Minicomputer Enhancement of Scintillation Camera Images Using Fast Fourier Transform Techniques

Mark J. Sebern, James D. Horgan, Robert C. Meade, Charles M. Kronenwetter, Philip P. Ruetz, and En-Lin Yeh

Veterans Administration Center, Wood, Wisconsin, Marquette University and the Medical College of Wisconsin, Milwaukee, Wisconsin

Minicomputer methods were developed to enhance lesions in scintillation camera images. This study was directed towards improving the diagnostic quality of liver images. A PDP-12 digital computer was interfaced to a Pho/Gamma HP III scintillation camera and programmed to carry out two-dimensional frequency-domain analysis and processing as an on-line operation. A two-dimensional fast Fourier transform (FFT) is generated, and a composite one-dimensional frequency spectrum is produced. An interactive program allows the operator to construct graphically a frequency-domain filter and apply it to the data matrix. The filter is optimized using the image of a known phantom and then applied unchanged to the clinical liver image. An inverse Fourier transform produces an enhanced image in the spatial domain. Significant enhancement of both phantom and liver images has been obtained.

This study sought to develop on-line minicomputer methods for improving the diagnostic quality of liver images produced by scintillation camera systems. In particular, it was desired to develop techniques that could be used in any laboratory operating a small computer in conjunction with a scintillation camera.

Brown et al (*1*) investigated resolution enhancement of rectilinear scans by using a large-scale digital computer to transform the data from the spatial to the frequency domain. They used two-dimensional fast Fourier transform (FFT) techniques and filter function, selected to be the reciprocal of the modulation transfer function at low spatial frequencies and a decreasing polynomial function at higher frequencies. Lotter et al (*2*) explored the use of a similar filter for the enhancement of scintiscans. Inouye et al (*3*) developed an alternative method using a large-scale digital computer to process the scintillation camera images: it employs nonlinear filtering with a square-root transform to remove random fluctuations without affecting the spatial resolution. Bal-

lard and Sklansky (*4*) described a method that enhances the edges of the tumor images in radiographs and radionuclide scans. It employs a large-scale digital computer with a core storage of approximately 100 thousand 36-bit words.

Brown et al (*5–7*), after investigating a number of other enhancement techniques, abandoned classic or logical means of resolution in favor of an empirical method: adaptive filtering. Higher-frequency transform coefficients are set to zero if their amplitude is below a threshold determined by the average of the high-frequency coefficients. An arbitrarily selected logarithmic filter function is used for the low spatial frequencies. This concept of empirically selecting the frequency-domain filter function is also used in the system to be described. The selection, however, uses a known phantom to optimize the
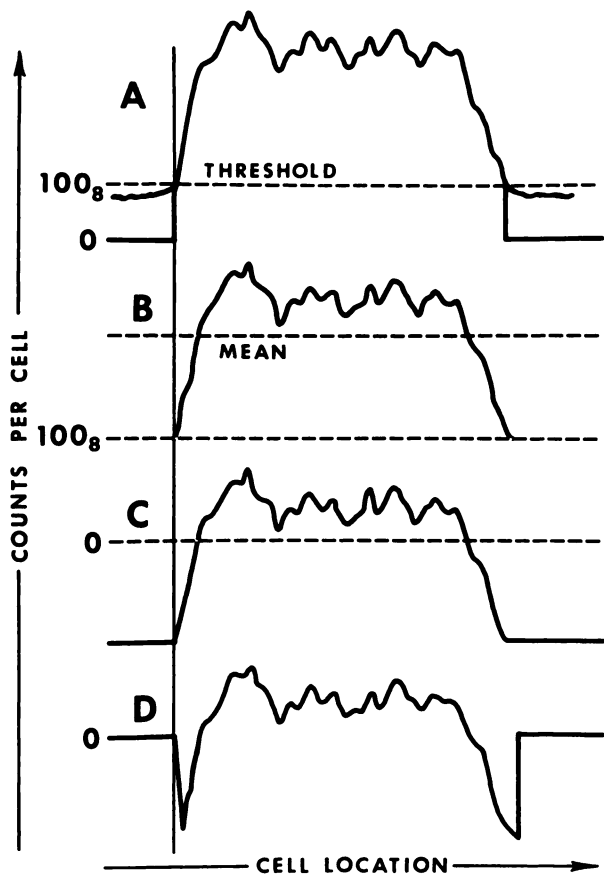
FIG. 1. (A) Typical profile through data matrix of liver. Matrix points below threshold (100_8) set equal to zero. (B) Mean of all cell counts above 100_8 is calculated. (C) Mean is subtracted from all cell counts. (D) Matrix points set to zero in Step A are again set to zero.

These steep edges cause relatively large-amplitude spectral lines to appear in the frequency domain in approximately the same range as those produced by organ defects. Figure 1A shows a typical profile through a liver image; Fig. 2A shows the corresponding one-dimensional frequency spectrum. The relatively large amplitudes at the lower spatial frequencies are due mainly to the steep edges of the image. To minimize this effect, the entire spatial image is examined with respect to a threshold equal to 100_8 (i.e., 64) counts, the minimum value necessary to eliminate background. All cell counts below this threshold are set equal to zero (Fig. 1A). The mean value of the remaining nonzero points in the entire data matrix is then calculated (Fig. 1B) and subtracted from each of the data points (Fig. 1C). In general, this mean is not equal to the mean of a particular profile. After subtraction, the points outside the image, which had been set to zero because they were below the threshold, are again set to zero. Figure 1D shows the profile of Fig. 1A after such preprocessing, and Fig. 2B shows the corresponding one-dimensional frequency spectrum. The higher-frequency components are unchanged by this proc-

filter. This filter is then applied unchanged to the clinical liver image.

MATERIALS AND METHODS

The subjects in the human studies were normal volunteers or patients referred for routine liver scans. Approximately 3 mCi of ^{99m}Tc-sulfur colloid is injected at least 20 min before data are recorded. Using a analog-to-digital converter to connect the Pho/Gamma HP III scintillation camera and the PDP-12 computer (programmed in assembly language), the liver image (approximately 500,000 counts) is stored on a magnetic disk as a 64 × 64 data matrix. A second matrix, derived from the system response to a uniform planar source (flood), is also stored.

Before transforming the data matrix to the frequency domain, two important preprocessing steps are executed in the spatial domain. The first is flood correction, which compensates for the nonuniform response of the camera. The second step is designed to reduce the effects of the steep edges of the liver image on the spectrum in the frequency domain.
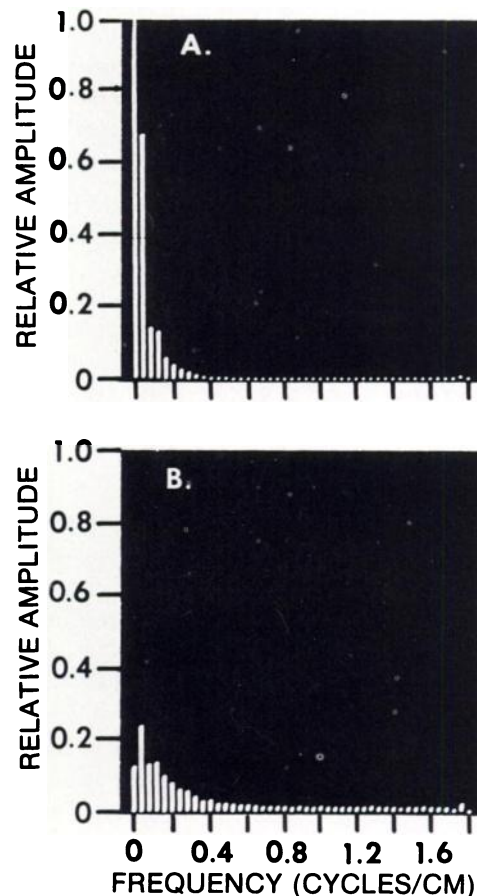

FIG. 2. (A) One-dimensional frequency spectrum corresponding to liver of Fig. 1A before preprocessing. Increments of frequency are 0.04 cycles/cm. (B) Spectrum of same liver after preprocessing.

essing, but the lower-frequency components are much reduced in amplitude.

After preprocessing in the spatial domain, the two-dimensional Fourier transform of the liver image is generated, using FFT techniques based on the procedures of Cooley and Tukey (8–10). The processing time for this step is about 30 sec. In the Fourier transform process, the liver image in the spatial domain is resolved into a large number of sinusoidal surfaces, each with a particular frequency, orientation, and amplitude. By averaging the amplitudes of all components of the same frequency, a one-dimensional frequency spectrum, showing the principal frequency components of the liver-image data, is generated.

This representation is considerably simpler than the two-dimensional spectrum and is particularly useful as an aid in selecting an appropriate digital filter. With the one-dimensional spectrum displayed on the cathode-ray tube, the operator, using a control lever on the computer console, can graphically construct a one-dimensional frequency response function for the filter, superimposing it on the one-dimensional spectrum (Fig. 3). This filter function is then applied to the data matrix to obtain the product of the amplitude of the frequency spectrum and the filter value, and an inverse Fourier transform is carried out to present the enhanced liver image to the operator. This step also requires approximately 30 sec of computer time.

The nature of the particular filter function that most effectively enhances the image will depend on the system being used and on the particular organ being studied. For processing liver images the function will represent a band-pass filter that attenuates the low-frequency components identified with the gross structure of the liver, emphasizes the intermediate frequencies associated with the liver defects, and attenuates the high-frequency components associated with random variations in the data. The exact shape of the optimum filter is best chosen with the aid of an appropriate phantom with precisely known features. The operator can then select a filter that gives the best delineation of "hot" or "cold" spots, as indicated by numerical profile studies as well as by visual inspection of the processed image. This filter is then applied unchanged to the clinical liver images. The filter function shown in Fig. 3, used to produce the results described below, was selected in this manner.

RESULTS

A schematic drawing of the absorption phantom studies is shown in Fig. 4. Circular absorption regions were constructed with thin layers of brass.
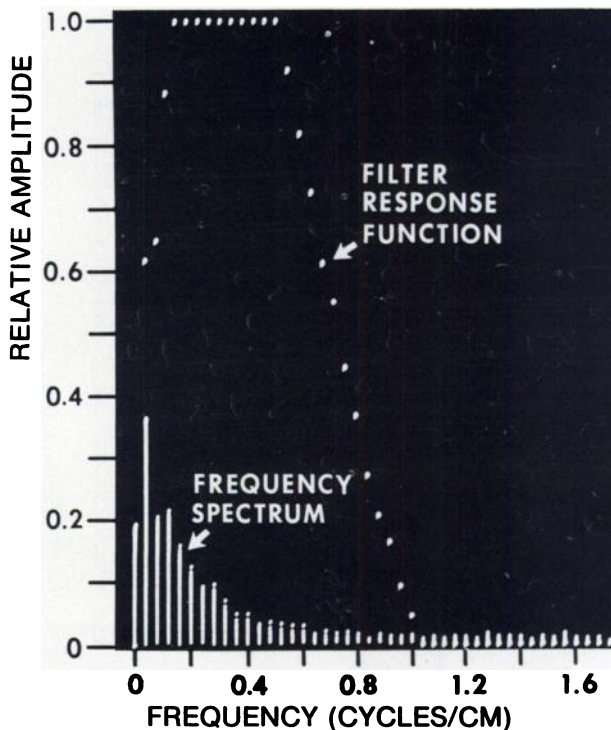

FIG. 3. One-dimensional preprocessed frequency spectrum with typical filter response function superimposed. Increments of frequency are 0.04 cycles/cm.
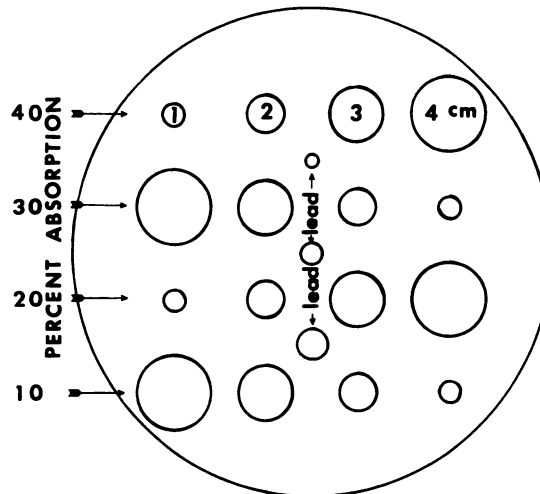

FIG. 4. Schematic drawing of absorption phantom.

They range over 1–4 cm in diameter and absorb 10–40% of the 140-keV gamma photons of ^{99m}Tc. The original triple-lens scintiscan of the phantom is shown in Fig. 5A and the computer-enhanced image is shown in Fig. 5B. The smaller 10% absorption areas, barely discernible on the scintiscan, appear clearly on the enhanced image.

Figure 6 shows the profiles through the 10% absorption area of Fig. 4. The areas of absorption are poorly identified in the profile of preprocessed data
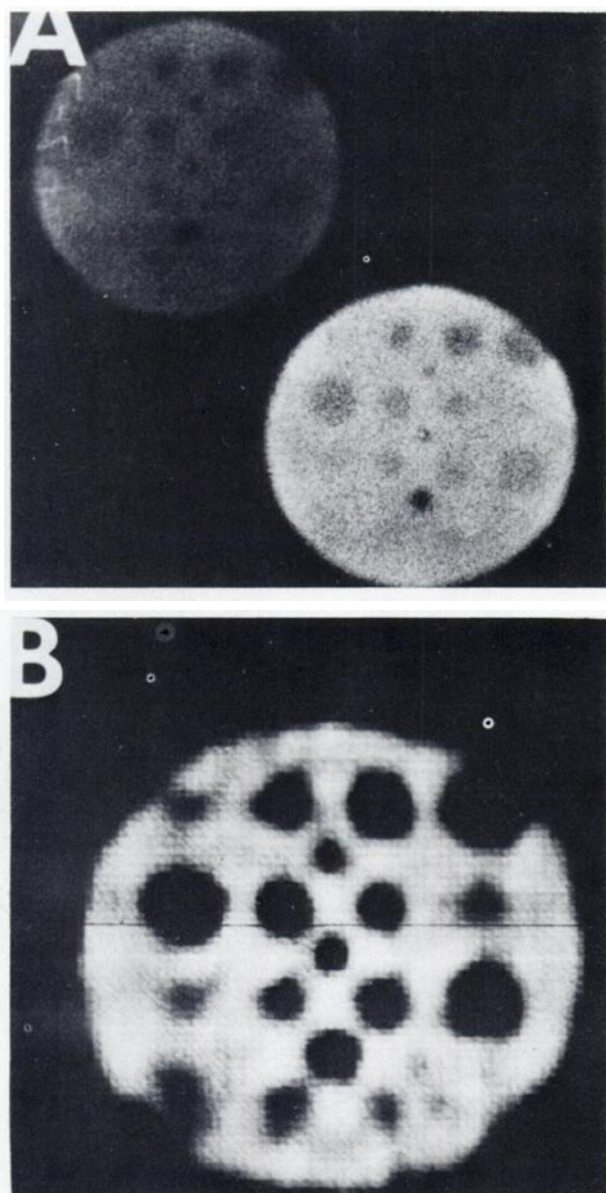
FIG. 5. (A) Triple-lens scintiscan of absorption phantom. (B) Computer-enhanced image.

liver was found to be very nodular with purplish lesions which were not biopsied. The computer-enhanced image (Fig. 9B) shows a larger number of easily detected defects. It also shows a defect in the area of the porta hepatis.

DISCUSSION

Some of the early attempts to enhance scintillation images involved data smoothing by averaging in the space domain. Such techniques are equivalent to using a low-pass filter in the frequency domain. The high-frequency random variations are reduced, but the variations associated with the liver defects are
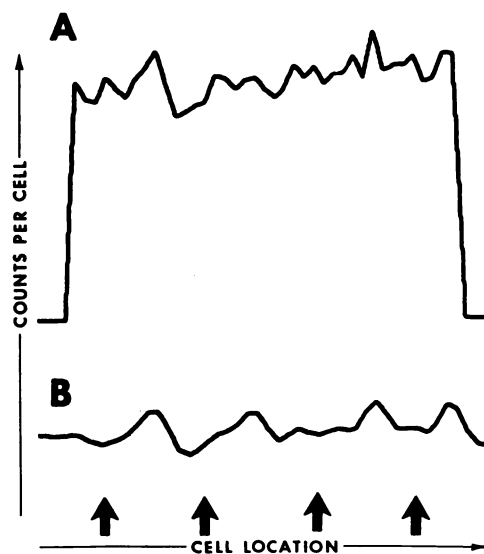

FIG. 6. Profiles through region of 10% absorption in Fig. 4. (A) Original data; (B) computer-enhanced data.
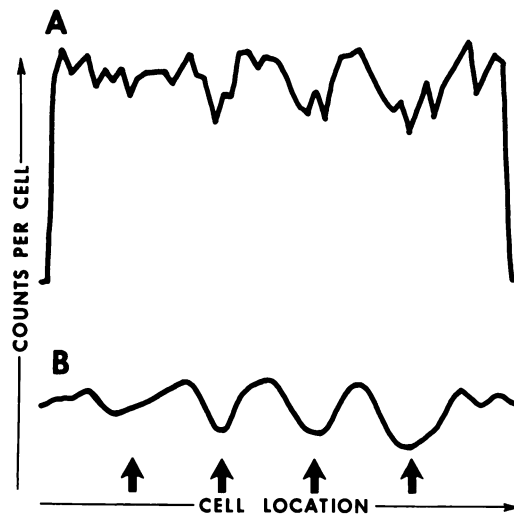
(Fig. 6A). However, the four low-absorption areas are clearly visible on the profile of the enhanced image (Fig. 6B). Figure 7 shows the profiles through the 20% absorption area. Although the preprocessed data identify the areas of absorption (Fig. 7A), they are more clearly identified on the profile of the filtered data (Fig. 7B).

Figure 8A is a flood-corrected scan of a normal liver. The computer-enhanced image (Fig. 8B) appears less homogeneous than the original scan. Anatomic landmarks, such as the interlobar fissure and gallbladder bed, are more clearly defined. Figure 9A is the flood-corrected scan of a patient with transitional-cell carcinoma of the bladder. At surgery, the

FIG. 7. Profiles through region of 20% absorption in Fig. 4. (A) Original data; (B) computer-enhanced data.
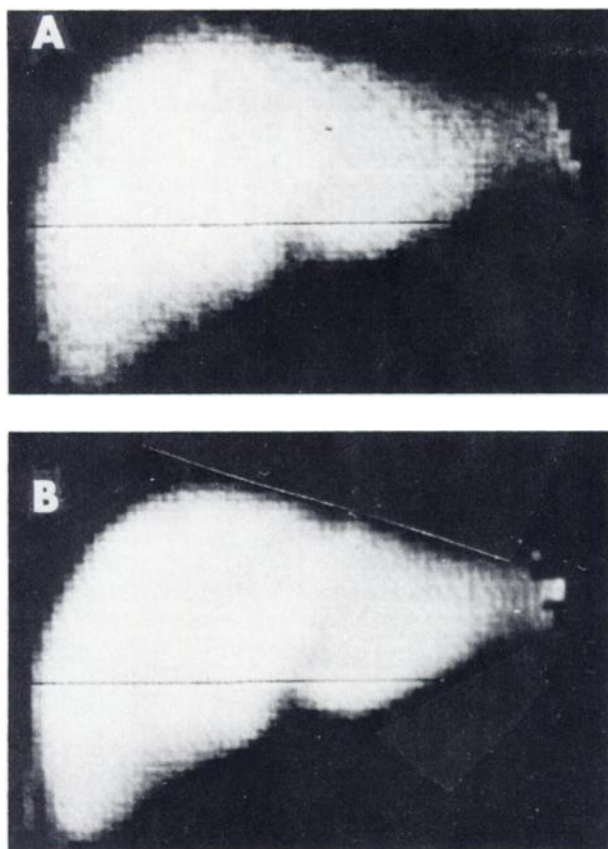
FIG. 8. (A) Scintiscan of normal liver. (B) Computer-enhanced image.
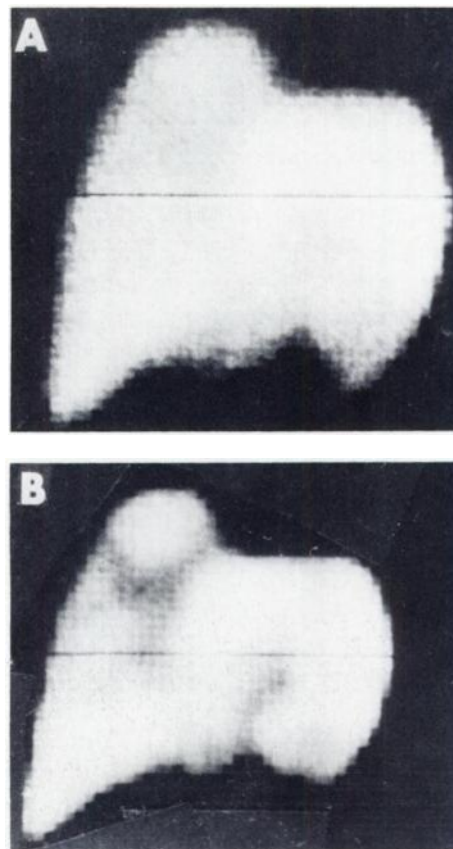

FIG. 9. (A) Scintiscan of abnormal liver. (B) Computer-enhanced image.

not amplified. In fact, they may themselves be reduced. This technique was later improved by combining data smoothing and uniform background subtraction to achieve noise reduction and contrast enhancement.

Other attempts have been made to emphasize defects by compensating for the poor response of the system at higher frequencies. A straightforward application of direct numerical deconvolution techniques was not very successful because the random variations were overemphasized. Fourier transform techniques make it possible to transform the data to the frequency domain and to utilize filters which emphasize a middle range of frequencies associated with variations around defects but which attenuate both the lower frequencies (associated with the gross structure of the organ) and the higher frequencies (associated with noise). Although the straightforward digital computation of the Fourier transform is relatively slow and inefficient, the introduction of FFT techniques permits the practical implementation of the process on a small laboratory digital computer (8). Execution times for the process are compatible with the time required for the collection of the data.

The characteristics of the appropriate band-pass filter depend on the particular camera system used. For this reason, the present method permits the filter to be adjusted to yield best results for a specific system; phantom studies are useful in making such an adjustment. Once an optimal filter has been developed for liver images, it can be used for processing all liver scans.

An important feature of the present system is the suppression of the effects of the steep edges of the liver image on the frequency spectrum. Because of the limited numerical precision available with small laboratory computers (three or four decimal digits), the small features of interest tend to be de-emphasized in comparison with these steep edges. The preprocessing helps to emphasize these features by subtracting a step function. A discontinuity at the edge of the image is introduced, but the magnitudes of the associated spectral components are less than before preprocessing. Thus, the frequency components due to the defects are more likely to be apparent.

Several enhanced scans have revealed focal defects that were poorly visualized or appeared insignificant in the routine camera studies. Nevertheless, the significance of this work in improving the diagnostic

quality of liver images can only be determined after study of many clinical cases. In our laboratory the basis for clinical interpretation continues to be the conventional liver image, although a number of positive interpretations have resulted from the enhanced images.

The phantom study (Fig. 7B) shows a few small areas of reduced counts, which might be interpreted as extra defects. These apparent artifacts are currently under study. The trade-off between true positives and false positives must be considered in any further study of diagnostic accuracy. New criteria for scan interpretation may need to be developed.

ACKNOWLEDGMENT

REFERENCES

1. BROWN DW, KIRCH DL, RYERSON TW, et al: Computer processing of scans using Fourier and other transformations. J Nucl Med 12: 287–291, 1971

2. LOTTER MG, MINNAAR PC, VERSTER F, et al: Digital data processing of images. S Afr Med J 48: 986–991, 1974

3. INOUYE T, SAWADA N, KASHIO E, et al: Computer methods for processing scintillation camera images. J Nucl Med 14: 411, 1973

4. BALLARD D, SKLANSKY J: Tumor detection in radiographs. Comput Biomed Res 6: 299–321, 1973

5. CYKOWSKI CB, KIRCH DL, POLHEMUS CE, et al: Image enhancement of radionuclide scans by optical spatial filtering. J Nucl Med 12: 85–87, 1971

6. KIRCH DL, BROWN DW, STERN DM: Nonlinear digital techniques for enhancement of radionuclide images. J Nucl Med 13: 441, 1972

7. BROWN DW, KIRCH DL, TROW RS, et al: Quantification of the radionuclide image. Semin Nucl Med 3: 311–325, 1973

8. SEBERN MJ: Minicomputer Enhancement of Gamma Camera Images. Doctoral dissertation, Marquette University, April, 1974

9. COOLEY JW, TUKEY JW: An algorithm for the machine calculation of complex Fourier series. Math Comput 19: 297–301, 1965

10. OPPENHEIM AV, SCHAFER RW: Digital Signal Processing. Englewood Cliffs, N.J., Prentice-Hall, 1975, pp 284–328