CHAPTER 2: INTRODUCTION TO ELECTRONIC IMAGING

We presume that the reader understands conventional radiography in which photographic methods capture radiographic images. Typically, a patient is positioned between an x-ray source and a film/screen cassette. The cassette contains an intensifying screen which, when exposed to radiation, emits light which exposes a photographic emulsion. The photographic film can be developed to provide an image to the observer. Traditionally, this is how radiographic images were obtained, and while many radiology departments still use conventional film-screen systems many have converted to digital imaging.

Figure 2-1. X-RAY IMAGING TECHNIQUES IN DIAGNOSTIC RADIOLOGY

CONVENTIONAL FILM-SCREEN
XERORADIOGRAPHY
IMAGE INTENSIFIER/VIDEO SYSTEMS
    a. Abdominal Studies
    b. Angiography
    c. Cardiovascular
DIGITAL VIDEO IMAGING
    a. Digital Subtraction Angiography (Temporal Subtraction)
    b. Dual-Energy Subtraction Angiography
DIGITAL PLANAR IMAGING
    a. Film Digitization
    b. Photostimulable Phosphor Technology
    c. Direct Digital Systems
    d. Scanning Detector Arrays
COMPUTED TOMOGRAPHY

There are many other ways that images are acquired in diagnostic radiology (Figure 2-1). Increasingly images in radiology are captured, processed, displayed, and stored electronically rather than on film. In this chapter, we will discuss several methods to obtain an electronic image from roentgenographic data. This discussion will be a survey of these techniques, since each of these topics will be covered more in depth in later chapters. However, they are provided here as the basis of our discussion of contrast, noise, and spatial resolution in medical imaging.

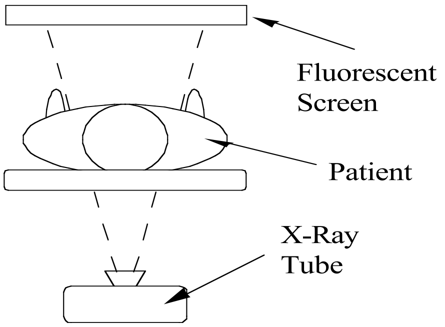
2.1 Common Detectors Used in Diagnostic Radiology

Image Intensifiers

While a film-screen cassette can be used to capture “static” images of stationary objects, “dynamic” images of moving body parts are important for certain diagnoses. This is especially true in the circulatory system where a series of images must be acquired to monitor cardiac motion, image or measure the flow of a contrast agent through the circulatory system, or follow pulmonary function.

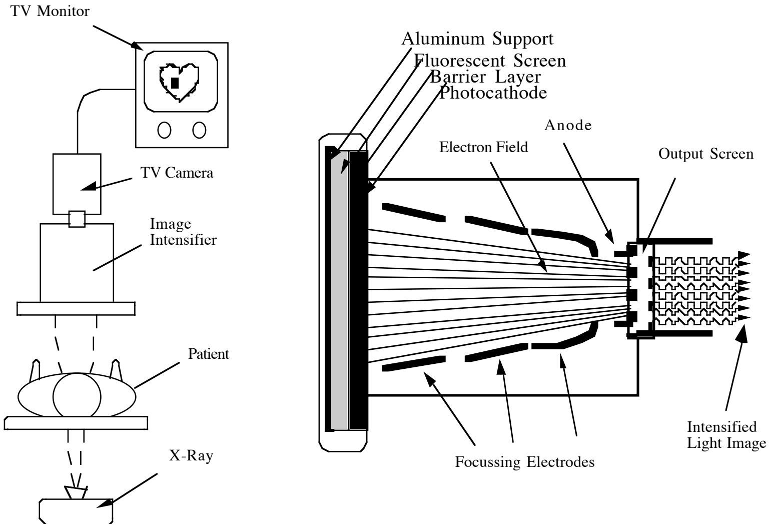
In the "old days", radiologists used fluorescent screens (Figure 2-2) to observe the motion of body parts under examination. A fluorescent screen (or intensifying screen) is one that emits light when exposed to radiation. The patient's body is positioned between the intensifying screen and the x-ray tube. X-rays emitted by the x-ray tube, pass through the patient's body and strike the intensifying screen, allowing the radiologist to view the x-ray image.

Figure 2-2. Conventional Fluoroscopy



There were three basic problems with this technique. First, the fluorescent image was very dim. The radiologist had to view the image in a darkened room. In fact, the radiologist had to stay in the darkened room, or wear dark red glasses, so that their eyes remained dark-adapted. Second, if the radiologist looked directly at the fluorescent screen, his or her head was in the path of the x-ray beam. This problem could be overcome by viewing the image through one or more mirrors, but this was an annoying complication. Finally, the image was so dim that it was impossible to record the fluorescent image with a movie or television camera.

Figure 2-3. Concept of Image Intensifier



In the late 1940's and early 1950's, the image intensifier was introduced as a way to eliminate both the dim fluorescent image and the requirement for direct viewing. It still is used today and provides a means by which dynamics processes in the body can be imaged. In most systems, the radiologist can move the image intensifier while trying to localize an object in the body. For example, an abdominal radiologist generally will use an image intensifier system to examine the gastrointestinal tract of a patient suffering from an ulcer. The patient is given a barium contrast agent either taken orally (or introduced into the GI tract by some other route). The radiologist then can look at the lining of the GI tract after it is coated with the contrast agent to find areas where the ulcer may be located.

As its name suggests, the purpose of the image intensifier is to amplify the light signal produced by a phosphor to generate a bright x-ray image which can be viewed easily or recorded with a photographic or video camera. The image formation process begins when x-rays strike the input phosphor of the image intensifier (Figure 2-3). The phosphor converts the radiation field into a light image, which is quite dim, but which otherwise could be seen by the human eye. The image intensifier increases the brightness of this image so that it can be viewed by the human eye, or more commonly by a television camera or movie camera. This light amplification is begun by a photocathode that converts the light field into a field of electrons. These electrons are accelerated across a high voltage potential in a way that preserves the geometrical configuration of the image. The electrons then are focussed onto a smaller output phosphor where they are absorbed and their energy converted back into light. The combination of the electronic acceleration and geometric minification produces an image at the output phosphor which is 1000 times to more than 5000 times brighter than the image produced at the input phosphor.

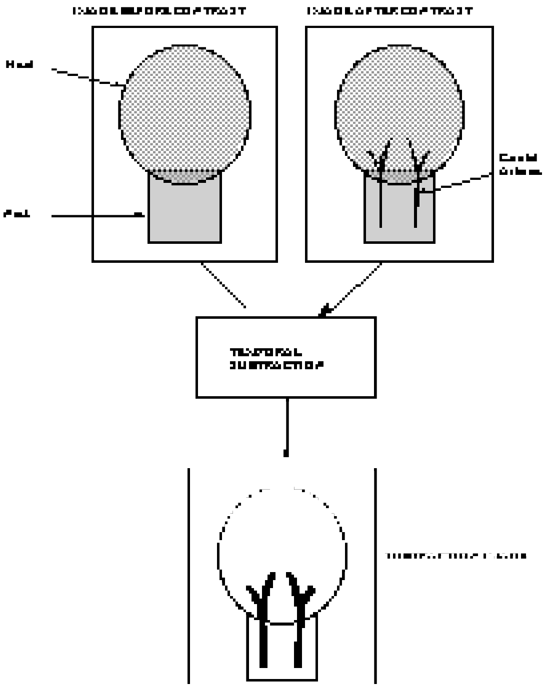


Figure 2-4. Temporal Subtraction

A television camera can record the light image produced by the image intensifier. The television signal then can be viewed on a video monitor by a radiologist. It also can be recorded by a video cassette recorder similar to those we use in homes to record football games and our favorite television programs. Alternatively, the image can be recorded with a movie or cine camera, producing an image on film with better spatial resolution and better contrast than video tape, but which must be chemically processed before it is viewed.

The most important aspect of the image intensifier is that it produces a "live" image in which the radiologist can watch dynamic changes such ventricular contraction in the heart. The image intensifier also can be moved across the patient to survey a large region of anatomy. This is especially useful in abdominal radiography where radiologists must locate a specific region of the patient's internal anatomy prior to obtaining high-resolution images on film. The video signal from the television-image intensifier system also can be sent to a digital image processor for temporal subtraction (Figure 2-4) or energy subtraction (Figure 2-5, Figure 2-6). We will discuss this technique later, in the section describing "digital subtraction angiography", after we finish our discussion of other types of electronic detectors used in radiography.

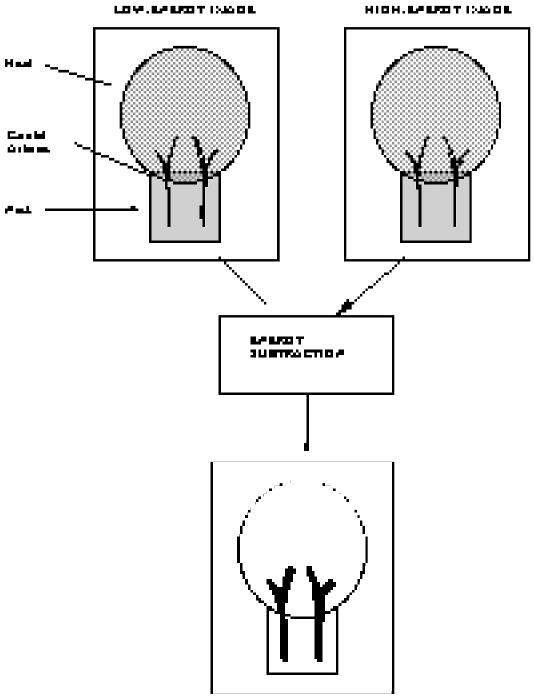


Figure 2-5. Energy-Subtracted Image

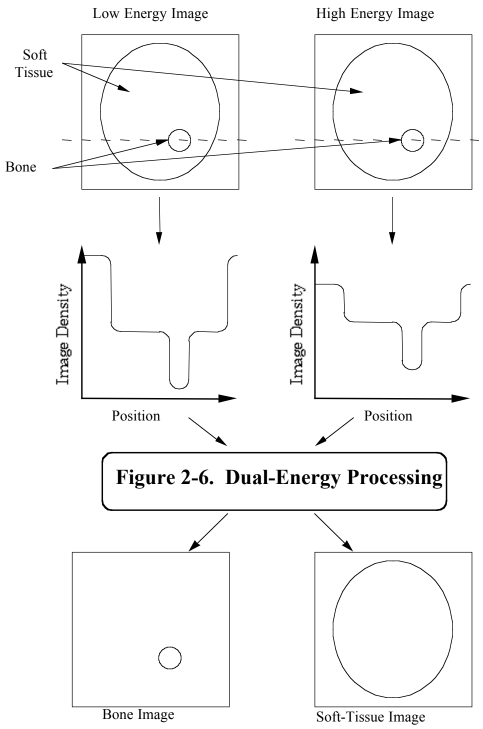


Figure 2-6. Dual-Energy Processing

Scintillation Detectors

Another type of detector found in digital radiographic systems is a scintillation detector (Figure 2-7). It is found in Positron Emission Tomography (PET), Nuclear Medicine, and some older computed tomographic (CT) imaging systems. A scintillator is a material such as sodium iodide or cadmium tungstate that emits light when exposed to radiation, where the brightness of the emitted light is proportional to the amount of radiation absorbed by the detector. To be useful in a digital imaging system, the light signal from the scintillator must be converted into an electronic signal by mounting the scintillator on a photomultiplier tube or a photodiode. Both devices generate an analog electronic signal proportional to the brightness of the light generated by the scintillator. The resulting electronic signal is delivered through an amplifier to an analog-to-digital converter (ADC), producing a digital signal that can be processed and stored in a digital image processor. Unlike the image intensifier that produces an image over a large area, a scintillation detector measures the radiation intensity only at the location of the scintillation crystal. An image is generated by moving the scintillation detector to record the x-ray intensity at each point in an area of interest, or by using an array of detectors to record the x-ray intensity at several points simultaneously. Because scintillation detectors are rather expensive, most medical imaging systems use scintillation detectors arranged in a line or in a circular arc. This detector array must be moved across or around the patient to generate a useful image of a larger area (Figure 2-8).

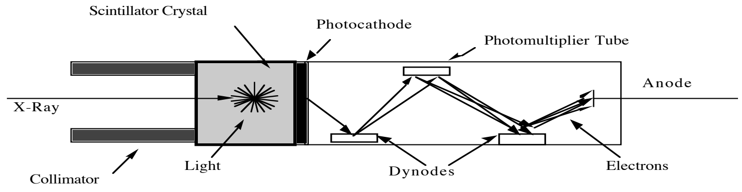


Figure 2-7. Scintillation Crystal/Photomultiplier Tube

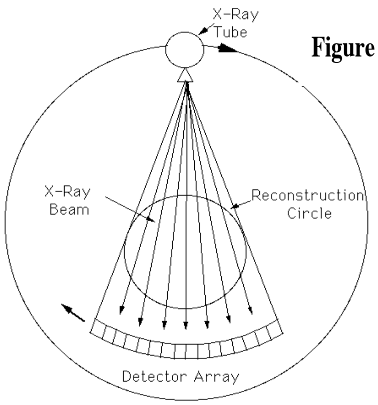


Figure 2-8

Third Generation CT Geometry
(Both Detector and Tube Rotate)

Gas Ionization Detectors

A final detector used in digital radiographic systems is the gas ionization detector (Figure 2-9), which is found in computed tomographic scanners as an alternative to scintillation detectors. Like the scintillation detector, this device measures radiation exposure at a single location. Therefore, these detectors usually are arranged in a line or a circular arc for imaging a patient. A gas ionization detector can be thought of as a volume of gas contained in an enclosure having at least one face that is transparent to x-rays and serves as the radiation window. The detector contains two electrodes, one at a negative electrical potential and one at a positive electrical potential. An x-ray enters the detector through its radiation window and dislodges electrons from the gas molecules in the detector. The electrons are collected by the electrodes and form an electrical current in the detector electronics that is proportional to the radiation exposure entering the detector.

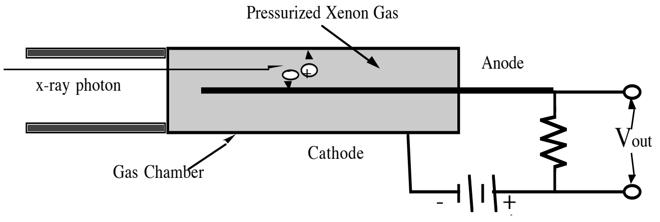


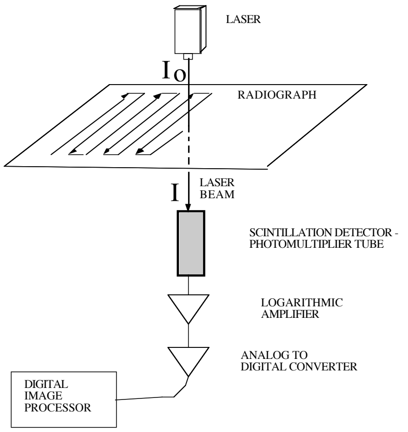
Figure 2-9. Gas Ionization Detector

The primary advantage of the gas ionization detector is that it is inexpensive and can be made relatively compact. Its primary disadvantage is that all gases are poor x-ray attenuators in comparison to solid crystalline scintillators. Therefore, xenon is used in gas ionization detectors since it has excellent x-ray absorbency in comparison to other gases. To increase its x-ray absorbency, the xenon commonly is pressurized to several atmospheres when used in a radiation detector.

2.2 Other Radiation Detection Systems

Image intensifiers are useful for dynamic studies of blood flow and cardiac function and scintillation and gas ionization detectors have been used in several “tomographic” imagers. However, various digital imaging methodologies, including (a) film digitizers, (b) photostimulable phosphor systems, (c) scanned detector arrays and (d) direct digital imagers, are evolving to replace static and dynamic analog imaging in radiology departments.

Figure 2-10. Laser Film Scanner For Film Digitization



Optical Density = -log(transmittance) = log(I_o/I)

Film Digitizers

Film digitization is a technique in which the radiographic film is placed in an instrument that scans the film to generate an electronic signal proportional to its optical density as a function of position. A typical system might be one diagrammed in Figure 2-10. In this instrument, a stable light source (usually a laser) is transmitted through the film, and is recorded with a photomultiplier tube. The optical density of the film is defined as:

$$\text{optical density} = -\log(\text{transmittance}) = \log\left[\frac{I_o}{I}\right] \qquad (2\text{-}1)$$

where

I_o = incident light intensity (w/o film) and I= transmitted light intensity.

If the incident light intensity is known, and the transmitted light intensity can be measured by the photomultiplier tube, then the system shown in Figure 2-10 can produce an electronic signal which is proportional to film density. The electronic signal can be digitized, usually to 10 or 12 bits, with a 1024 x 1024 matrix or finer to provide sample spacing appropriate for the resolution desired.

The advantage of this technique is that it supports conversion of film images to a digital format where they can be managed within a departmental picture archiving and communications system (PACS).

Photostimulable Phosphors

A digital imaging alternative to a film-screen system is provided by photostimulable phosphor screen. A photostimulable phosphor screen (Figure 2-11) emits light (a phosphor) when exposed (or stimulated) by another light (hence photostimulable). When exposed to x-rays, about half of the energy in the x-ray beam is trapped in the photostimulable screen as a "latent image" which is stable over several days. This energy is released by scanning a red laser beam across the plate, generating blue light (or luminescence). The brightness of the luminescence is proportional to the intensity of the x-rays originally striking the screen. The stimulated blue light signal can be measured with a photomultiplier tube, producing an electronic signal representing the radiographic image. The analog signal is digitized and presented to a digital image processor for calculation or storage.

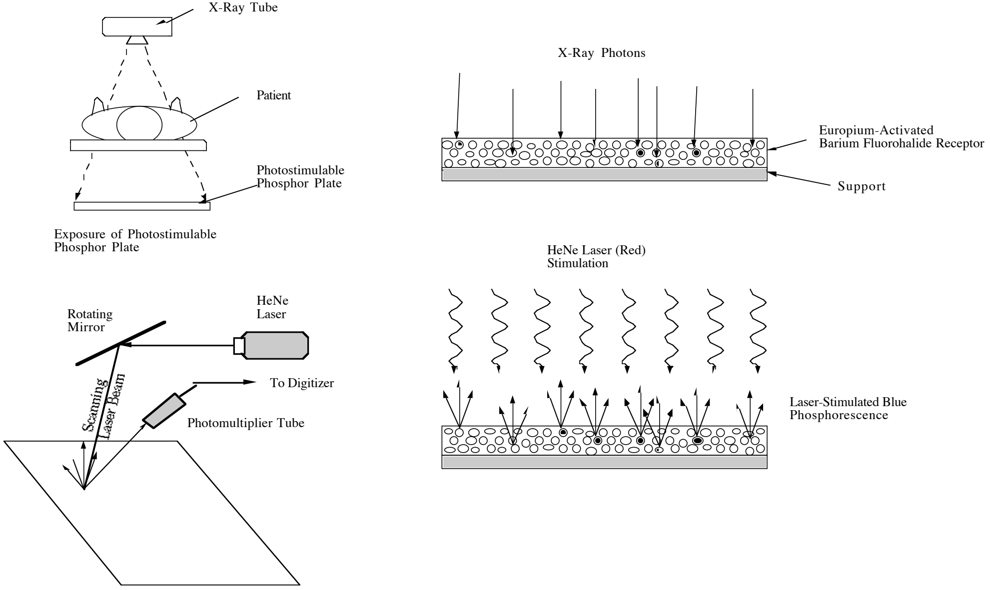


Figure 2-11. Development of Photostimulable Phosphor Plate

Photostimulable phosphors screens have some advantages over film and film digitizers. First, photostimulable phosphors are less prone to be underexposed or overexposed, and capture a wider range of radiation exposure than film (4-5 decades). This is an advantage especially in portable or emergency chest radiography where the detector must record high exposures behind the lung simultaneously with low exposures through the abdomen. Second, the response of photostimulable phosphors is linear. That is, doubling the x-ray exposure will double the response of the phosphor, an important property for quantitative measurements. Finally, the image is obtained electronically, without chemical processing, and the photostimulable plates are reusable. Also, photostimulable phosphor screens can be mounted in conventional size cassettes and can serve as a direct replacement for film-screen cassettes. Among the disadvantages of photostimulable phosphor systems are their cost ($500,000 or more), and the slight degradation in spatial resolution in comparison to film-screen systems, which may be critical in several diseases such as the detection of pneumothorax or diffuse interstitial disease.

Scanning Detector Arrays

Both scintillation and gas ionization detectors can be used in scanned projection systems. In these systems, the radiation beam is collimated to a fan-beam (Figure 2-12) or pencil-beam geometry. The x-ray beam passes from the x-ray tube, through the patient, and onto the detectors which are arranged in a linear array or circular arc. Scanning the x-ray beam and the detector array across the patient forms an image. The electronic detectors accommodate a wide range of x-ray

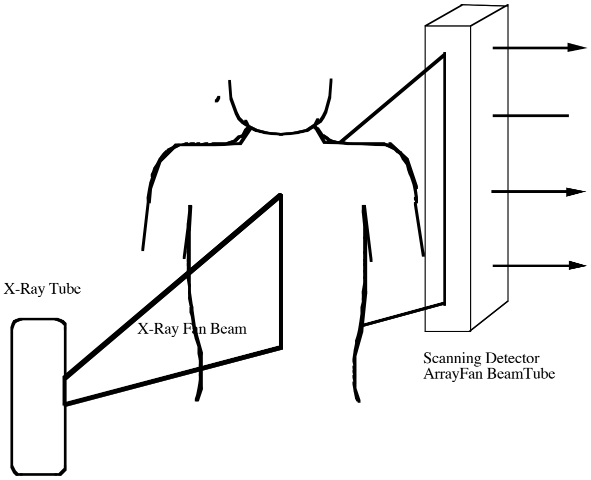


Figure 2-12. Scanning Detector System.

exposures making them less prone to overexposure or underexposure than film. In comparison to broad area geometries, the narrow x-ray beam greatly reduces scatter, thereby improving the detection of subtle differences in x-ray absorption by tissues in the patient. Scanned detector array systems have not been adopted widely for routine clinical studies because they have poorer spatial resolution than film or photostimulable phosphors, and are expensive to purchase and maintain in comparison to conventional radiographic systems. Additionally, x-ray CT systems can be used as scanned detector array systems, as exemplified by scout studies used to select body range for CT acquisition.

Direct Digital Imaging

In support of the growing trend in digital radiography departments, modern systems have emerged to provide direct conversion from analog to digital x-ray images. These systems consist of a flat panel containing a large array of small detection units. X-ray detection units can be indirect type, with a scintillation region paired with photosensitive detector, or direct type where the detector and electronic recording parts are combined. In either case the x-ray image is converted to an array of electronic signals, which are subsequently stored as digital images. Direct digital imaging of projection x-ray images is provided for both static and dynamic imaging needs.

Computed Tomography (CT). Historically CT has principally been use to refer to x-ray CT, but any approach that computes tomographic section images (tomo means cut) can be classified as computed tomography. CT therefore includes popular modern radiological imaging methods such as single photon emission computed tomography (SPECT) used in Nuclear Medicine departments, Positron Emission Tomography (PET) used for cancer imaging of F^{18} FDG, Magnetic Resonance Imaging, and even some forms of ultrasound imaging. CT images due to their computed nature are digital images, the topic introduced in Chapter 3.

Homework Problems.

1. The optical density difference between two regions of an x-ray film is 0.7. What is the ratio of the intensity of transmitted light between these two regions? Assuming that the film-screen has a gamma of 0.76, what is the ratio of the radiation exposure between these two regions?

2. Describe the process of latent image generation for a photostimulable phosphor plate digital radiographic system. Why is the dynamic range of these image receptors so high compared to film-screen systems?

3. Estimate the reduction in scatter produced by a 1 mm collimated x-ray scanning system (like in Figure 2-12) compared with a conventional x-ray beam. Assume that the object is 100x100x100-cm phantom that is placed on the image detector and that the entire phantom is imaged. Make your calculation based only on the geometrical efficiency at the center of the detector. Report the ratio of scatter using the slit to that without it.

4. The relationship between pixel value and exposure for a photostimulable phosphor plate is given by the following equation:

   $$\text{Pixel Value} = C_1 \log_{10}(E) + C_2$$

   where E is the exposure of the plate in mR and C_1 & C_2 are constants for the plate.

   You make the following measurements in an attempt to determine the relationship between pixel value and exposure:

Pixel Value	1019	738	485	260	7
Exposure (mR)	100	10	1	0.1	0.01

   While the exposure was measured quite accurately the pixel values were a bit noisy.

   a. Use a least square error fit method to estimate the values of C_1 and C_2.

   b. Plot the raw data and the fitted data on a graph of pixel value vs. $\log_{10}(E)$.

   c. What is the R^2 value for the fit?