# Chapter 7 : Contrast Resolution

I. Introduction

- A major task for all medical imaging systems is to detect lesions (i.e., a zone of tissue with impaired function as a result of damage by disease or wounding) of varying size and contrast from surrounding tissue. In diagnostic ultrasound, examples include the detection of tumors in the liver, myocardium infarction and breast masses. Therefore, contrast resolution, defined as the imaging system's ability to resolve objects with different amplitudes from the background, is a critical component in determining the system's imaging performance. It is often more important than spatial resolution in clinical situations.

- Based on Weber's law, human's ability to resolve two visual stimuli with different intensities, I and $I+\Delta I$, is determined by the ratio $\Delta I /I$ over a wide intensity range. In other words, a larger intensity difference (ΔI) is needed for a target to be detected in a brighter background of intensity I. However, our analysis in this chapter will ignore the effects of the background intensity level to the human visual system. This is due to the fact that the display level (related to I) and the display dynamic range (related to ΔI) can be arbitrarily set by the manufacturer and the user based on personal preferences. Therefore, we will only focus on the **system's** fundamental capabilities to detect objects with different size and contrast.
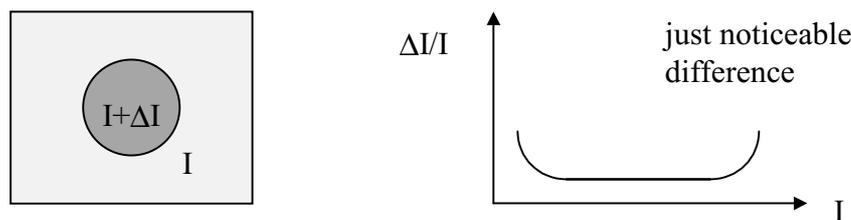

- In diagnostic ultrasound, it has been shown that one of the primary limitations on detecting low contrast lesions is image speckle resulting from coherent interference of scatterers (i.e., linear summation by the transducer of echoes from scatterers) in a finite resolution volume. Speckle formation and its statistical properties have been discussed previously. In this chapter, we will discuss the interaction between spatial and contrast resolution, the influence of speckle on contrast resolution. Finally, we will discuss how one can improve contrast resolution.

II. Contrast-Detail Analysis

- Intuitively, an imaging system's ability to detect objects of low contrast against the background is determined by the following three factors : contrast (i.e., mean brightness difference between the object and its surrounding background), size of the object and noise level. Therefore, it is straightforward to see that detectability can be improved with higher contrast, a bigger object or lower noise. However, to quantitatively characterize the effects of the three factors on detectability is by no means a trivial task.

- As mentioned previously, contrast resolution refers to the ability to resolve objects with different mean brightness. Unlike detectability, which is more or less a binary function, contrast resolution is related to the fundamental performance of an imaging system. Nevertheless, we will use the same approach to quantitatively characterize these two concepts and use these two terms interchangeably in this chapter.

- In order to study detectability of objects with varying size and contrast empirically, a "contrast-detail" phantom may be used to enable the measurement of threshold detection of simulated objects as a function of lesion size and contrast. A possible design of such a tissue-mimicking phantom is illustrated in the following figure. The phantom includes a series of cones and each cone contains a different tissue mimicking material so that a range of echo intensities is covered. Additionally, cross section scans (i.e., the plane perpendicular to the axes of the cones) at various positions along the lengths of the cones result in images of disks of a constant diameter within a plane, but varying in contrast.
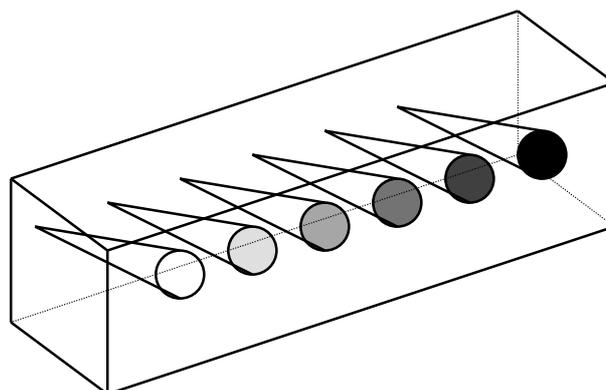
contrast -detail phantom
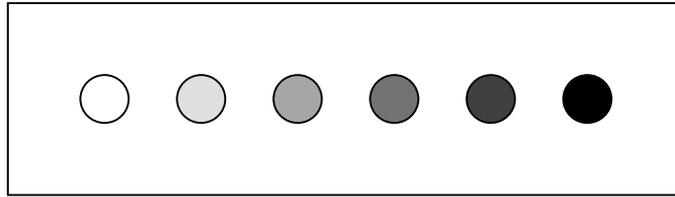
a cross section image

- Observer evaluation of such images enables the measurement of threshold detectability as a function of diameter and echo amplitude relative to the background. Viewers are instructed in advance as to the pattern of the objects and they are allowed to adjust the viewing position. The viewers are then asked to determine the presence of each disk in the ultrasound images and a criterion of 50/50 likelihood is used for detectability. By doing so, the detection threshold of objects of varying contrast and size can be determined.
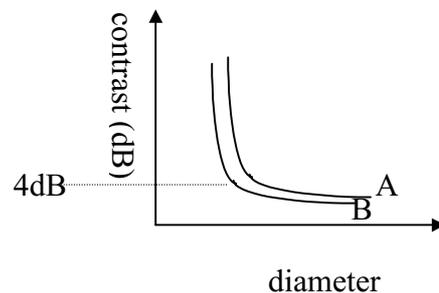
- The above figure shows typical results of the measurements of threshold detection for two different imaging conditions (e.g., different systems or different frequencies on the same system). Very roughly speaking, these curves are simply a shifted version of each other. In addition, the "knee point" (the breakpoint between two distinctly different regions) is all around 4dB. This observation is consistent with our previous analysis on speckle noise (~4.34 dB on a logarithmic scale). In other words, when the contrast is larger than the speckle noise, the detection threshold is mainly determined by the system's spatial resolution (relative to the system's point spread function). When the contrast is low, the detection threshold is strongly dependent on the contrast itself.

- In each region separated by the knee point, the curve can be approximated by the

following equation

$$Cd^{m} \approx \text{constant}$$

where C is the object contrast, d is the disk diameter and m is determined by curve fitting. Generally speaking, a curve towards the left (i.e., a smaller constant) represents a system with better contrast resolution.

III. Contrast-to-Noise Ratio

- Detectability of a lesion in a noisy background is determined by local contrast and the properties of the system. Local contrast, defined as the ratio of the intensity variation to the averaged background intensity, can be artificially enhanced, e.g., by re-mapping gray levels of the envelope. Noise characteristics, on the other hand, inherently limit detectability. To detect a finite lesion, statistics of averaged signals over the regions of interest (i.e., area-wise statistics instead of point-wise statistics) must be derived. More specifically, a contrast-to-noise ratio (CNR), defined as the ratio of the local intensity variation to the standard deviation of the averaged background intensity, is suitable for a quantitative measure of detectability.

- If *I(x',z)* denotes the intensity at a point *(x',z)* in the image plane, the averaged signal over a target is

$$I_{A} = \frac{1}{S} \int\int_{-\infty}^{\infty} W(x',z) I(x',z) \, dx'dz \,,$$

where $W(x',z)$ is a weighted function describing the target and

$$S = \int\int_{-\infty}^{\infty} W(x',z) \, dx'dz \,.$$

 In our analysis, $W(x',z)$ is chosen to be unity inside and zero outside so that the constant S simply represents the target area. Assuming the **point-wise** first order moment (i.e., statistical mean) of $I(x',z)$ in the target region is a constant $\langle I \rangle$, i.e., the target has a homogeneous distribution of scatterers, the mean value of the **area-wise** averaged signal I_{A} is also $\langle I \rangle$ according to the principles of superposition.

- The second order moment of I_A is

$$\langle I_A^2 \rangle = \frac{1}{S^2} \iiint \int_{-\infty}^{\infty} W(x_1',z_1) W(x_2',z_2) \langle I(x_1',z_1) I(x_2',z_2) \rangle dx_1' dx_2' dz_1 dz_2,$$

where $\langle I(x_1',z_1) I(x_2',z_2) \rangle$ represents the autocorrelation function of the intensity. Assuming the speckle pattern is spatially stationary, we have

$$\langle I_A^2 \rangle = \frac{1}{S^2} \iiint \int_{-\infty}^{\infty} W(x_1',z_1) W(x_2',z_2) R_I(x_1'-x_2', z_1-z_2) dx_1' dx_2' dz_1 dz_2$$
$$= \frac{1}{S^2} \iint_{-\infty}^{\infty} W(x_1',z_1) [W(x_1',z_1) ** R_I(x_1',z_1)] dx_1' dz_1$$

where $**$ denotes two-dimensional convolution and $R_I(x_1'-x_2', z_1-z_2)$ is the autocorrelation function evaluated at $(x_1'-x_2', z_1-z_2)$. The above equation can be re-written as

$$\langle I_A^2 \rangle = \frac{1}{S^2} [W(-x',-z) ** R_I(x',z) ** W(x',z)]\big|_{x'=0, z=0}.$$

Since $\sigma_{I_A}^2 = \langle I_A^2 \rangle - \langle I_A \rangle^2$,

$$\sigma_{I_A}^2 = \frac{1}{S^2} [W(-x',-z) ** C_I(x',z) ** W(x',z)]\big|_{x'=0, z=0},$$

where $C_I(x',z) \equiv R_I(x',z) - \langle I \rangle^2 = R_I(x',z) - \langle I_A \rangle^2$ is the autocovariance function of $I(x',z)$.

- Assuming the target area is much larger than the area where $C_I(x',z)$ is appreciable,
$$\sigma_{I_A}^2 \approx \frac{1}{S^2} [W(-x',-z) ** W(x',z)]\big|_{x'=0, z=0} \iint_{=\infty}^{\infty} C_I(x',z) dx' dz.$$

Since $W(x',z)$ is unity inside the target area and zero outside, the above equation reduces to

$$\sigma_{I_A}^2 = \frac{1}{S} \iint_{=\infty}^{\infty} C_I(x',z) dx' dz.$$

- Defining the normalized correlation cell area S_c of the target as

$$S_c \equiv \iint_{-\infty}^{\infty} \frac{C_I(x',z)}{C_I(0,0)} dx'dz \,,$$

the standard deviation of the averaged signal, I_A, is

$$\sigma_{I_A} = \frac{\sigma_I}{N^{1/2}} \,,$$

where $\sigma_I^2 = C(0,0)$ is the variance of $I(x',z)$ and N represents the number of independent speckle spots in the target region defined by

$$N = \frac{S}{S_c} \,.$$

Note that N is directly related to spatial resolution.

- For the above analysis, it is clear that the standard deviation of the area-wise signal (i.e., I_A) is determined by both the standard deviation of the point-wise signal (i.e., I) and the number of speckle correlation cells. Therefore, defining ΔI as the local intensity variation (i.e., $\Delta I \equiv I_1 - I_2$), then the CNR is

$$CNR = \frac{\langle \Delta I \rangle}{\sigma_{I_A}} = \frac{\langle \Delta I \rangle}{\sigma_I} N^{1/2} \,,$$

where I_1 and I_2 are mean intensities of the target and the background respectively. Note that the speckle cell size, S_c is independent of the mean intensity of the signal.
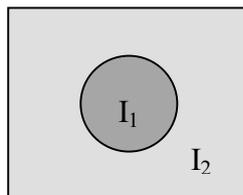
- As discussed previously, intensities of fully developed speckle have exponential

statistics and speckle fluctuations can be viewed as additive noise on a logarithmic display, which is normally the case in most ultrasonic imaging systems. On a logarithmic scale, the intensity of each pixel (D) and the original intensity (I) have the following relation

$$D(dB) = f(I) \equiv 10\log_{10}(\frac{I}{I_0}),$$

where I_0 is an arbitrary reference intensity. Therefore, the brightness variation between the target and the background on a logarithmic display is

$$\Delta D = 10\log_{10}(\frac{I_1}{I_0}) - 10\log_{10}(\frac{I_2}{I_0}) = 10\log_{10}(\frac{I_1}{I_2}).$$

Using σ_D to denote the standard deviation of the display intensity on a logarithmic scale, we have

$$CNR = \frac{\Delta D}{\sigma_D}N^{1/2} = \frac{10\log_{10}(\frac{I_1}{I_2})}{\sigma_D}N^{1/2}.$$

Finally, recall that the speckle variation on a logarithmic scale is approximately 4.34dB, we obtain

$$CNR = \frac{10\log_{10}(\frac{I_1}{I_2})}{4.34}N^{1/2}.$$

In other words, the detectability of a lesion in a speckle beakground is determined by three factors : the ratio of mean intensities between the target and the background, the total number of speckle correlation cells in the target, and speckle variations. These three factors are consistent with our intuition described previously.

IV. Contrast Resolution Enhancement

- The definition of CNR provides a quantitative measure of contrast resolution. It can also be used to gain insights of factors affecting detectability. For example, it shows that contrast resolution is determined by the local contrast, noise level and

the size of the target relative to the system's spatial resolution.

- Although local contrast is determined by inherent properties of the object, the actual contrast displayed by the system (i.e., ΔD) can be arbitrarily varied. For example, a simple linear mapping function $f(D) = kD$, where k is a constant, increases the display contrast by a factor of k. In a noisy environment, however, the noise is also increased. For more complex mapping functions (assuming monotonic), linear approximation is often valid in the context of detecting low contrast lesions. Therefore, detectability (or CNR) can not be improved by re-mapping since the speckle noise is also increased by the same amount.

- Contrast measured by the system is fundamentally limited by the differences in reflectivity between the region of interest and the surrounding tissue. In practice, the measured contrast is even lower than the true contrast due to imperfection in beam formation. Sidelobes, resulting from a finite aperture size, receive signals from directions outside of the main beam direction and therefore may decrease the contrast assuming the signal level in the main beam direction is lower than the background. There are other sources of sidelobes, such as beam formation errors, contributing to degradation of contrast resolution via a similar mechanism. Therefore, lower sidelobe levels often produce higher contrast resolution and this explains why non-uniform apodization, instead of uniform apodization, is often preferred.

- Contrast resolution can be enhanced by decreasing speckle variations (i.e., σ_D). Since speckle is due to coherent interference from scatterers within a sample volume, speckle variations can only be reduced by processing **in-coherently** (i.e., discarding phase information). For example, speckle variations can be reduced by low pass filtering the **post-detection** data, since phase information has been discarded at this point and any processing done thereafter is in-coherent. Speckle variations cannot be reduced by low pass filtering the **pre-detection** data since linear combinations of phase sensitive data do not change the statistical properties of the resulting signals. Note that although speckle noise can be reduced by low pass filtering the post-detection data, contrast resolution enhancement is usually gained at the price of spatial resolution and edge definition.

- Compounding techniques (another type of in-coherent averaging) have also been proposed for speckle reduction by introducing in-coherence during image formation. Common compounding techniques include spatial compounding,

frequency compounding and temporal compounding. These techniques exploit the fact that the ensemble average of a speckle image is the same as the in-coherent average of the original object. Hence, speckle variations are reduced by in-coherently averaging partially correlated measurements without affecting the original intensity contrast. Generally speaking, partially correlated images can be obtained in spatial, frequency or temporal domain.

- The following figure illustrates the basic principles of spatial compounding with lateral array displacements.
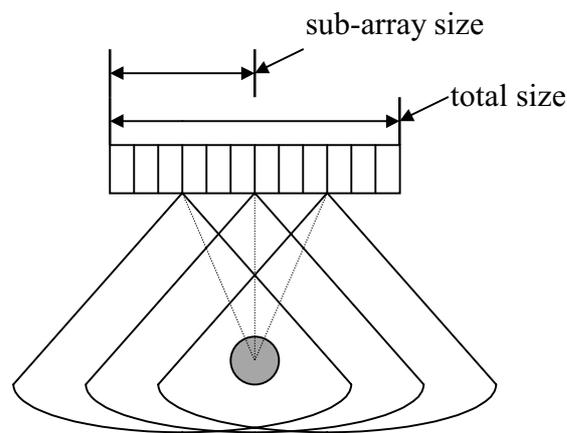

 Since the distance between individual scatterers and the observer (i.e., array center) varies as the sub-array displaces laterally, the sub-array images are partially correlated (or even totally un-correlated, depending on the displacement). Therefore, by averaging the post-detection images, speckle variations can be reduced. In the case where N individual images are totally un-correlated to each other, speckle variations are reduced by a factor of $1/\sqrt{N}$. Apparently, lateral resolution is degraded due to the aperture size reduction. Edge definition, however, is the same or even better than the image using the full aperture. (The relation between the spatial correlation and the lateral displacement is determined by the van Cittert-Zernike theorem. Please refer to the literature of statistical optics if interested.)

- Spatial compounding can also be performed in the elevational direction with two-dimensional arrays. In other words, partially correlated images can be acquired by steering the image plane elevationally. The elevational steering angle determines the correlation between images and therefore the effectiveness of speckle reduction. Such a compounding scheme is particularly of interest if the array is coarsely sampled in elevation in order to reduce the total channel count. Since

only small steering angles are required, the grating lobes artifacts are not critical.

● Partially correlated images can also be obtained in the frequency domain. Consider the following spectrum,
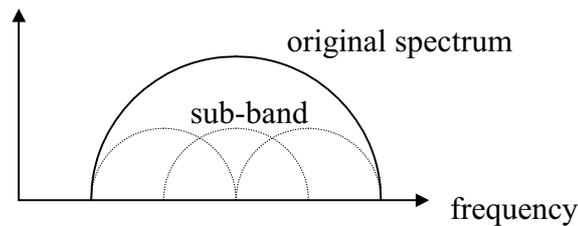
frequency compounding is based on the fact that a speckle pattern changes when the frequency of sound waves changes. Therefore, in frequency compounding, a pulse of a specific bandwidth is passed through a filter bank which divides the pulse into a number of sub-bands. The spectra of the various sub-bands can overlap. The correlation between images from different sub-bands, however, increases as the overlap increases. The compounded signal intensity is then obtained by summing the intensities of the individual sub-bands. Similar to spatial compounding, speckle variations are reduced by a factor of $1/\sqrt{N}$ when there are N sub-bands with no overlap. Since the axial resolution is determined by the pulse width and the pulse width is in turn proportional to the reciprocal of the bandwidth, axial resolution is degraded with frequency compounding.

● Temporal compounding is a form of multi-rate signal processing as the final pixel intensity is made up of the intensity samples which were sampled initially at a higher rate. Similar to frequency compounding, the correlation between samples and the potential reduction in speckle variations are determined by the sampling interval. In general, speckle reduction is also gained at the price of axial resolution.

● Another way to enhance contrast resolution is to increase the number of independent speckle spots (i.e., correlation cells) in the region of interest. Since the size of the object is fixed, this can only be achieved by decreasing the size of speckle spot. In other words, by improving the spatial resolution, there are more independent samples within the object available to the observer (either the imaging system or the human visual system) such that speckle variations can be

smoothed out by additional processing. Given a transducer size and a carrier frequency, it is usually impossible to achieve optimal spatial and contrast resolution simultaneously due to the interaction between mainlobe and sidelobes. Therefore, the only way to improve both spatial and contrast resolution at the same time is to change the fundamental diffraction limits by increasing the aperture size and using a higher carrier frequency.