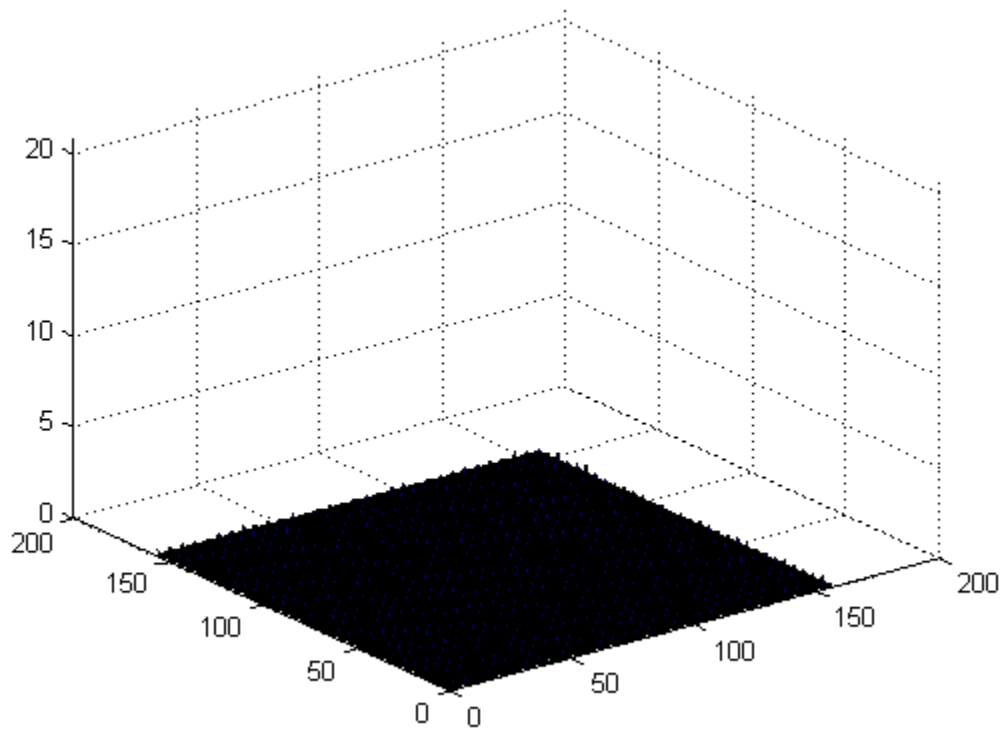


Figure (5.4.41): X-image surface and Y-image surface plot for image number 8

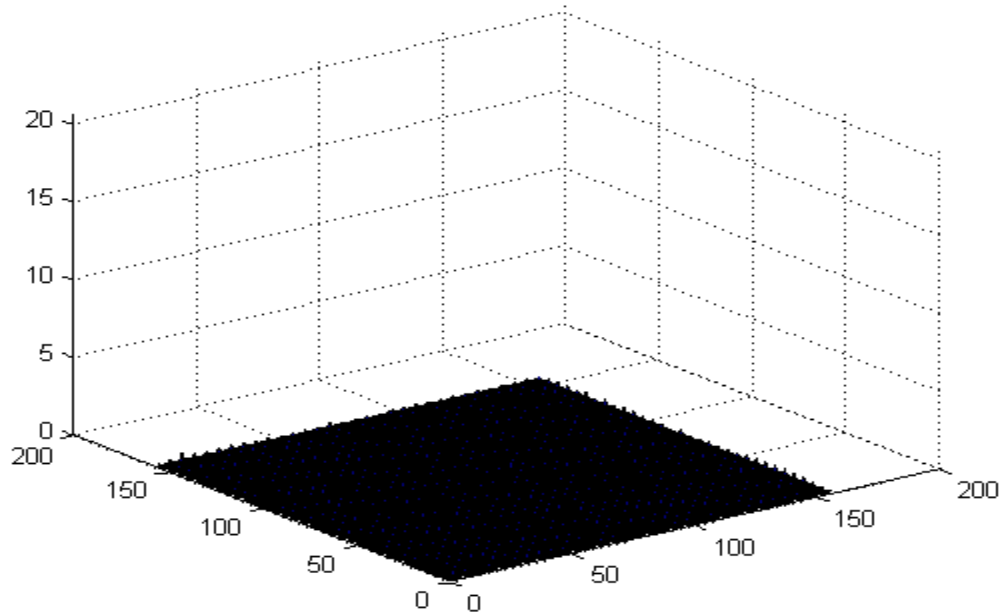


Figure (5.4.42): X-image surface and Y-image surface plot for image number 9

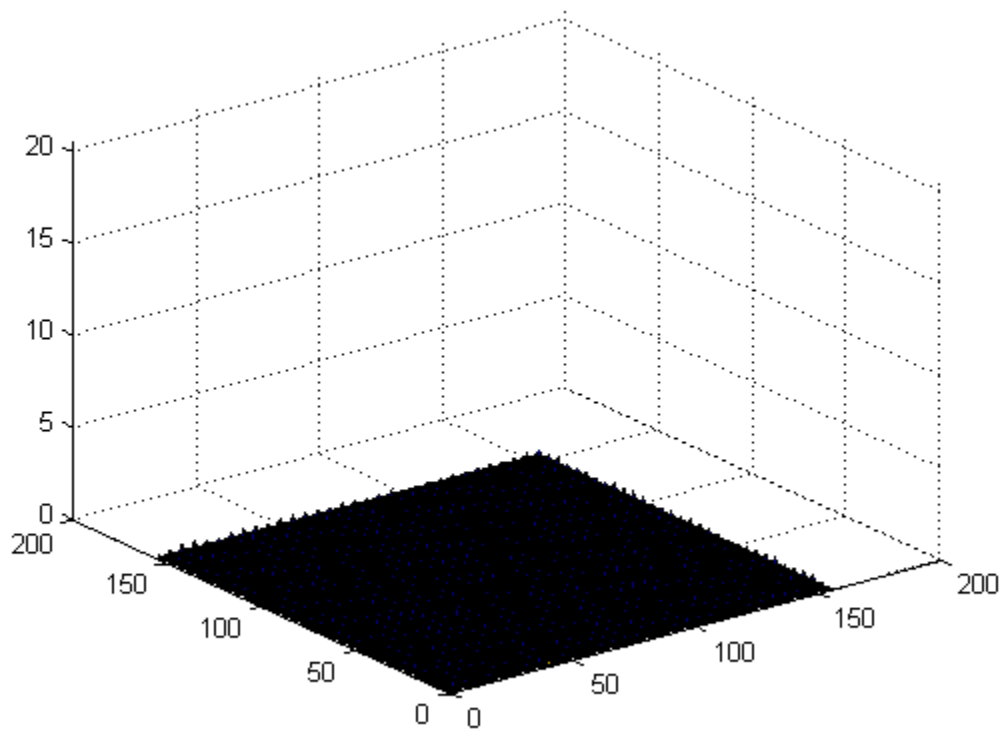


Figure (5.4.43): X-image surface and Y-image surface plot for image number 10

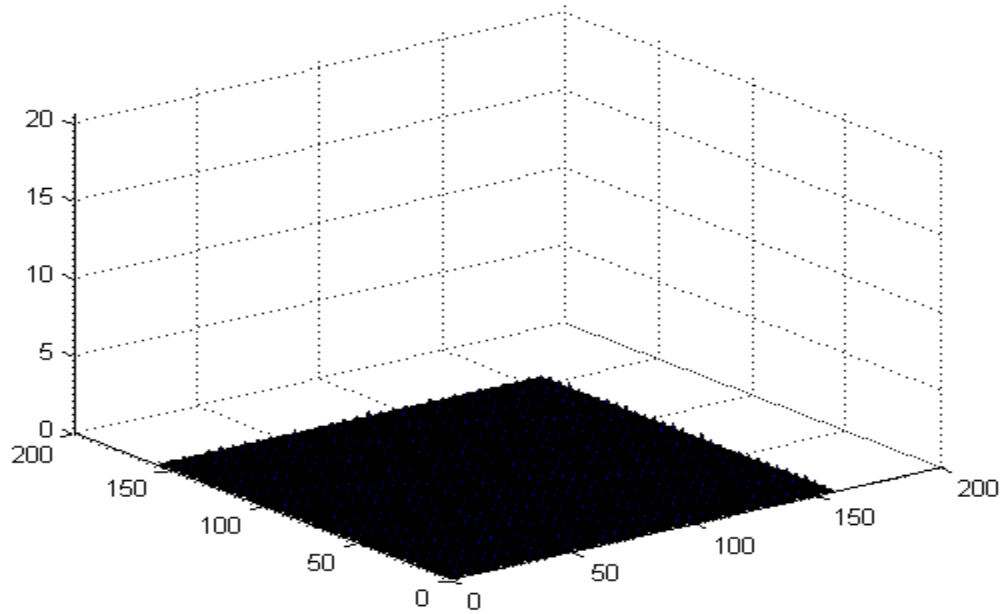


Figure (5.4.44): X-image surface and Y-image surface plot for image number 11

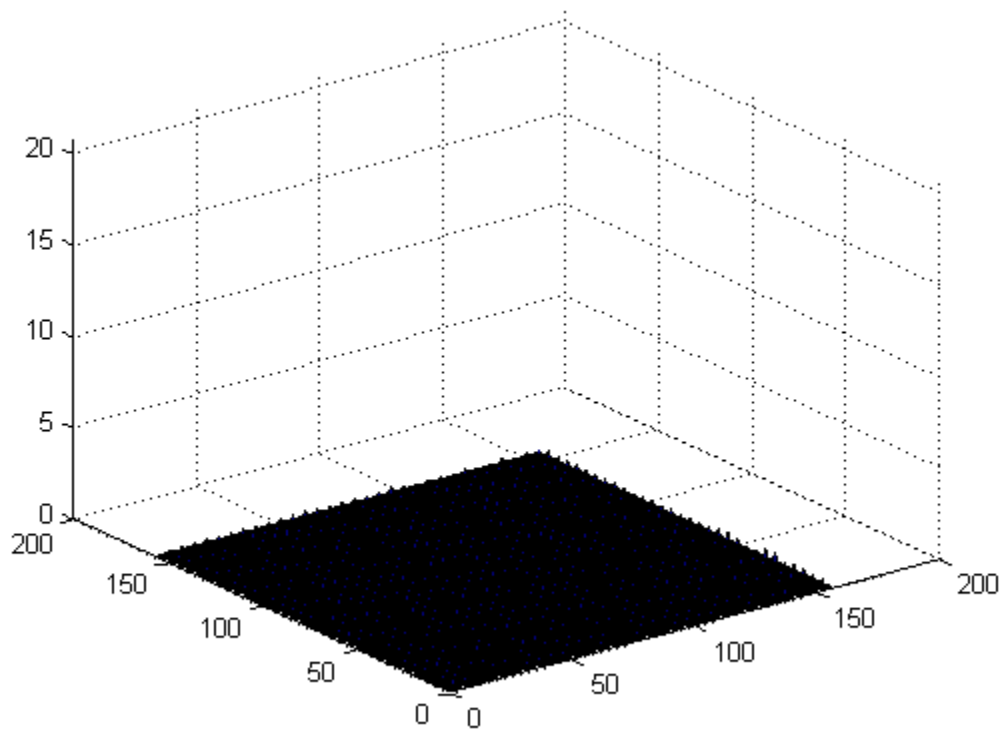


Figure (5.4.45): X-image surface and Y-image surface plot for image number 12

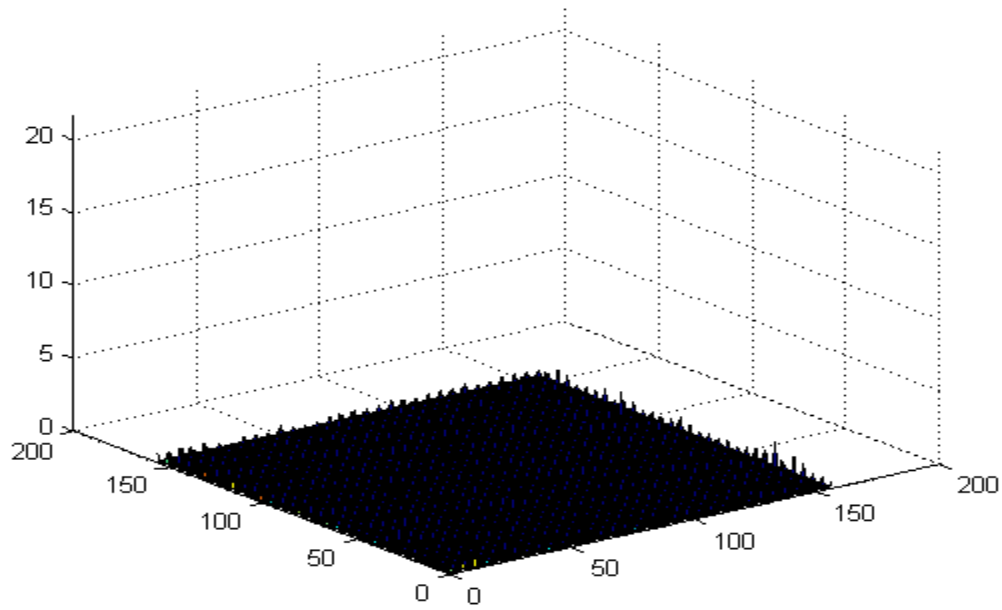


Figure (5.4.46): X-image surface and Y-image surface plot for image number 13

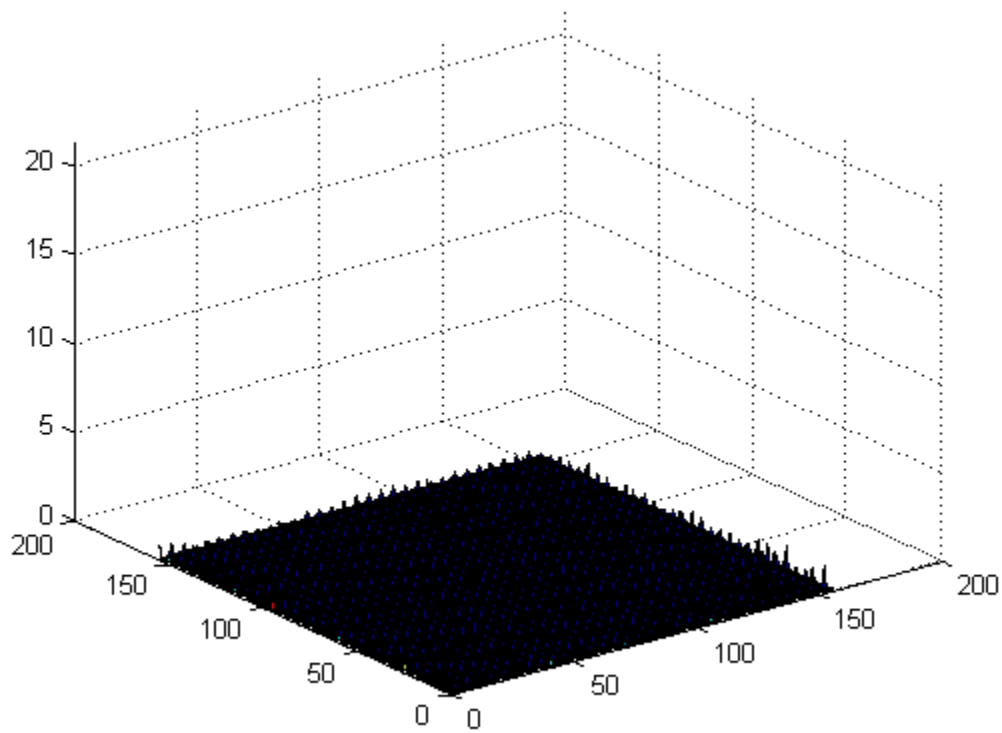


Figure (5.4.47): X-image surface and Y-image surface plot for image number 14

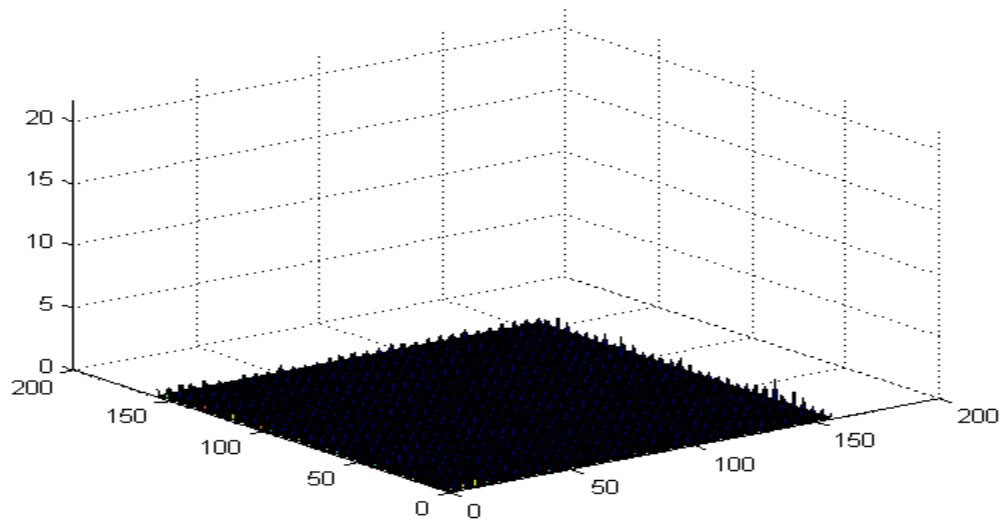


Figure (5.4.48): X-image surface and Y-image surface plot for image number 15

