DISCUSSION

The simulations without noise and without attenuation were designed to illustrate the effects of detector motion between two adjacent views in CSSM. The artifacts were clearly seen in Figures 2 and 3, particularly for the $4 + 4$ and $1 + 4$-sec CSSM acquisitions that were arbitrarily defined as fast CSSM acquisition because a 64-view study with a dual-head SPECT system could be finished in less than 4 min. The motion artifacts will be reduced by using more than 64 views due to a smaller angular step. In simulations with noise, the relatively minor artifacts of detector motion were covered by the fluctuation of the counts. By separate simulations, we isolated the effect of each of the different factors.

The quality of a noisy image largely depends on the number of counts of projection data. More counts reduce noise and hence improve image quality. A CSSM acquisition provides more counts than the corresponding SSM due to extra data collected as the detector moved from one view to the next. Therefore, the image quality of CSSM, especially fast CSSM, is better than that of the corresponding SSM because of more counts acquired in CSSM. With longer acquisition times (25 + 4 and 15 + 4 sec per view) that are most often used in current clinical studies, the image quality of CSSM is marginally better than that of the corresponding SSM since the total counts are comparable.

With attenuation, image quality obtained from both CSSM and SSM is further degraded. Due to the quality degradation, the improvement in image quality resulting from CSSM became less significant as compared to that obtained without attenuation. Therefore, to take full advantage of CSSM, attenuation effects should be corrected.

CONCLUSION

Image quality obtained from CSSM was appreciably improved compared to that obtained from SSM in fast or dynamic SPECT due to higher counts, but was only marginally better for relatively long-time SPECT due to comparable counts. Improvement in image quality due to CSSM was most noticeable when the attenuation effects were not present. Therefore, attenuation compensation should be performed in image reconstruction of fast SPECT. The results encourage us to conduct phantom and patient studies with CSSM acquisition in the near future.

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Nonstationary Scatter Subtraction-Restoration in High-Resolution PET

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Although removal of object scatter has been shown to improve both contrast and quantitation accuracy, subtraction of detector scatter leads to marginal contrast enhancement and negligible resolution recovery at the expense of reduced sensitivity and increased statistical noise. Since detector scatter has correct information about radioactivity but slightly erroneous information about source location, we suggest that this component should be restored to preserve sensitivity and improve resolution. Methods: A scatter correction model that consecutively removes object scatter and restores detector scatter is proposed. The scatter components are processed in the spatial domain using nonstationary scatter kernels. The detector scatter restoration kernel is obtained by piecewise inversion in the Fourier space. The model was tested using line source and hot spot phantom measurements. Results: Object scatter subtraction increased contrast substantively with no effect on resolution. Detector scatter restoration recovered resolution almost completely with modest contrast enhancement in small lesions.

Spillover effects were reduced to less than 5% for hot spots $\geq 3 \times$ FWHM, at the expense of moderate noise amplification. Conclusion: While subtraction of object scatter is necessary for contrast enhancement and quantitation accuracy, restoration of detector scatter preserves sensitivity and improves quantitation accuracy by reducing spillover effects in high-resolution PET. Key Words: PET; scatter components; scatter correction; subtraction-restoration; nonstationary


Image quality in PET is degraded, among other factors, by scatter contamination from the object, partial volume effects and noise due to low counting statistics. There have been two approaches to minimize these effects in conventional PET systems. In the first method, the effects of scatter are reduced by direct subtraction using integral transformation of the projections with scatter functions characteristic of the imaging system ($1-4$). Since the scatter distributions estimated by this method are fairly smooth, the process does not add significantly to the noise and a low-pass filter to suppress noise is often not
required (5,6). In the second method, blurring due to scatter and partial volume effects are restored by deconvolution filtering using the inverse of the imaging system's modulation transfer function (MTF). The effects of noise are minimized by preserving all detected events and by using a low-pass filter. This approach has been used extensively in SPECT imaging for similar purposes (7–9) and its extension to PET was only made recently (10–13). Implicit in this approach is the concept that object scatter does contain certain information about source location that should be preserved. Deconvolution filtering requires that the imaging system be shift invariant, which is not a valid assumption in most situations. In addition, object scatter restoration complicates the attenuation correction (14), since an effective broad-beam geometry coefficient that accounts for the scatter buildup in the object must be assumed (7,12,15). Due to the large size of the detector crystals, the spillover effects resulting from scattering of the photons in the detection system have been ignored in both cases.

As the resolution of PET scanners is improved by using ever smaller cross-section detectors, a significant proportion of scattered events is formed by 511-keV photon spillover from primary to secondary crystals (16–18). Since the spatial distribution and intensity of events formed by detector scatter differ substantially from that of object scatter, it was possible to make reliable estimates of the individual distributions from line source measurements and use them to correct for object, collimator and detector scatter contributions in projections by consecutive convolution subtraction (19,20). Whereas subtraction of object and collimator scatter improved image contrast, further subtraction of detector scatter amplified noise with marginal improvement of image resolution and contrast. These results raised the question of whether events formed by detector scatter should be removed or repositioned in the image (20,21).

To overcome this problem, we propose a scatter subtraction-restoration technique that maximizes the benefits of the two scatter correction approaches: (a) removal of object scatter to improve contrast and quantitative accuracy and (b) restoration of detector scatter to improve resolution and sensitivity. Other degrading effects, such as positron range, intrinsic resolution and penetration that can be processed concurrently to minimize the partial volume effect (22–24), were not performed here to evaluate the specific benefits of detector scatter restoration.

**MATERIALS AND METHODS**

**Phantom Measurements**

The measurements were made with the Sherbrooke PET camera simulator (25) set to simulate a small-animal PET scanner as described in earlier work (20,26). The lower energy threshold on each detector was set at 172 keV to acquire a large fraction of scattered photons. Measurements were acquired simultaneously with their randoms in a delayed-coincidence time window. Efficiency calibration measurements and subtraction of randoms, the tomographic data were rebinned into 128 projections of 63 parallel lines of response separated by a distance of 1.9 mm. Images were reconstructed by filtered backprojection on a 128 × 128 grid using projection data interpolated to 0.95 mm and a ramp filter of cutoff frequency 0.53 mm⁻¹. No attenuation correction was performed.

Three sets of measurements were made for the following purposes. The first set, from which the system response functions were obtained, was made with a line source of ⁴⁰⁸Na at 50 mm from the center of a cylindrical acrylic phantom of 110 mm diameter and 25.4 mm high. The position-dependent spread functions and fractions of the true or geometric (g), object scatter (o) and detector scatter (d) components were derived from the projections measured at the various incidence angles as described elsewhere (20). The desired kernel $F_i$ (i = g, o, d) at position $x_s$ is the product:

$$F_i(x_s, x) = f_i(x_s) H_i(x_s, x)$$

Eq. 1

of the component fraction $f_i$ (such that $f_g + f_o + f_d = 1$) and the spread function $H_i$. Examples of object and detector scatter kernels are shown in Figure 1 and the fractions $f_i$ as a function of $x_s$ for the system used in this study are plotted in Figure 2.

**FIGURE 1.** Examples of (A) object and (B) detector scatter kernels for different source positions in the projection.

**FIGURE 2.** Fractions of object scatter, detector scatter and geometric components as a function of source position in the projection of a small animal-sized high-resolution PET system with an 11-cm diameter object in the FOV. The dashed curve shows the fraction of detected events contributing to the image when restoring detector scatter.
The second set of measurements for assessing the effects of detector scatter restoration on spatial resolution was made with the line source at 0 mm and 30 mm from the center of the phantom. The third set of measurements used to assess image quality in terms of contrast enhancement and quantitative accuracy recovery was performed with a 110-mm diameter-phantom with hot cylinders of various diameters (Table 1) located at 28.3 mm from the scanner axis. At this distance from center, the system has a resolution of about 3.0 mm FWHM without sampling motion (26).

Object Scatter Subtraction

The random-corrected projection \( P_m \) is the sum of true events (T), object (S_o) and detector (S_d) scattered events (20, 21):

\[
P_m = T + S_o + S_d
\]

Eq. 2

After the method of Bergström et al. (1), the projection \( P_o \) \( \approx \) T + S_o free of object scatter, is estimated from \( P_m \) by the nonstationary convolution subtraction:

\[
P_o = P_m \otimes (\delta - F_o)
\]

Eq. 3

where \( \otimes \) and \( \delta \) denote a nonstationary convolution operation and the Dirac delta function, respectively.

Detector Scatter Restoration

Restoration of the detector scatter contribution can be performed by inverse filtering of the imaging system's detector scatter function. The initial step to evaluate the detector scatter kernel is to subtract object scatter from the system's response and renormalize the resulting distribution to unit source strength as follows:

\[
H'(x_s, x) = \frac{f_g}{f_g + f_d} H_g(x_s, x) + \frac{f_d}{f_g + f_d} H_d(x_s, x)
\]

Eq. 4

where \( f_g + f_d = 1 \). In the absence of object scatter, the effect of detector scatter is to blur the geometric component. Its distribution can be represented by the nonstationary convolution:

\[
H_d(x_s, x) = H_g(x_s, x) \otimes K_d(x_s, x)
\]

Eq. 5

where \( K_d(x_s, x) \) is a blurring function dependent on the scattering characteristics of the detection system. Substituting in Equation 4, we have:

\[
H'(x_s, x) = H_g(x_s, x) \otimes \left[ \frac{f_g}{f_g + f_d} \delta(x_s, x) + \frac{f_d}{f_g + f_d} K_d(x_s, x) \right]
\]

Eq. 6

Restoration for the blurring effects of the detector scatter component to reposition events in the geometric component would normally require inversion of Equation 6 in Fourier space. Such a procedure is feasible only if \( K_d \) is stationary. However, since this is not the case here (Fig. 2), the assumption \( K_d(x) \approx K_d(x_s, x) \) was made to allow an inverse kernel to be calculated for each source position \( x_s \) in the projection (see Discussion). By using this approximation, a piecewise inversion of the term on the right of the convolution can be performed and the following shift-variant restoration kernel is obtained:

\[
R_d(x_s, x) = FT^{-1}\left\{ \frac{f_g + f_d}{f_g + FT[f_d K_d(x_s, x)]} \right\}
\]

Eq. 7

where FT and \( FT^{-1} \) are the forward and inverse Fourier transforms, respectively. Neglecting the blurring effects of the geometric component, the term \( f_g K_d \) in Equation 7 can be replaced by the measured detector scatter kernel \( F_d(x_s, x) \) defined in Equation 1. In practice, the off center \( F_d(x_s, x) \) is not perfectly symmetrical. The right and left slopes were thus made equal by arithmetic averaging to eliminate the small asymmetries, thereby rendering the imaginary part of the restoration kernel negligible. Figure 3 displays examples of the position-dependent restoration kernels computed using Equation 7.

Restoration of the detector scatter is then performed by convolving the object scatter-corrected projection \( P_o \) obtained in Equation 3 with the symmetric but shift-variant kernel \( R_d(x_s, x) \):

\[
P_{od} = P_o \otimes R_d
\]

Eq. 8

to obtain the projection \( P_{od} \) free of object scatter and blurring effects induced by scatter in the detector. Since the geometric component was excluded in the restoration process, the resolution recovery by inverse filtering converges to the resolution of the geometric component. Thus, unlike other approaches where full-resolution recovery based on the system response (including the geometric detector response) has been attempted (12, 13), a low-pass filter to suppress noise amplification was not required.

Indices of Image Quality

The ability of the technique to improve image quality was assessed by indices of resolution, contrast, quantitative accuracy and noise characteristics. Resolution recovery was measured by the FWHM, the FWTM and the modulation transfer function (MTF) in the reconstructed images of a line source at the center and at 30 mm from the center. The FWHM and FWTM were calculated by linear interpolation between nearest pixels on a radial profile through the reconstructed line source. The MTF was taken as the Fourier transform of the same profile normalized to unity. The geometric
response reconstructed from the true component \( F_0(x_0, x) \) in the projected response function was used as an estimate of the highest achievable resolution without considering the effects of positron range and nonparallel flight of the annihilation photons.

The percent image contrast (IC) used to evaluate image quality was calculated from the hot spot phantom images as:

\[
IC(D) = 100 \frac{CP_h(D) - CP_c}{CP_h(D)} \quad \text{Eq. 9}
\]

where \( CP_h \) is the average counts/pixel in the selected ROI for each circular hot spot of diameter \( D \) and \( CP_c \) is the mean counts/pixel in a background ROI in the cold region surrounding the hot spots. Circular ROIs with the size given in Table 1 were selected manually at the center of each spot to determine the count density in the reconstructed images. The ROI in the background region consisted of 349 pixels.

The third index of image quality is the relative recovery factor (RF). It measures the quantitative accuracy loss due to spillover effects and it is evaluated according to the formula:

\[
RF(D) = 100 \frac{CP_h(D)}{CP_h(D_{\text{max}})} \quad \text{Eq. 10}
\]

where \( D_{\text{max}} \) is the diameter of the largest hot spot. The ROI sizes indicated in Table 1 were used again in evaluating the count density for each hot spot.

The last index, the standard deviation (STD), was measured to estimate the amplification of statistical noise resulting from the scatter corrections. The mean counts/pixel (M) and STD were evaluated in the ROI defined in the largest hot spot and in a cold region. The standard deviation was evaluated as:

\[
\text{STD} = \sqrt{\frac{1}{N} \sum_{i=1}^{N} (C_i - M)^2} \quad \text{Eq. 11}
\]

where \( C_i \) is the number of counts in pixel \( i \), and \( N \) is the total number of pixels in the ROI.

**RESULTS**

**Spatial Resolution**

Table 2 presents the FWHM and FWTM evaluated from the profiles of the line source at 0 mm and 30 mm from the center, reconstructed without scatter correction (Data), with object scatter subtraction (Obj) and with the combined object scatter subtraction and detector scatter restoration (Obj + Det). The highest possible resolution, obtained by reconstructing only the geometric component, is also given for comparison. From these data, it is evident that removal of object scatter has no effect on resolution while restoration of detector scatter improves resolution. The MTFs shown in Figure 4 also support this observation, although complete resolution recovery is not achieved. For the line source off center (Fig. 4B), the restoration undercorrects for the spreading at the base of the distribution (low frequency) but slightly overcorrects for the spatial resolution (high frequency), which agrees with the resolution data in Table 2.

The effects of scatter correction on resolution can also be assessed qualitatively from the images presented in Figure 5. Detector scatter restoration sharpens the edges of the hot spots as it brings counts from immediate surroundings (dark circles) to the top (light circles) of the hot spots (Fig. 5B, top right). For smaller hot spots with diameters comparable to the resolution, the restoration has an appreciable effect, as it significantly increases the counts per pixel in the hot regions.

**Image Contrast**

The effect of scatter on contrast can also be appraised from the images given in Figure 5. With scatter correction, the hot spots have higher contrast than those of the uncorrected image because the background level is relatively lower and the smaller hot spots are heightened by the restoration process. According to the profiles through the spots of 3.4 mm and 9.7 mm diameters shown in Figure 6A, object scatter subtraction decreases counts in both the hot and cold regions, as expected. Restoration of the detector scatter, however, increases counts in the hot region at the expense of the surrounding cold region, especially for the smaller spots. Figure 6B shows the corresponding profiles taken from the residual images of Figure 5B.

**TABLE 2**

Resolution Measured from Reconstructed Images of Line Source at Center and at 30 mm off Center

<table>
<thead>
<tr>
<th></th>
<th>FWHM (mm)</th>
<th>FWTM (mm)</th>
<th>FWHM (mm)</th>
<th>FWTM (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Data</td>
<td>2.23</td>
<td>4.54</td>
<td>3.20</td>
<td>5.03</td>
</tr>
<tr>
<td>Obj</td>
<td>2.23</td>
<td>4.54</td>
<td>3.20</td>
<td>5.03</td>
</tr>
<tr>
<td>Obj + Det</td>
<td>2.11</td>
<td>4.35</td>
<td>3.12</td>
<td>4.88</td>
</tr>
<tr>
<td>Geometric</td>
<td>2.12</td>
<td>4.38</td>
<td>3.23</td>
<td>5.06</td>
</tr>
</tbody>
</table>

*The line source was in a scattering media of 110 mm diameter.*
The amount of the object scatter subtracted is broadly distributed over the entire field of view. The restored detector scatter closely matches the shape of the activity distribution. The evaluation of image contrast as a function of hot spot diameter indicates that the correction for object scatter is mainly responsible for the contrast enhancement, as shown in Figure 7. This result is expected since blurring due to detector scatter inside large uniform regions cancels out. However, for the smaller hot spots, the restoration of detector scatter moves counts from the.

**FIGURE 5.** (A) Images of the hot spot phantom without correction (top left), with object scatter removed (top right) and with object scatter removed and detector scatter restored (bottom left). (B) Residual images obtained by subtracting the object scatter corrected image from the uncorrected image (top left), the object scatter corrected image from the object and detector scatter corrected image (top right) and the uncorrected image from the object and detector scatter corrected image (bottom left). The arrow indicates the position of the profile shown in Figure 6. Note the inverted position of the two smallest spots in the sequence.

FIGURE 6. Profiles (A) through the images of the hot spot phantom and (B) through the residual images shown in Figures 5A and B, respectively. The profiles are through the hot spots of diameters 3.4 mm and 9.7 mm (see arrow in Fig. 5). The small peaks at positions ±53 mm are due to a thin film of radioactive solution surrounding the phantom in the container (also visible as a ring in Fig. 5).

surrounding to more central parts of the distribution and, thus, the correction yields some contrast enhancement.

**Recovery Factor**

The relative recovery factors evaluated according to Equation 10 are presented in Figure 8 as a function of hot spot diameter. The comparison indicates that object scatter subtraction does not improve quantitation recovery, while restoration of detector

**FIGURE 7.** Comparison of image contrast as a function of hot spot diameter for images reconstructed without scatter correction (Data), with object scatter subtracted (Obj) and with object scatter subtracted and detector scatter restored (Obj + Det).
scatter significantly improves recovery by minimizing the spillover effects. The resolution enhancement resulting from the restoration allows accurate quantification of radioactivity (more than 95% recovery) in objects with dimensions as small as $3 \times$ FWHM ($D \geq 3$ mm). Without restoration, the same accuracy can only be achieved in objects of dimensions larger than $5 \times$ FWHM ($D > 15$ mm). For further improvement of the recovery factor below $3 \times$ FWHM, restoration of the geometric detector response would be required to compensate for the partial volume effect.

Statistical Noise

The effects of the scatter corrections on the statistical noise can be noted from the changes in the texture of the noise in the hot spots and the background in the images of Figure 5. The statistical fluctuations represented by STD in the uncorrected and scatter corrected images are given in Table 3. The images were reconstructed using a ramp filter without low-pass filter since one of the objectives of this study was to test the effect of detector scatter restoration on spatial resolution. This has resulted in some amplification of the noise magnitude in the original, uncorrected images that amounts to a relative s.d. of nearly 7% in the largest hot spot and 27% in the background. These contributions add up to that generated by the scatter corrections. Correction of object scatter by convolution subtraction did not significantly increase STD, in accordance with what other investigators have reported (6). Restoration of the detector scatter increased noise by 31% in the hot region and 38% in the cold region, relative to the object scatter corrected image. These quantities become 34% and 39%, respectively, relative to the original uncorrected image. Since the mean values in the hot region and the background remain nearly unchanged after detector scatter restoration, the increase in STD is a result of the high-frequency noise amplification of the restoration filter. These effects are apparent in the images of Figure 5 and the profiles shown in Figure 6.

**DISCUSSION**

**Shift-Variant Scatter Kernels**

The dependence of the scatter kernels $F_i$ on source position is through the variation of the scatter fractions $f_i$ (Fig. 2) and the scatter spread functions $H_i$ with position in the projection. By following a method that has been validated for cylindrical objects with uniform attenuation ($1, 2, 3$), subtraction of position-dependent object scatter contributions was achieved by nonstationary convolution with the shift-variant object scatter kernel $F_{oi}$.

The problem central to the nonstationary restoration of the detector scatter was to find the inverse kernel. Since a nonstationary restoration kernel cannot be derived by standard Fourier deconvolution procedures, the restoration cannot be carried out as a deconvolution. However, since a blurring function

$$H'(x_s, x) = \left(\frac{f_g}{f_g + f_d} \delta(x_s, x) + \frac{f_d}{f_g + f_d} K_d(x_s, x)\right)$$

Eq. 12

exists for each source position $x_s$ in the projection, an inverse function $R_d(x_s, x)$ would equally exist such that

$$H'(x_s, x) * R_d(x_s, x) = \delta(x_s, x)$$

Eq. 13

where * holds for the standard (stationary) convolution operation. The value of $R_d(x_s, x)$ for a given position $x_s$ could be calculated in Fourier space by assuming that $H'(x_s, x)$ in the Fourier transforms was stationary and equal to its value at position $x_s$. By repeating this calculation for each position $x_s$ in the projection, an approximate shift-variant inverse kernel $R_d$ can be found. This piecewise inversion in Fourier space can be performed provided the detector scatter kernel $K_d$ is considered stationary and symmetric for each position of the source in the projection. While the piecewise inversion is not formally exact, the errors introduced by this approximation are expected to be minimal and the position dependence of the blurring function accounted for rather accurately if the variations of the kernel and projection data with position are not too important. This approximation may thus be responsible for the slight overcorrection of resolution observed when the line source is off center, since the assumption of a symmetric detector scatter kernel gets worse as we move from the center towards the edge of the field.

**Image Quality**

As was previously observed by several workers ($1, 2, 3$), subtraction of object scatter is mainly responsible for contrast enhancement. Such contrast improvement has a substantive effect on detectability of small hot spots, as can be seen in Figure 7, and facilitates detection of large weak sources. However, the main benefit of scatter subtraction is to improve quantitation accuracy by removing contamination of events with inaccurate positional information. Our data show that the presence or absence of object scatter does not improve or degrade resolution (Table 2 and Fig. 4) or has any effect on the recovery factor that is a measure of the loss of quantitative accuracy due to resolution (Fig. 8). These observations are consistent with the postulate that little useful information about the source position and intensity can effectively be recovered from this component.
The disadvantages of subtracting the detector scatter component were manifest in our earlier work (20), as it was demonstrated that its removal from images substantially lowered signal with insignificant gain of resolution and contrast. On the contrary, restoration of the detector scatter preserves sensitivity (Fig. 2), enhances resolution (Table 2 and Fig. 4) and improves the recovery factor (Fig. 8). As a result, the detectability and quantitation of small lesions are improved and the detection of larger objects is facilitated by the sharper edge response. These features are evident in Figure 5A. The benefits must be traded off with an increase of statistical fluctuations, but the possibility of using a low-pass filter to suppress high-frequency noise amplification always exists (8–13).

Rationale for Differential Scatter Correction
By definition, an ideal PET image would be formed by annihilation photons emitted by radioactive distributions suspended in air and detected at the points of first interaction in the detection system. Since the formation of events in this manner is not feasible in practice, attenuation and scatter correction methods were introduced to minimize effects due to the presence of the object. Because detector scatter is formed by photons that have not interacted with the object, it is logical to remove the effects of object scatter before correcting for detector scatter (27).

Detector scatter events are formed by annihilation photons measured at a short distance from the primary detector that corresponds to the source position in the object. These events would actually be registered as true events, had the detection system been ideal. Since such events possess the desired information about the activity but slightly inaccurate information about the source location, it is legitimate to compensate for the inherent deficiency of the measuring instrument by restoration because it brings the event registration closer to ideal detection. The main benefit associated with a specific treatment of object and detector scatter is the ability to correct for the loss of image contrast and quantitation accuracy due to object scatter as we recover the loss of resolution and sensitivity due to detector scatter. Since the energy of the detector scattered photons can be low, as has become evident in multispectral imaging (28), the proposed differential scatter correction enables the use of broader energy windows than conventional to maximize detection of annihilation photons in high-resolution PET systems based on narrow individual crystals.

CONCLUSION
A method that allows sequential removal of object scatter and restoration of detector scatter has been described and validated using line source and hot spot phantom measurements. Removal of object scatter improves contrast while restoration of detector scatter preserves sensitivity and improves spatial resolution with relatively little penalty on image noise. Although modest, the resolution recovery obtained was sufficient to overcome the spillover effects, thereby improving detectability, contrast and quantitation accuracy in small ROIs. With the proposed detector scatter restoration, recovery factors better than 95% were obtained for objects as small as 3 × FWHM, whereas the same accuracy could only be achieved in objects larger than 5 × FWHM without restoration. The method permits the use of broader energy windows than commonly used in PET. This feature is particularly important in high-resolution PET, in which a large portion of annihilation photons are detected with low energy.

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