AAPM Tutorial

CT Image Detail and Noise[1]

Perry Sprawls, PhD

Two important characteristics of the computed tomographic (CT) image that affect the ability to visualize anatomic structures and pathologic features are blur and noise. Increased blurring reduces the visibility of small objects (image detail); increased visual noise reduces the visibility of low-contrast objects. Sources of blurring in CT include the size of the sampling aperture (which can be regulated by the focal spot size and the detector size), the size of the voxels, and the reconstruction filter selected. Noise is caused by the variation in attenuation coefficients between voxels. Use of small voxels and edge-enhancing filters helps reduce blurring and improve visibility of fine details. However, small voxels absorb fewer photons and therefore result in increased noise. Noise can be reduced by using large voxels, increasing radiation dose, or using a smoothing filter, but this filter increases blurring. An optimized protocol for a specific clinical study must take these physical principles into account and be adjusted to give proper balance among detail, low noise, and patient exposure.

INTRODUCTION

Computed tomography (CT) is characterized by a high-contrast sensitivity that makes it possible to depict the differences among soft tissues within the body. However, CT has two other characteristics that affect the ability to visualize anatomic structures and pathologic features: (*a*) blurring of the image, which affects the visibility of small objects (image detail), and (*b*) visual noise, which reduces visibility of low-contrast objects.

This article shows how visibility is influenced by these two characteristics, identifies the sources of blurring and noise, and shows how they can be controlled by the operator to optimize image quality for different clinical requirements.

LESION CHARACTERISTICS

Lesions have two characteristics that affect their visibility: inherent contrast and size. In CT, inherent contrast is the extent to which the attenuation coefficient value of a lesion differs from that of the surrounding tissue. If a lesion has attenuation values that differ greatly from those of surrounding tissue, it has high contrast and will be

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[1] From the Department of Radiology, Emory University School of Medicine, 1364 Clifton Rd NE, Atlanta, GA 30322. From the 1991 RSNA scientific assembly. Received May 28, 1992; revision requested June 9 and received July 2; accepted July 6. **Address reprint requests** to the author.

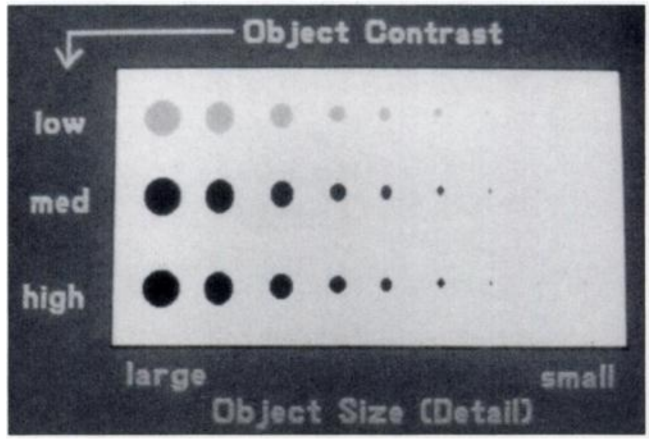


Figure 1. Potential lesions (objects) are arranged in a matrix according to their inherent contrast and size.

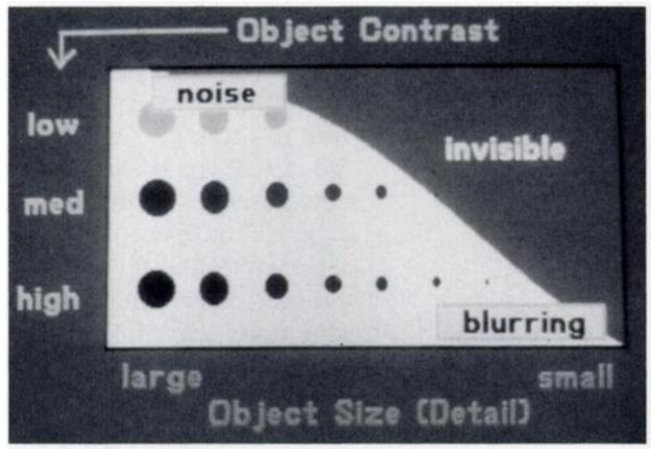


Figure 2. The matrix of potential lesions shown in Figure 1 is depicted here with a curtain of invisibility, which covers the small low-contrast lesion in the upper right corner, as well as many other lesions. Because of image blurring and noise (the components of this curtain), some lesions in the body will not be visible.

easily visualized. Physical size is also thought of in terms of detail to be discerned. Generally, the smaller the object, the greater the capabilities must be for detecting detail, specifically anatomic detail.

Figure 1 illustrates how inherent contrast and size interact and affect our ability to see lesions. The large, high-contrast lesion in the lower left corner is very easy to see. The challenge would be to detect the lesion that is both small and low contrast at the upper right.

IMAGE QUALITY

Each time an imaging procedure is conducted, there is actually a large range of potential lesions that are not visible. There is a distinct dividing line between those lesions in the body that are visible and those that are invisible. This boundary is one form of a ***contrast detail curve,*** which can be thought of as a "curtain of invisibility" (Fig 2).

We are generally concerned with image quality and improving image quality, which entails pushing the curtain of invisibility back so that more potential lesions or other objects are revealed. By changing certain imaging factors, we can raise the curtain and increase visibility. In doing this, however, we must be aware of other compromises in the imaging process. Two other specific characteristics of an image besides contrast have an impact on visibility: (*a*) visual noise, which affects visibility of low-contrast objects, and (*b*) blurring, which affects visibility of small objects and anatomic detail (Fig 2).

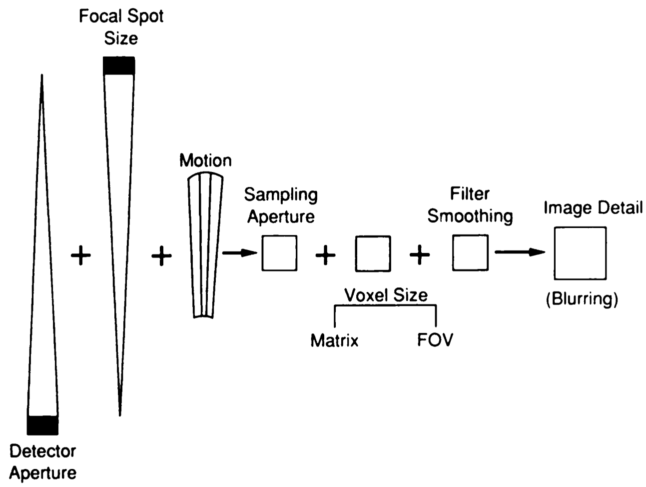


Figure 3. Diagram illustrates the sources of blurring that limit visibility of detail in a CT image. *FOV* = field of view. (Reprinted, with permission, from Sprawls P. The physical principles of medical imaging. Rockville, Md: Aspen, 1987.)

All image-forming procedures suffer from blurring. In human vision, we occasionally suffer from some level of blurring, which limits our ability to see small objects, fine print, and other details. Operators and users of CT can control the amount of noise and the amount of blurring by selecting appropriate values for various imaging protocol factors. In general, as we set up a protocol for a specific clinical procedure, we are adjusting the level of noise and the level of blurring. We are affecting the range of potential lesions that will be visible.

Blurring

As blurring increases, the visibility of the smaller objects is reduced. In each imaging modality, there are specific sources of image blurring. In conventional radiography, visibility of detail is limited by blurring from three sources: the focal spot size, the type of receptor, and motion. Unlike radiography, which is affected by blurring only during image acquisition, CT can be affected in the scanning stage and in the image reconstruction stage (Fig 3).

During the scanning step of the image-forming process, blurring is produced by the finite size of the focal spot (as in radiography) and by the finite size of the radiation detector, similar in principle to the blurring associated with radiographic receptors. For example, a ray is the segment of an x-ray beam defined by the size of the focal spot on one end and by the size of the detector on the other end. The cross-sectional dimension of a ray, often referred to as the sampling aperature, depends on the distance from both the focal spot and the detector. If the sampling aperature (ie, the ray) is wider than the object being imaged (ie, the ray is "seeing" not just the object but also the surrounding tissue), the image of this small object will be blurred.

In some CT scanners, the user can choose the focal spot size and, obviously, would select the small focal spot to give the greatest detail. In a few scanners, the user can select and change the effective size of the detector; again, for maximum detail, a relatively small detector aperture should be used.

In the image reconstruction process, the section of tissue that was imaged is subdivided into individual tissue voxels, each of which has a dimension. All structures within an individual voxel are mixed together and represented by a single CT number. In principle, a voxel is a three-dimensional blur. We cannot see any detail within a voxel; when we look at images, we are seeing voxels side by side. To achieve an image with high detail, it is necessary to use small voxels.

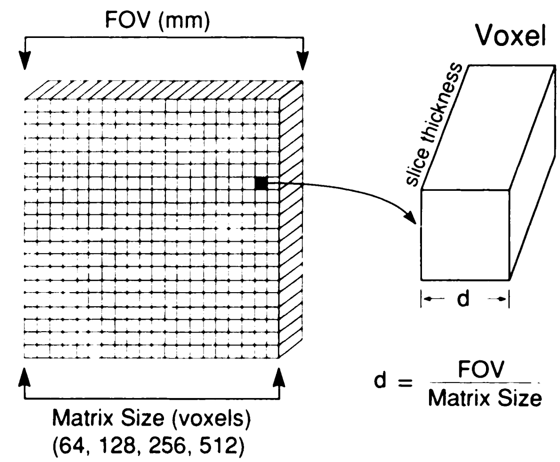
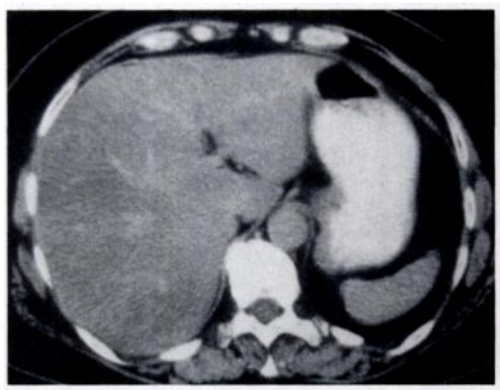


Figure 4. Diagram illustrates how the size of an individual voxel is determined by three factors: field of view (*FOV*), matrix size, and section thickness. (Reprinted, with permission, from Sprawls P. The physical principles of medical imaging. Rockville, Md: Aspen, 1987.)



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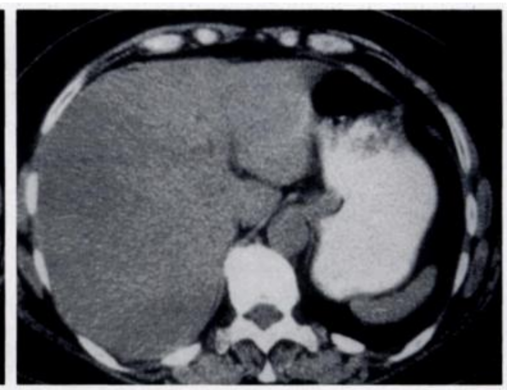

6.

Figures 5, 6. **(5)** CT scan with a moderate level of image noise demonstrates vascular structures within the liver. **(6)** CT scan with a higher level of noise demonstrates how noise reduces the visibility of the vessels.

There are three protocol factors that can be adjusted that control voxel size and image detail: field of view, matrix size, and section thickness (Fig 4). The face dimension of a voxel (sometimes erroneously referred to as the dimension of the pixel) is the ratio of the field of view (the actual size of the anatomic region that we are imaging) to the matrix size (the number of voxels across the dimension of the image). The other dimension of the voxel is the section thickness, which, in most instances, will be the largest dimension. Section thickness is typically adjusted from about 1 to 10 mm.

The final factor that determines the amount of blurring produced in a CT image is the reconstruction filter. Reconstruction filters are mathematical operations that are used to alter characteristics of the image. In certain cases, it is desirable to use a so-called smoothing filter, which, in principle, is a blurring filter.

In summary, the total blurring in an image that limits our ability to see the small features, potential lesions, or other anatomic structures has three basic sources. During scanning, the amount of blurring is determined by the sampling aperture (regulated primarily by focal spot size and detector size). At the time of reconstruction, blurring is determined by the size of the voxel and the type of filter. We need to recognize the characteristics of blurring. If we introduce a large amount of blurring at one point, there is no way to counteract it by decreasing the blurring at other points in the process. A chain is no stronger than its weakest link.

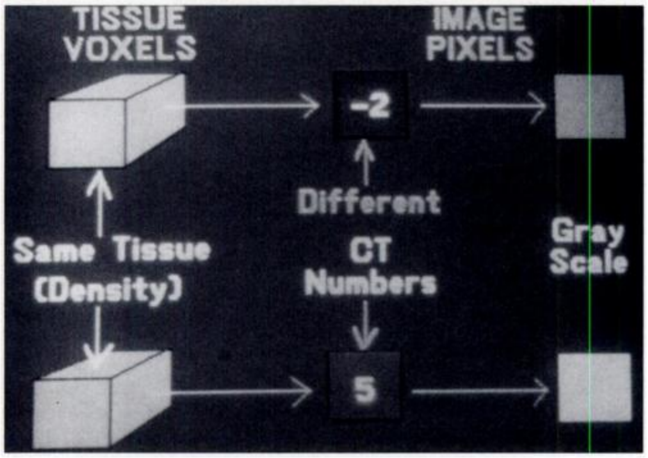

Figure 7. Diagram illustrates how noise results because the same tissue can produce different CT numbers. Note that the pixel in the lower right corner is brighter than the other one.

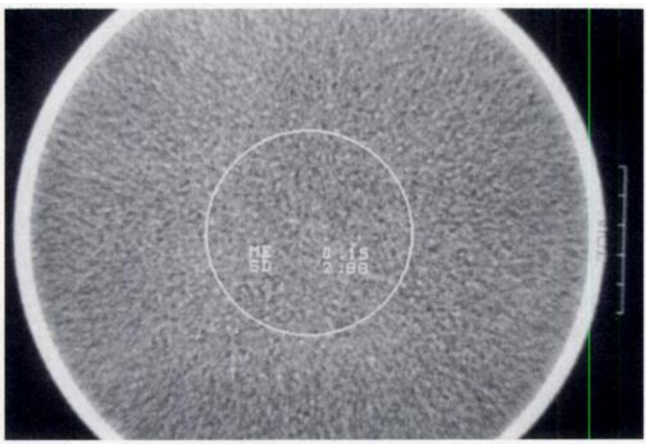

Figure 8. CT scan of a container of water in which a circular area of interest is indicated. Values for the mean (*ME*) and standard deviation (*SD*) of the CT numbers are shown.

Noise

Visual noise affects visibility of low-contrast objects. In an image with low noise, we can see more low-contrast lesions. As the image noise increases, fewer and fewer low-contrast objects are visible (Figs 5, 6).

Noise is caused by the variation in attenuation coefficients between individual voxels. In CT, we are measuring the attenuation coefficients of the individual tissue voxels. If we have two voxels of identical tissue, we would expect to measure identical attenuation coefficient values, and, when these are translated into CT numbers, we would expect to get similar numbers. In reality we do not. In Figure 7, two voxels of the same tissue produce different CT numbers. This is a statistical variation that is visualized as image noise. One way of evaluating noise is to produce a CT image of a container of water. Water is homogeneous and has a defined CT number of zero. However, a CT scan of water reveals variations in pixel brightness because of the noise (Fig 8).

Noise can be quantified. If we count and do a statistical analysis of the number of pixels within a region of interest and the various CT numbers, the resultant statistical distribution would most likely be a bell-shaped curve. The amount of spread or deviation can be evaluated by determining the standard deviation, which is the easiest and an excellent way of quantifying the amount of noise in CT images. All CT systems have the software capability of defining a region of interest and measuring and calculating the standard deviation.

To decrease noise, we must measure the attenuation coefficient values for each voxel more precisely. A more precise measurement can be made by increasing the number of x-ray photons absorbed in each voxel during the imaging process. The more photons in a voxel, the more precise the measurement will be, and this will reduce the CT number error that appears in the image as noise.

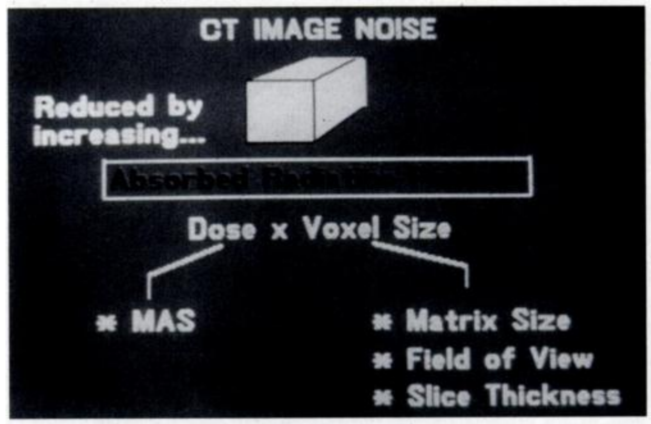

Figure 9. Diagram illustrates how CT image noise is reduced by increasing the quantity of x-ray photons absorbed by each tissue voxel. This is determined by the exposure (*MAS*) and the voxel size.

The quantity of radiation photons absorbed can be controlled in two ways: adjusting the size of the voxel and the exposure (Fig 9). Obviously, a large voxel will intercept and absorb more photons than a small voxel and will result in decreased noise in the image. The other way of controlling the quantity of absorbed radiation photons per voxel is by increasing the total radiation imparted to the patient. This is controlled through the milliampere seconds. By increasing the milliampere seconds, we deposit more photons in each voxel of tissue, which results in less statistical variation, therefore less noise in the image.

The mathematical filters selected in the image reconstruction process can also be used to control noise. Many CT systems have 10 or 12 filters with different characteristics. The smoothing filters are used to reduce noise. Unfortunately, as we decrease noise by using a smoothing filter, we increase the blurring and decrease visibility of detail. The so-called edge-enhancing or detail-enhancing filter has essentially an opposite characteristic. A filter of this type might be selected when the primary goal is visualization of small detail. CT equipment manufacturers generally recommend which filters are useful for specific clinical procedures.

Unfortunately, noise is a form of image detail. Anything we do to sharpen an image and increase detail, such as use of small voxels or edge-enhancing filters, will also increase visibility of noise.

CONCLUSION

We are concerned with the visibility of objects within the body, primarily lesions. As we consider lesions with respect to their size and inherent contrast, we have found that blurring of the image from several sources limits the visibility of the smaller objects and that visual noise limits visibility of the low-contrast objects. We can improve visibility of detail by using small voxels. However, when we use small voxels to push up the curtain of invisibility on the right and increase visibility of detail, the curtain will drop down on the left because those small voxels increase the noise (Fig 2). In controlling image quality, we must make compromises.

An optimized protocol for a specific clinical study must take these physical principles into consideration and be adjusted to give a proper balance among detail (visibility of small objects), low noise (visibility of low-contrast objects), and patient exposure.

SUGGESTED READINGS

Newton TH, Potts DG, eds. Technical aspects of computed tomography. St Louis: Mosby, 1981.

Sprawls P. The physical principles of medical imaging. Rockville, Md: Aspen, 1987.