Enhancement of SPECT Images by Fourier Filtering the Projection Image Set

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Tomographic images from rotating gamma camera systems are often difficult to interpret because of poor contrast and high noise levels. A method is presented for improving the quality of these images by Fourier filtering the projection image set prior to reconstruction. A two-dimensional circularly symmetric Gaussian function is used as the spatial frequency filter. This filter can be optimized to enhance contrast and suppress noise in the projection image set in a straightforward and simple manner from the power spectra of representative projections. Preprocessing of the projections makes it possible to use a ramp reconstruction filter. The resulting tomographic sections show a dramatic improvement in image quality.

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Rotating scintillation cameras are now widely available and are being used in many institutions to acquire single photon emission computed tomographic (SPECT) studies. There are several advantages to these systems (*1*). Besides being able to provide conventional planar images, multiple tomographic sections are available from a single study. This allows the generation of sagittal, coronal, or any oblique angle set. The major problems associated with these systems are the constraints imposed by the collimators (*2,3*). Resolution in the reconstructed image is limited both by the collimator resolution and by the reconstruction filter required to yield an acceptable noise level. This is illustated in Fig. 1, which shows one section from an iodine-123 iodoamphetamine ([^{123}I]IMP) brain study reconstructed by several commonly used reconstruction filters (*4*). Although the image produced by using the ramp filter has the best contrast, the statistical variations are great, making diagnostic interpretation difficult. These variations can be reduced by using reconstruction filters which roll off at high spatial frequency, but at a loss of contrast.

A solution to the above problem is to apply a filter to

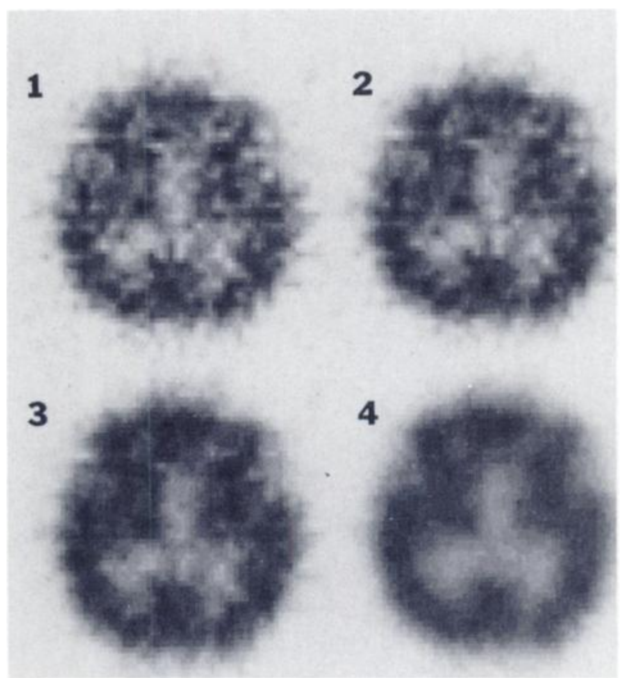

FIGURE 1
Example of IMP brain section as produced from four different reconstruction filters. Numbers on images correspond to reconstruction filter used. (1) Ramp; (2) Shepp-Logan; (3) Hanning; (4) Modified Shepp-Logan. See Ref. *4* for details on filters

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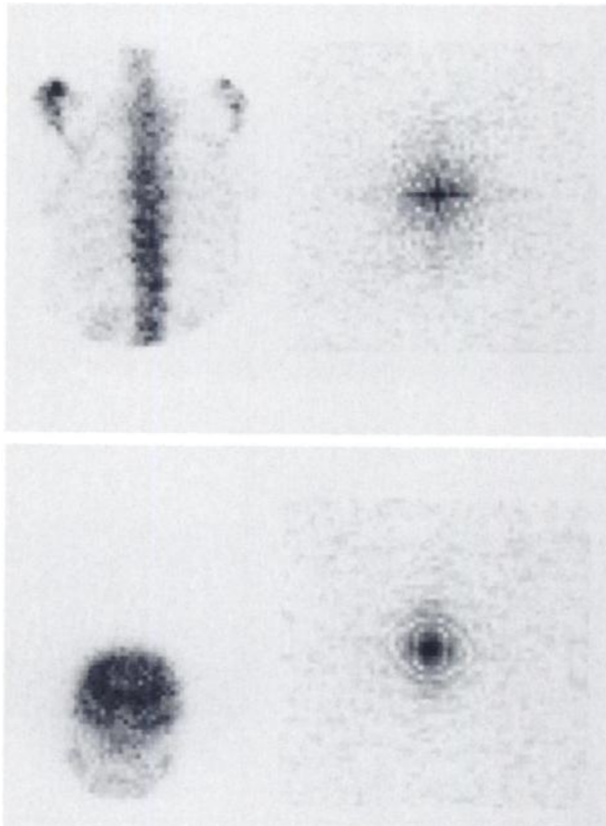

FIGURE 2
Images of [^{123}I]IMP brain and [^{99m}Tc]MDP bone studies and their respective power spectra. Both images were acquired under similar conditions and have same total counts

the projection images prior to reconstruction to enhance contrast and suppress the Poisson noise. With this processing, recontructions can be performed using a ramp filter to yield high quality SPECT without amplifying image noise.

THEORY

The problem is to find a filter which will improve the quality of the projection images as discussed above. Recently, King et al. have published papers on related work in which Wiener and Metz filters were used (*5,6*). These filters incorporate the inverse of an approximate system modulation transfer function (MTF) in order to recover resolution losses along with a high frequency roll-off to suppress noise. The roll-off of the Wiener filter depends on the noise and object power spectra, while that of the Metz filter depends on total image counts. The success of these filters is demonstrated in the substantial improvement seen in the SPECT images. However, the design of these filters is somewhat arbitrary since no one response function can fully describe the collimator resolution when sources are distributed over large volumes (*6*). In addition, the Metz filter dependence on total image counts is not necessarily sufficient in order to judge where the filter should go to zero. The spatial frequencies of the object must also be considered as they are in the Wiener filter. This is illustrated in Fig. 2, which shows images of an IMP brain and a technetium-99m methylene diphosphonate ([^{99m}Tc]MDP) bone and their corresponding power spectra. These images all have approximately the same total counts, but the power spectra are significantly different. Information in the bone image extends out to higher spatial frequencies. Thus a filter designed for the bone study applied to the IMP brain would include a significant noise component, while a filter specifically designed for the brain would compromise useful information.

Instead of relying on resolution recovery filters, we have investigated alternate band pass spatial frequency filters which do not require knowledge of the system MTF. The particular filter studied is a two-dimensional circularly symmetric truncated Gaussian function. The

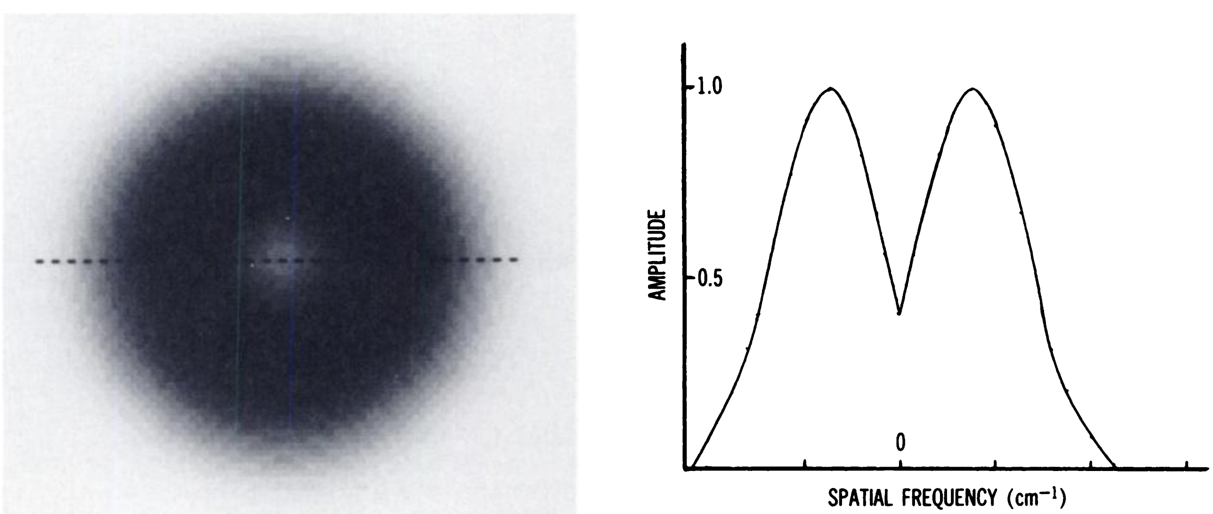

FIGURE 3
Image of typical two-dimensional Gaussian spatial frequency filter. Graph on right is plot through axis of symmetry

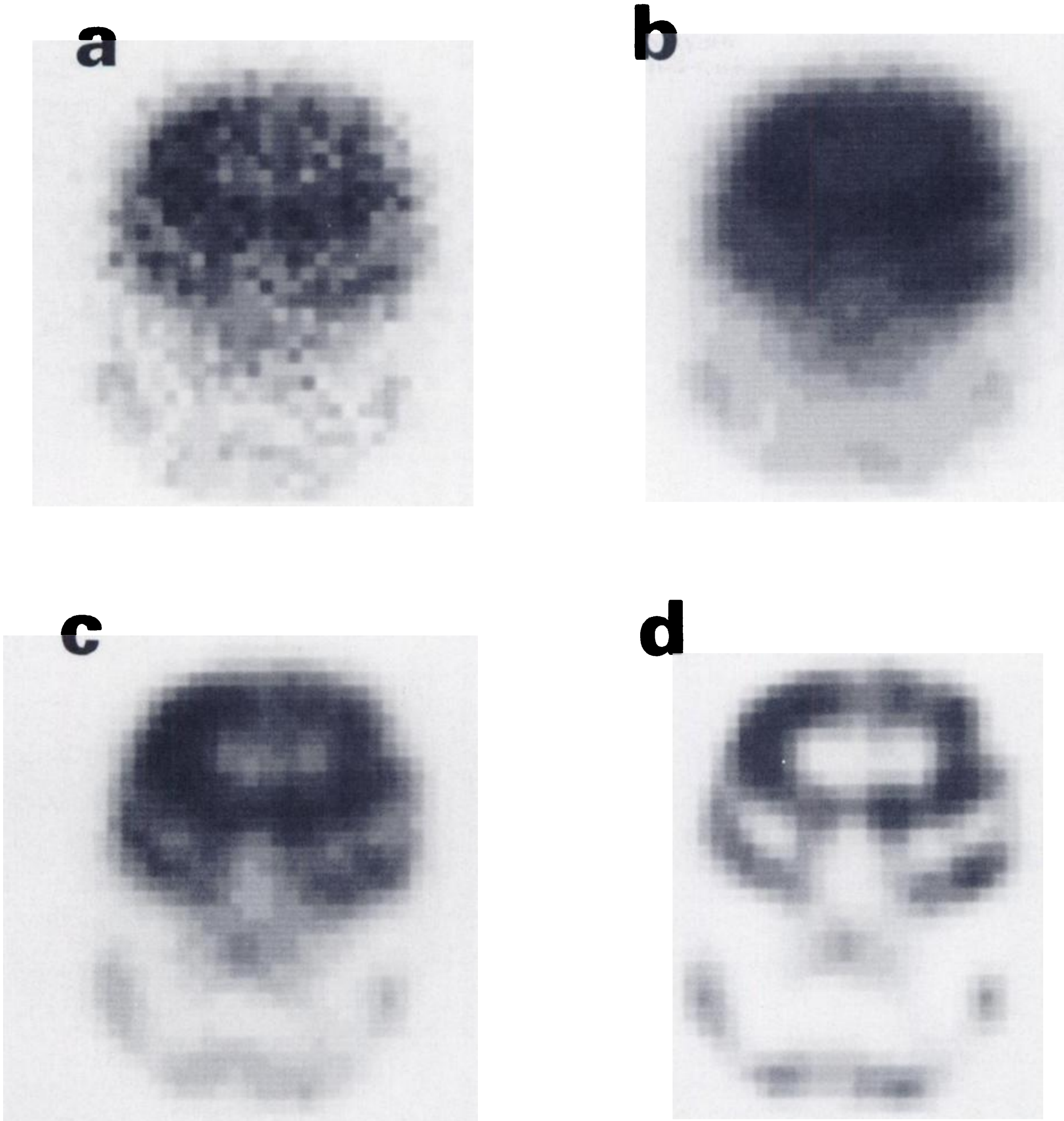

FIGURE 4
Examples of filtered projection image. Image (a) is raw [^{123}I]IMP brain projection image. Image (b) has had high frequency noise suppressed, but no contrast enhancement. Image (c) has been processed by Gaussian filter described in text. Image (d) has been processed by Gaussian filter which over enhances contrast resulting in count saturation

function can be specified in one dimension as

$$F(u) = e^{-(u-u_0)^2/2\sigma^2}\ u \geq 0 \qquad (1)$$

where u is spatial frequency, u_0 is the displacement of the Gaussian from the origin and σ specifies the spread. An image of a typical filter is shown in Fig. 3.

This choice has the following advantages. The Gaussian function is easily computed and optimized because the free parameters, (u_0,o) are independent. Also, since the coordinate space representation (i.e., the convolution kernel) is a Gaussian modulated by a cosine function (*7*), application of the filtering in the spatial domain is straight forward.

The filter is optimized from the evaluation of a one-dimensional power spectrum created from averaging over annuli on the two-dimensional power spectrum

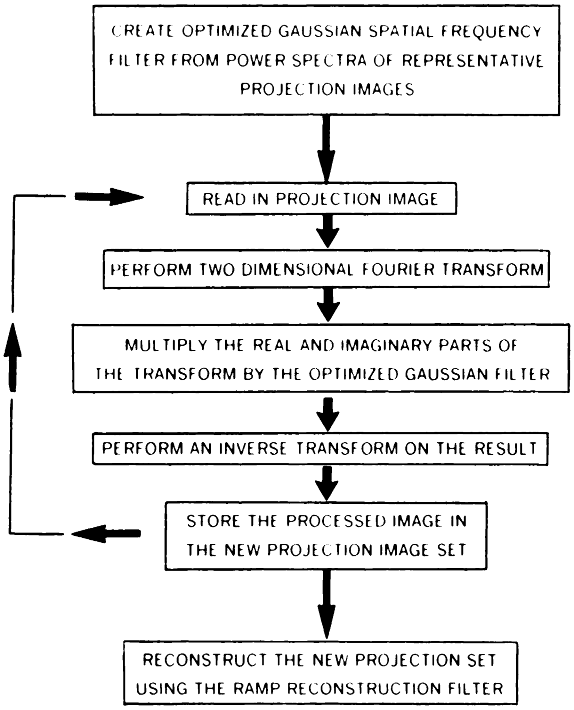

FIGURE 5
Block diagram of SPECT preprocessing method

associated with a representative projection image (*6,8*). The parameters are adjusted so that the following conditions are met.

Condition 1. The filter goes to 0 at the spatial frequency where the magnitude of the power spectrum approaches the average noise amplitude. Determination of the noise level for Condition 1 is simply done by examining the high frequency components of the power spectrum. This works for the following reasons. The power spectrum of Poisson noise is essentially flat oscillating about an average value (*5*). The power spectrum of the image is band limited, restricted not only by the actual range of contrast but by the modulation transfer function of the system. Therefore, the amplitudes at the high frequency end of the power spectrum correspond to the Poisson noise. In practice, Condition 1 is met by finding the spatial frequency (denoted by uc) where the average power spectrum is twice the noise level. This point is assumed to be 2 s.d. from the Gaussian mean, i.e.,

$$u_c - u_0 = 2\sigma. \quad (2)$$

Condition 2. The magnitude of the filter is 0.3 at the origin. Condition 2 was arrived at empirically as a compromise between two extremes. If the magnitude of the filter at the origin is 1 (i.e., maximum value of the Gaussian), then no contrast enhancement is seen. If the Gaussian is shifted too far to the right, negative saturation can occur. This is demonstrated in Fig. 4 which shows a raw projection image and three filtered images each illustrating the examples given above.

Mathematically, Condition 2 can be written as

$$e^{-u_0^2/2\sigma^2} = 0.3 \text{ which implies } \sigma = u_0/1.552. \quad (3)$$

Along with Eq. (2) this is sufficient to specify the Gaussian in terms of u_c

$$\sigma = u_c/3.552 \text{ and } u_0 = 0.437\, u_c. \quad (4)$$

Because u_c is easily determined from a search of the one-dimensional average power spectrum, the generation of the filter is easily automated. For studies in which the source distribution is more or less uniform about the center of rotation, the projection image selected to optimize is unimportant. However, if the sources are asymmetrically located near the periphery of the body, the power spectra will have an angular dependence. For such cases, interactive trials may be necessary to obtain the filter which best preserves the information of the high count projections without creating artifacts in the other projections.

MATERIALS AND METHODS

The SPECT studies are acquired on a rotating gamma camera system interfaced to a 16-bit minicomputer. Sixty-four 64 × 64 projection images are collected and stored in a disk file. The method (shown in the block diagram of Fig. 5) begins with the generation of power spectra for representative projection images. A two-dimensional Gaussian spatial frequency filter is produced by computer program using the criteria discussed above and is automatically stored on disk. The method then consists of serially reading each projection image, applying the filter to the Fourier transform of the projection, performing the inverse transform and storing the filtered image in the study. This is done for all 64 projection images. At the completion, the filtered set is reconstructed using a filtered backprojection algorithm with the ramp reconstruction filter. The method is also illustrated in Fig. 6.

RESULTS

The advantage of filtering the projections prior to reconstruction is demonstrated by comparisons with conventionally processed SPECT studies. An [^{123}I]IMP brain study and a [^{99m}Tc]MDP bone study of the lower spine were selected as examples.

Each of the studies was first reconstructed using the raw, unfiltered projection images. The reconstruction filter was chosen from those available in the software package* to yield the best images based on subjective comparisons. The projection data sets were then filtered using the method discussed and reconstructions were

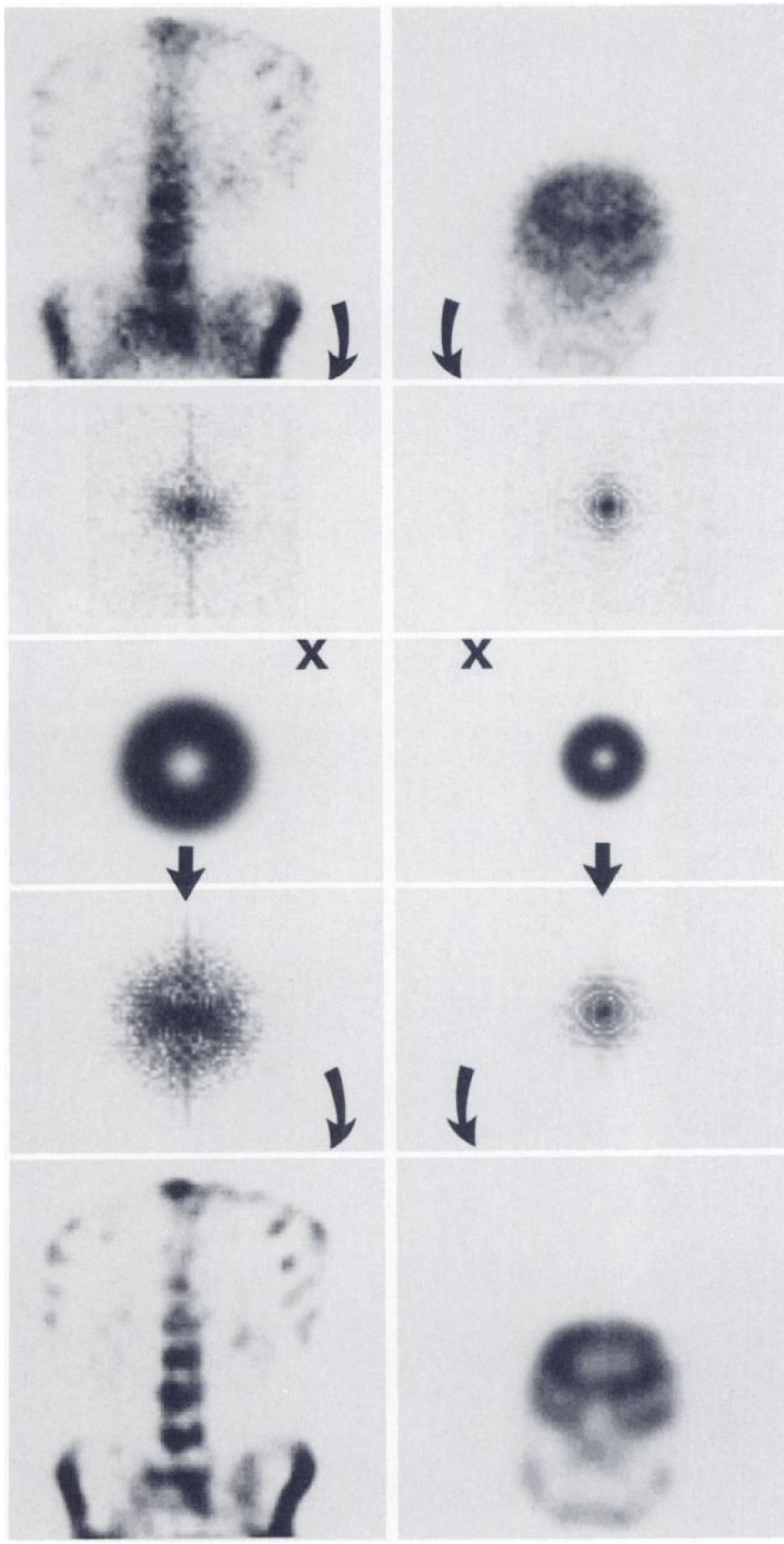

FIGURE 6
Pictorial representation of SPECT preprocessing method. IMP brain and MDP bone studies are used as examples to illustrate importance of filter optimization. All processing occurs in spatial frequency domain through multiplication of Fourier transformed projection image by Gaussian filter function

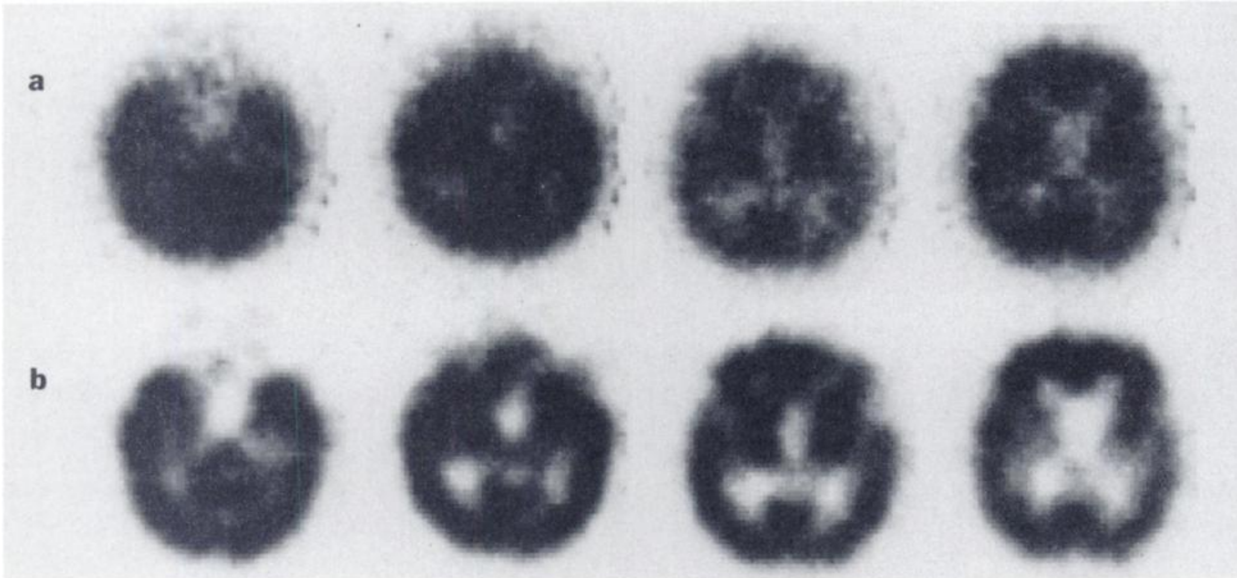

FIGURE 7
a: Conventionally processed transverse sections for [^{123}I]IMP brain study. b: Transverse sections of same study in which projection image set was preprocessed and reconstructed with ramp filter

performed using a ramp filter. Sagittal and coronal sections were obtained from the transverse images.

Figures 7 and 8 show comparisons for transverse and coronal sections of the [^{123}I]IMP brain. The images in which the projections have been processed are sharper and the magnitude of statistical fluctuations has been significantly reduced. All the structures which appear in the unprocessed images are clearly delineated, and no apparent artifacts are evident.

Similar results are seen for the bone images in Figs. 9 and 10. In addition to the enhanced contrast and suppressed noise level, the shape of the vertebral body and spinal canal are more precisely and symmetrically defined.

DISCUSSION

The SPECT images generated from the filtered pro-

FIGURE 8
a: Coronal slices generated from transverse images shown in Fig. 7a. b: Coronal slices from transverse images shown in Fig. 7b

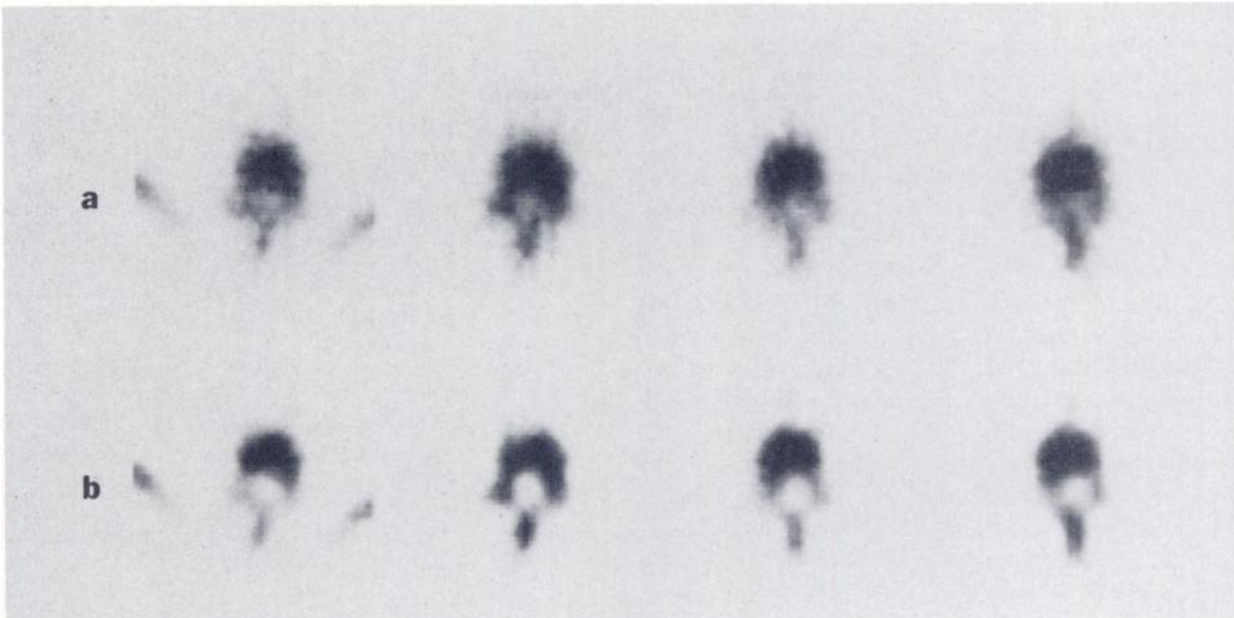

FIGURE 9
a: Conventionally processed transverse sections for [^{99m}Tc]MDP bone study. b: Transverse images from same study in which projections were preprocessed and reconstructed with ramp filter

jection set show significantly more detail without distracting statistical variations than the conventional SPECT images. We have used this method on a variety of studies with the same results. However, it is very important that the filter function be optimized. Factors which have an influence on the filter are the same as those which affect the quality of the projection images, e.g., the collimation, the number of acquired counts, the source distribution, and the source to collimator distance. For any given study a previously successful filter may be used as long as the above parameters remain constant.

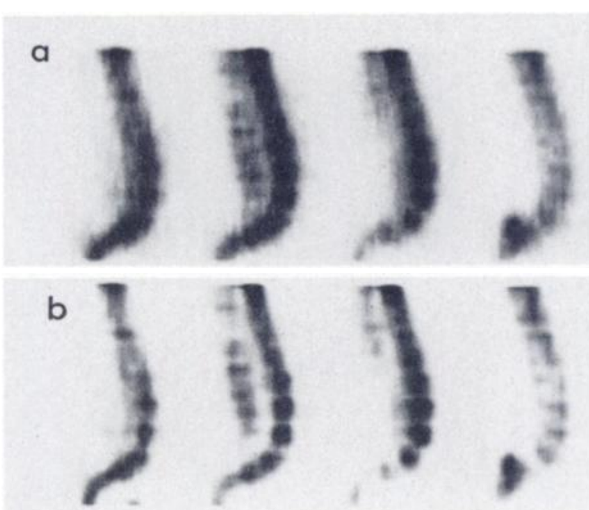

FIGURE 10
a: Sagittal slices from transverse images shown in Fig. 9a
b: Sagittal slices from transverse images shown in Fig. 9b

Any significant changes will require alterations of the filter.

Although the criteria for generating the filter were empirically derived, their selection was guided by the following principles. The first condition assures that the filter will not amplify frequencies in which noise dominates. Subtle changes in the falling portion of the Gaussian function are not expected to seriously alter the effectiveness of the filter. The second condition results in the enrichment of the middle frequencies in which much of the useful information about the image is located. Decreasing the magnitude of the zero frequency of the filter progressively sharpens the contrast, but can cause negative saturation. In images where the useful count information varies by more than a factor of 3, the condition given by Eq. (3) may have to be relaxed. Other criteria may provide better results for particular images and this is an area for further study. It should also be noted that the filter as described alters the total image counts. This will require a normalization procedure if quantitation is desired.

The success of this method suggests that some of the earlier estimates of count requirements for SPECT imaging were overly pessimistic (*9*). It is well known that the reconstruction process amplifies noise in the projections (*10*). However, much of the noise power occurs at frequencies beyond which any useful image information exists. Elimination of this noise can only improve the section images. The limitation becomes not one of statistical variations, but of resolution. As the number of acquired counts in the projections decrease, the spatial frequency at which the noise becomes dominant also decreases. Although the noise can be removed, the image information in the noise dominant frequencies cannot be recovered. We note that a formal analysis along these lines could be used to determine the criteria for optimal acquisition times in rotating camera SPECT imaging.

The time required to spatially filter a single 64 × 64 projection image is 20 sec on our system. Therefore, a set of 64 projections occupies the computer in excess of 20 min. Although this is not an insignificant amount of time, the improvement in the SPECT images warrants it. An array processor can reduce this time by more than a factor of 10 (*11*). Also, assuming the decisions about the filter parameters could be made from the first several projections, the filtering could be performed during acquisition. In this case, virtually no additional processing time would be evident to the user. An array processor would be a necessity in any facility where large number of SPECT studies are routinely performed or in gated cardiac SPECT studies where large numbers of projections are collected.

CONCLUSION

In conclusion, we have shown that a Gaussian spatially frequency filter applied to the projection images yields dramatic improvement in the quality of SPECT images. The filter is easy to compute and can be optimized in a simple and straightforward manner.

FOOTNOTE

* Digital Equipment SPETS 11 software.

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