Radionuclide Image Minification Can Compensate for Coarse Digitization: Concise Communication

W. Robert Pitt, Peter F. Sharp, Robert B. Chesser,* and Philip P. Dendy†

University of Aberdeen; Foresterhill, Aberdeen AB9 2ZD, Scotland

Although it is common practice to digitize radionuclide images onto the finest matrix available, their low count densities and poor spatial resolution suggest that quite large pixels should be adequate. Observers find these large pixels visually obtrusive, but minification of the image can reduce this effect. Experiments reported here have investigated how minification (achieved by increasing viewing distance) affects the perceived quality of images digitized onto different sized matrices. Observers' subjective preference for different pixel sizes was measured at various viewing distances using clinical bone images as test patterns. An objective measure of image quality was made by comparing the detectability of computer-generated focal areas of increased activity both in simple noisy backgrounds and in clinical bone images. The results show that a 128 × 128 matrix is adequate when the image is 8 cm² and is viewed from 1 and 2 m. A finer matrix failed to produce better results.

J Nucl Med 24: 1046–1054, 1983

Digitized radionuclide images have many advantages over analog images. As well as permitting quantitative analysis, a digitized image is more conducive to visual interpretation, since it gives an observer the opportunity to vary at will the intensity, contrast, and gray or color-scale levels of the display. In addition they allow the possibility of applying more sophisticated image-processing facilities. None of this manipulation is possible on an analog image.

The major disadvantage of a digitized image is that the matrix of discrete elements (pixels) imposes an artificial regular structure on the displayed data. In an attempt to reduce this problem, some authors recommend digitizing data onto the finest matrix possible so that the images appear similar to analog ones (*1*). Although this approach may be appropriate for high-quality images where the only requirement is to retain fine spatial detail, difficulties arise with radionuclide images because of their low count densities. Small pixels contain few counts, so their random noise content is higher. Rollo and Harris (*2*) state that each pixel in an image must contain a statistically meaningful number of counts, and this consideration limits the smallest size of pixel that may be used (*3,4*).

The combination of statistical restraints with the poor spatial resolution of a gamma camera leads to the choice, on physical grounds alone, of a relatively large pixel for digitizing radionuclide images. Todd-Pokropek (*3*) suggests that a pixel whose side length represents as much as 5 mm on the camera face is adequate for any clinical image in which spatial resolution has been degraded by the scattering effects of soft tissue.

Clearly the use of a finer matrix than is necessary merely increases costs (in terms of computer memory, time for data manipulation, and data storage space) without improving diagnostic effectiveness. A choice of pixel size made on physical criteria alone, however, may not be the best for visual interpretation. Sharp and co-

Received Feb. 16, 1983; revision accepted June 1, 1983.

For reprints contact: Dr. P. F. Sharp, Dept. of Bio-Medical Physics and Bio-Engineering, University of Aberdeen, Foresterhill, Aberdeen AB9 2ZD, Scotland.

* Present address: M & D Technology, Unit 1, Whitemyers Ave., Mastrick, Aberdeen, Scotland.

† Present address: Dept. of Medical Physics, Addenbrooke's Hospital, Hills Road, Cambridge, England.

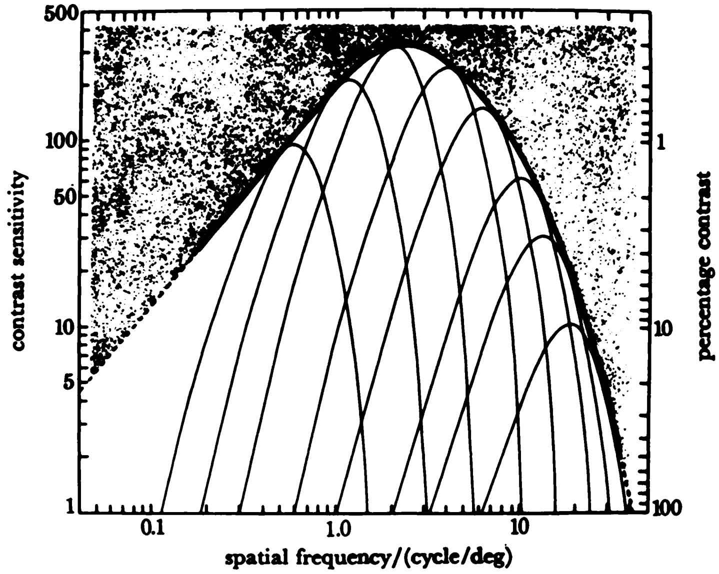

FIG. 1. Thick curve represents contrast sensitivity (defined as reciprocal threshold contrast) of human visual system (measured with sinusoidal gratings) plotted against spatial frequency. Lighter curves represent channels sensitive to narrow range of spatial frequencies. (Reprinted from Ref. 8, by permission.)

workers (*5*) have shown that although large pixels are adequate for simple perceptual tasks, observers find the pixels themselves to be visually obtrusive when more complex data are displayed. For this reason data initially collected on coarse matrices are often interpolated onto finer ones for display. Although these images retain the lower statistical noise levels of the collection matrix, they appear blurred and are also visually unsatisfactory (*5*).

A matrix that is visible in coarsely digitized images contains higher spatial frequencies than the underlying data. Although the sharp edges of the pixels contain very high spatial frequencies, it is believed that those artificially introduced frequencies within two octaves of the signal spectrum cause most disruption (*6*). It is the problem of these higher spatial frequencies that must be overcome to make coarsely digitized images visually acceptable.

The response of the human visual system varies greatly with observed spatial frequency. Figure 1 shows that it peaks at a few cycles per degree (*7,8*). Thus, by varying the visual angle of an image, for example, by changing the viewing distance, it should be possible to enhance the signal and at the same time reduce the interference from the pixel matrix. This is discussed more fully elsewhere (*9*).

The effect of minifying radionuclide images has been discussed in the literature (*10–14*) but not specifically with the intention of improving the appearance of digitized images. This paper reports experiments to determine whether it is necessary to use very fine digitization matrices, or whether, after suitable image minification, observers are able to obtain the same amount of diagnostic information from coarser matrices that are computationally more convenient. To avoid image distortion, reduction of the visual angle was achieved by increasing the viewing distance rather than by a series of lenses or by reducing the physical size of the display.

METHODS

Three experiments have been performed to investigate the subjective and objective visual effects of minifying radionuclide images.

In each experiment data were first recorded on a 256 × 256 matrix, and images equivalent to those obtained by digitizing onto 128 × 128 or 64 × 64 matrices were produced by suitable merging of the original pixels. Clinical images were collected on a gamma camera interfaced to a data processor.

The effect of minification was achieved by keeping the image size constant and increasing the viewing distance from 1 to 12 m using a system of front-silvered mirrors. Square images with side lengths of 32 cm were displayed on a monochrome TV using 128 grey levels. When a 64 × 64 matrix was displayed, the range of minification reduced the visual angle subtended by a pixel side from 17 min of arc at 1 m to 1.4 min of arc at 12 m. The

FIG. 2. Simple test pattern used in Experiment 2. This image is digitized onto 128 × 128 matrix.

background room lighting was kept at the normal working level, although care was taken to avoid direct reflection of light sources from the TV screen. The TV monitor was interfaced to a computer through a microprocessor-controlled display that permitted interactive image manipulation (*15*). The observer had remote control of the image brightness, contrast, and the gray scale's dynamic range, and could position a flashing cursor on the screen by moving a roller-ball.

Experiment 1: Subjective preference using clinical images. To investigate whether the strong preference for finely digitized images reported elsewhere (*5*) persists when the visual angle of the image is reduced, three sets of bone studies, each containing ten different images, were shown to six experienced observers. Upper and lower posterior views were used as test patterns because these images contain the finest spatial detail found in nuclear medicine. The images were of normal patients and of patients with confirmed skeletal metastases.

The three digitized versions of each image were displayed consecutively in a random order to an observer placed 1, 4, 8, or 12 m from the TV screen. The observer altered the order of display of the three versions of each image using a computer keyboard. He was not instructed to adopt any particular criterion for his judgment of image quality but left to draw on his own experience of what constituted a good clinical image. When the observer was satisfied that the images were ranked in order according to his preference, the result was automatically recorded.

Experiment 2: Detectability of abnormalities in a noisy structureless background. The fact that observers prefer a particular type of image presentation does not necessarily mean that it yields the most diagnostic information. The second experiment used simple computer-generated test patterns to obtain a more objective measure of observer performance. Since by far the most common sign of malignant disease in a radionuclide bone image is the appearance of an area of increased count density, it was decided to concentrate only on these. Accordingly the test pattern comprised focal areas of increased activity superimposed on a structureless background containing random (Poisson) noise. The mean background count density was 20 counts per element on a 256 × 256 matrix, corresponding to 1280 counts/cm^2 on the TV face. The focal areas of increased radioactivity had Gaussian count-density profiles and were of two sizes, with full widths at half-maximum counts (FWHM) of 7.5 and 15 mm on the TV face (i.e., six and 12 pixels respectively on the 256 × 256 matrix). The small focal areas corresponded approximately to the point spread function (PSF) of a gamma camera under clinically realistic conditions. The contrast (defined in terms of the peak count density) increased in seven equal steps from 15% to 45% for the small abnormality and from 5% to 35% for the large one. The lower limits were chosen to provide focal areas of increased activity that were barely visible under any viewing conditions, while the upper limits gave spots that were nearly always visible.

Thirty different patterns were created, grouped into three distinct sets of ten. Each set of patterns contained ten focal areas of increased activity of each size and contrast (a total of 140 abnormalities) and between 11 and 17 of these were randomly located in each pattern. The original data were created on 256 × 256 matrices, and images on 128 × 128 and 64 × 64 matrices were created by merging adjacent pixels. An example of one of these images is shown in Fig. 2.

The approximate sizes of the focal areas of increased activity were demonstrated to each observer, but no other information, such as the minimum or maximum number of targets present or their probable positions, was given. In order to keep each observer's detection threshold constant, both within a viewing session and between sessions, he was asked to mark with the cursor only those positions in an image where he was certain that there was

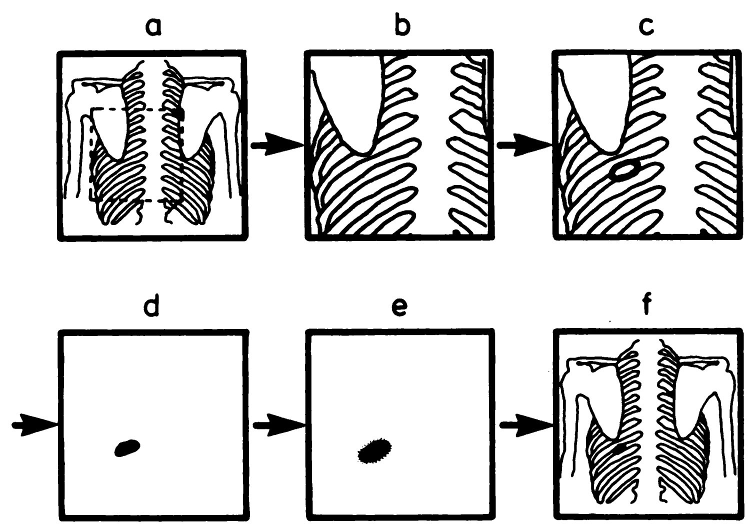

FIG. 3. Schema of method used to place foci of increased activity in normal bone images: (a) choose region of interest in bone image; (b) expand this region and smooth once; (c) mark outline of proposed focus of increased radioactivity; (d) extract outlined area; (e) smooth area twice, scale, and use these values as means to generate Poisson data; (f) superimpose area as focus of increased radioactivity in original image.

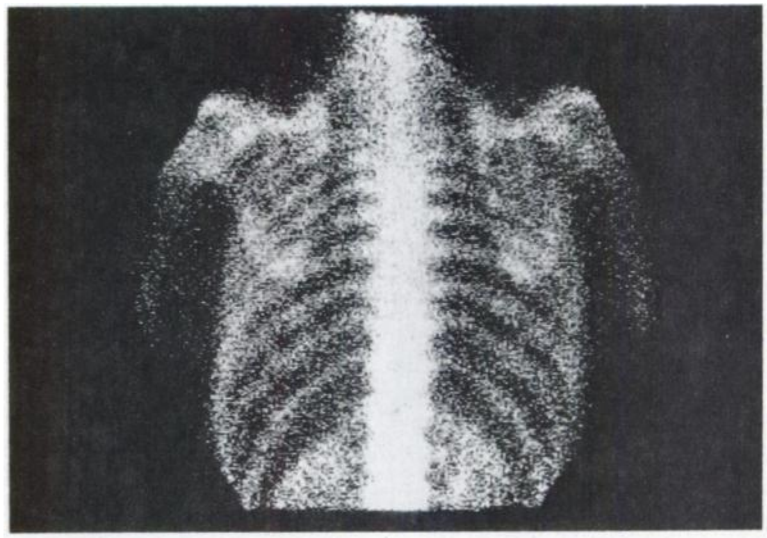

FIG. 4. Normal bone image on 256 × 256 matrix.

an abnormality. Doubtful areas were not to be marked.

Six observers took part in this experiment, and images were viewed from distances of 1, 4, and 8 m. Every observer underwent three viewing sessions at each distance. In a single session the observer saw all thirty different patterns, with one set of ten displayed on a 256 × 256 matrix, another on a 128 × 128 matrix, and the third on a 64 × 64 matrix. On the next occasion that this viewing distance was used, the same three sets were displayed but on matrices different from the previous one (e.g., 128 × 128, 64 × 64, 256 × 256). By the end of the third session at a particular viewing distance, an observer had seen all three sets of patterns on all three matrices, thus giving a maximum possible number of true-positive responses of 1260 at each distance. To minimize any distortion of the results caused by observers becoming familiar with the patterns, the order in which the 30 images were presented in each session was randomized. In addition, the order of the viewing distances used by each observer was also randomized, and the nine separate viewing sessions were spaced at least two days apart.

Experiment 3: Detectability of abnormalities in a highly structured clinical image. The previous experiment used a simple pattern in which any definite local increase in count density represented an abnormality, and the only noise present was due to random fluctuations. Clinical images, however, provide an irregular background due to normal anatomical or physiological features. This "anatomical noise" is extremely obvious in radionuclide bone images (as in Fig. 4) and observers must be able to distinguish these normal variations in count density from those resulting from pathological changes.

The third experiment used upper posterior bone images of normal patients as the background pattern, and clinically realistic foci of increased activity representing metastatic deposits were added under computer control. The method used to introduce the abnormalities is outlined in Fig. 3. A portion of the normal image, digitized

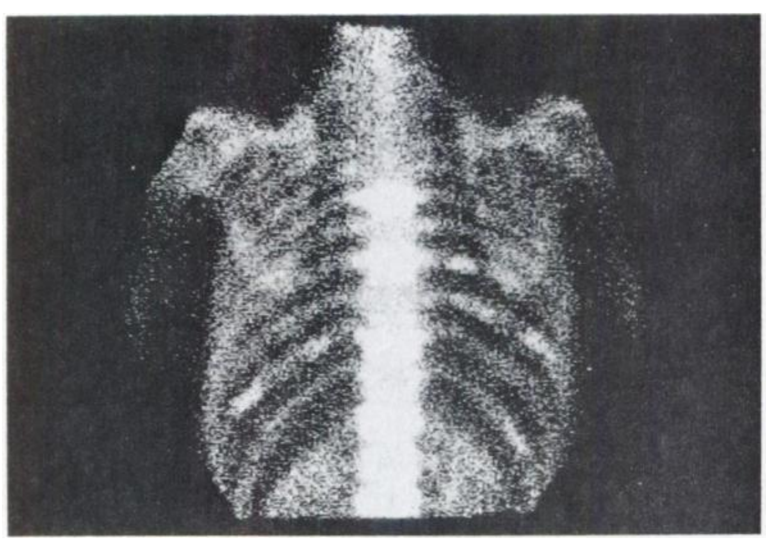

FIG. 5. Data from Fig. 4 with several artificial, high-contrast abnormalities added.

onto a 256 × 256 matrix, was expanded to twice its size and a single nine-point smoothing operation was performed. The outline of the proposed focus of increased activity was then marked on the expanded image and the spatial coordinates and contents of each pixel within this region were extracted by the computer. This extracted area, representing an image of the "lesion" on an empty background, was then given two more nine-point smoothings to blur the edges, then scaled to give the required "lesion" contrast. Poisson-distributed data were generated from these scaled values and this noisy focus of activity was added to the original image.

Two sizes of foci of increased activity were produced: small ones were placed in the ribs while large ones encompassed a whole vertebral body. As before, contrasts were chosen to cover the range from hardly visible to obvious. Thirty patterns were altered in this way, each finally containing from seven to 12 "lesions" (mean of ten). Figures 4 and 5 show, respectively, a pattern before and after artificial lesions had been introduced.

The viewing procedure for the second experiment was used, and a strict decision criterion was again adopted. Observers were asked to mark carefully the position of every region which, if it had been the only abnormality in the image, would have been sufficient to record the image as unequivocally positive. Equivocal areas were not to be marked.

Only observers who were experienced in examining bone images were eligible for this test, and the author responsible for selecting the positions for the foci of increased activity had to be excluded. For these reasons only four observers were available to undertake this part of the experiment.

RESULTS

Experiment 1. At each viewing distance a simple sign test (*16*) was used to assess the preference expressed for each image when viewed on two different matrices. A "+" was allocated whenever the finer matrix was preferred, and a "−" otherwise. No ties were allowed in the experiment. Results from all six observers were pooled and the "+" scores were calculated for each pair of matrices at each viewing distance. The results, expressed as percentages, are shown in Table 1. The null hypothesis, that there is no preference for one of the matrices, corresponds to a score of 50%, and the 95% confidence level calculated from the two-tailed binomial distribution (with n = 60 and expected probability 0.5) is ±13%.

TABLE 1. PERCENTAGE OF OCCASIONS WHEN A FINER MATRIX WAS PREFERRED TO A COARSER ONE

Preference	Viewing distance 1 m	4 m	8 m	12 m
256 over 128	100	62*	60*	55*
256 over 64	100	98	90	77
128 over 64	100	100	90	77

* Not significant at 95% confidence level.

Experiment 2. Each of the nine combinations of matrix size and viewing distance produced a different "viewing condition" (VC), and by the end of the experiment each observer had seen every abnormality under all nine VCs. The strict decision criterion resulted in a false-positive rate of no more than one in every ten images, which was considered to be negligible. For each observer and target size the analysis considered each increased focus in turn. For any two VCs the comparison between the results from each of the 210 abnormalities of a particular size was allocated to one of the four positions shown in Table 2: either the focal area of increased activity was seen under both conditions (k), seen under VC1 but not under VC2 (s), seen under VC2 but not under VC1 (r), or not seen at all (m). The responses in k and m are tied pairs, so the comparison of interest is between s and r. Thus a two-tailed binomial test was carried out on the proportion r/(r + s), with an expected value of 0.5 (*16*). Any deviation with a confidence level greater than 95% was considered significant.

Variation of detection rate with viewing distance. The effect of keeping the digitization matrix fixed and varying the viewing distance is shown in Table 3. For each observer and size of target, the viewing distances are listed according to their relative detection rates, with the distance producing the highest rate placed at the top. For example, Table 3 shows that Observer 1 saw more foci of increased activity of both sizes on the 64 × 64 matrix when the viewing distance was increased from 1 to 4 m, or from 1 to 8 m, but there was no significant difference between his results on moving from 4 to 8 m.

Results from coarse matrices at long viewing distances. To find out whether images digitized onto coarse matrices can, under suitable viewing conditions, yield as much information as those digitized onto finer ones,

TABLE 2. STATISTICAL TEST TO COMPARE THE NUMBER OF ABNORMALITIES SEEN UNDER ANY TWO VIEWING CONDITIONS (VCs)

		VC1 Seen	VC1 Not seen	Totals
VC2	Seen	k	r	r + k
	Not seen	s	m	s + m
	Totals	k + s	r + m	N

Null hypothesis: that r/(r + s) follows the binomial distribution, with expected probability of 0.5.

TABLE 3. RELATIVE RATES OF DETECTION OF TARGETS IN A STRUCTURELESS BACKGROUND (FIG. 2) AT VIEWING DISTANCES OF 1, 4, AND 8 m

Small targets Matrix size	Observer number 1	2	3	4	5	6
64	8 = 4 1	8 4 1	8 = 4 1	8 4 1	8 4 = 1	8 = 4 = 1
128	8 = 4 = 1	8 4 1	8 = 4 1	8 = 4 = 1	8 = 4 1	8 4 1
256	8 4 = 1	8 = 4 1	8 4 1	8 = 4 = 1	8 = 4 1	8 = 4 1

Large targets Matrix size	Observer Number 1	2	3	4	5	6
64	8 = 4 1	8 4 = 1	8 = 4 1	8 = 4 1	8 = 4 1	8 = 4 = 1
128	8 = 4* 1 = 4	8 4 1	8 = 4 1	8 = 4 1	8 = 4 1	8 4 = 1
256	8 4 = 1	8 4 1	8 4 1	8 4 = 1	8 4 1	8 4 1

* This means that significantly more targets were detected at 8 m than at 1 m, but results of 4 m were not significantly different from those at 8 m or 1 m.

Top figure is distance at which most targets were detected; bottom figure is that at which fewest targets were detected; and equality signs indicate no difference.

results from the 64 × 64 and 128 × 128 matrices viewed from 8 m were compared with those from every other VC. The conclusions are shown in Figs. 6 and 7 respectively. Figure 6 shows that some observers found several VCs better than the 64 × 64 matrix at 8 m for the small foci of activity, but no VC was better than this for the large abnormalities. There are no stars in Fig. 7, however, showing that no VC produced significantly better results than the 128 × 128 matrix at 8 m for either size of abnormality.

Experiment 3. The results from this experiment were analyzed in exactly the same way as those from the second experiment.

Variation of detection rate with viewing distance. Table 4 is analogous to Table 3, although the pattern of detection rates is less consistent, as might be expected from a more complex image.

Results from coarse matrices at long viewing distances. Comparisons of results from the 128 × 128 matrix viewed from 8 and 4 m with all other VCs are shown for individual observers in Figs. 8 and 9, respectively. These figures also show the comparisons of the pooled results of all four observers.

DISCUSSION

Experiment 1. The results in Table 1 show that observers disliked the coarsest matrix (64 × 64) even when

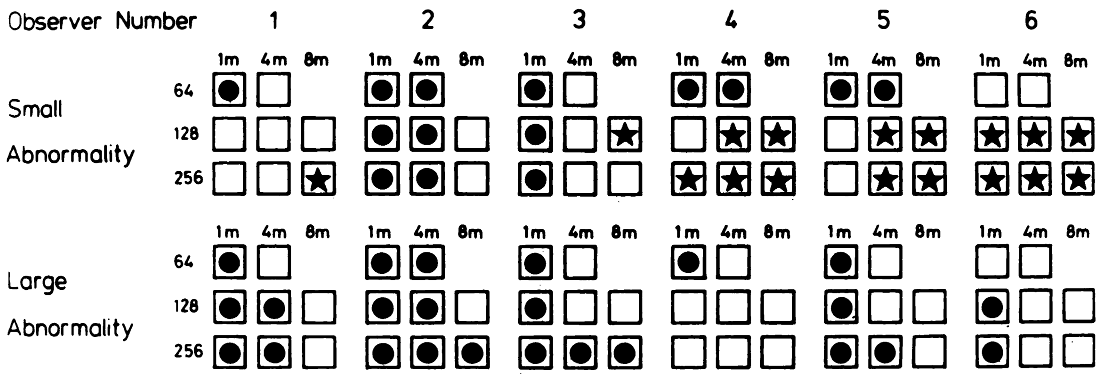

FIG. 6. Comparison of results from viewing 64 × 64 matrix from 8 m with other combinations of viewing distance and pixel size. Test pattern was of targets in structureless background, as shown in Fig. 2. Solid circles (●) indicate worse results than 64 × 64 at 8 m, stars (★) indicate better results, and empty squares (□) no difference (all at 95% confidence level).

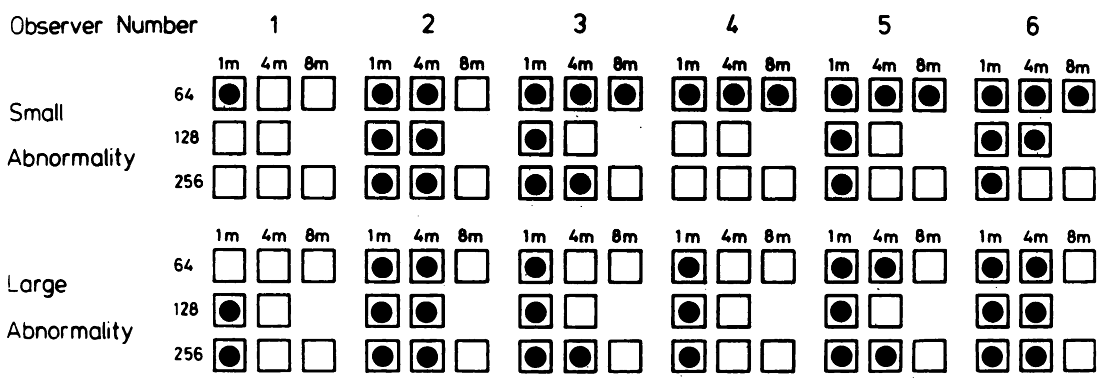

FIG. 7. Comparison of results from viewing 128 × 128 matrix from 8 m with other combinations of viewing distance and pixel size. Test pattern was of targets in structureless background, as shown in Fig. 2. Symbols are as in Fig. 6.

viewed from 12 m. There was no significant preference for the 256 × 256 matrix over the 128 × 128, however, for viewing distances of 4 m or more.

Experiment 2. Table 3 shows that, despite considerable interobserver variability, the rate of detection of simple foci of increased activity in a noisy, structureless background generally increased as the viewing distance increased up to 8 m. It is expected that there will be a viewing distance beyond which detectability starts to decrease [as reported elsewhere (*17*)], but there was no evidence that this limit had been reached in this experiment.

The object of the experiment was to test whether very fine matrices were needed to produce the best visual response. Since the best results were obtained from 8 m (Table 3), comparisons of the results from the two coarser matrices viewed from this distance were made with all other VCs. Figure 6 shows that, for the large abnormality, a 64 × 64 matrix viewed from 8 m gave results that were always as good as, and often significantly better than, those from the finer matrices at any distance. This was not the case for the small abnormality, however, for which finer matrices often gave better results.

Figure 7 shows that the 128 × 128 matrix viewed from 8 m gave results as good as, or better than, all other VCs for both sizes of abnormality. In particular it can be seen that no improvement was ever obtained by using a 256 × 256 matrix at any distance. Thus it appears that a 128 × 128 matrix is adequate for this experiment, and no improvement results from the use of a finer one.

Experiment 3. Results from the clinically realistic images were less straightforward than those from the simple test pattern. Although there was some overall improvement in detection rates as the viewing distance increased (Table 4) particularly for the large lesions, it was less marked than in the simple case (Table 3). Indeed, Observer 1 detected significantly more small lesions at 4 m than at 8 m when the two finer matrices were used.

Figure 8 shows that detection of large vertebral lesions using the 128 × 128 matrix at 8 m was always as good as, and often better than, under any other VC. The same was true for the small lesions when the results from all four observers were pooled together, but interobserver variability was such that Observer 1 recorded more success with the 128 × 128 and 256 × 256 matrices at 4 m than with the 128 × 128 at 8 m.

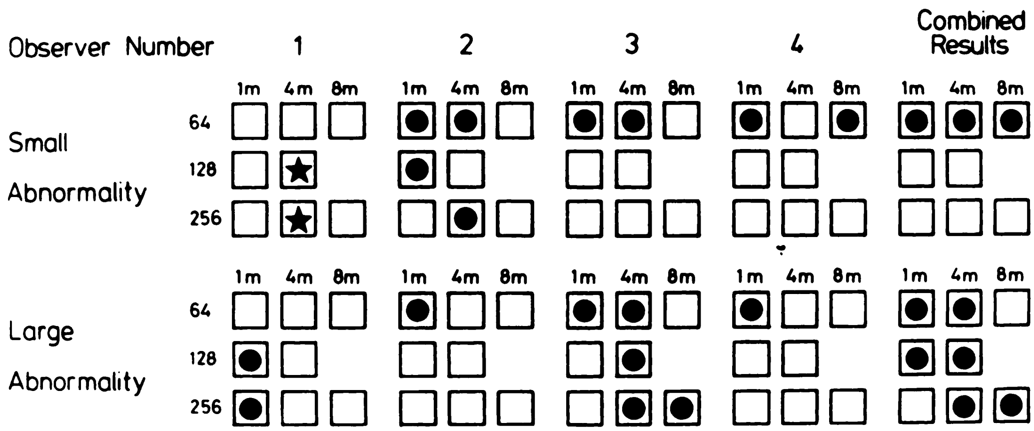

FIG. 8. Comparison of results from viewing 128 × 128 matrix from 8 m with other combinations of viewing distance and pixel size. Patterns were clinical bone images, as shown in Fig. 5. Symbols are as in Fig. 6.

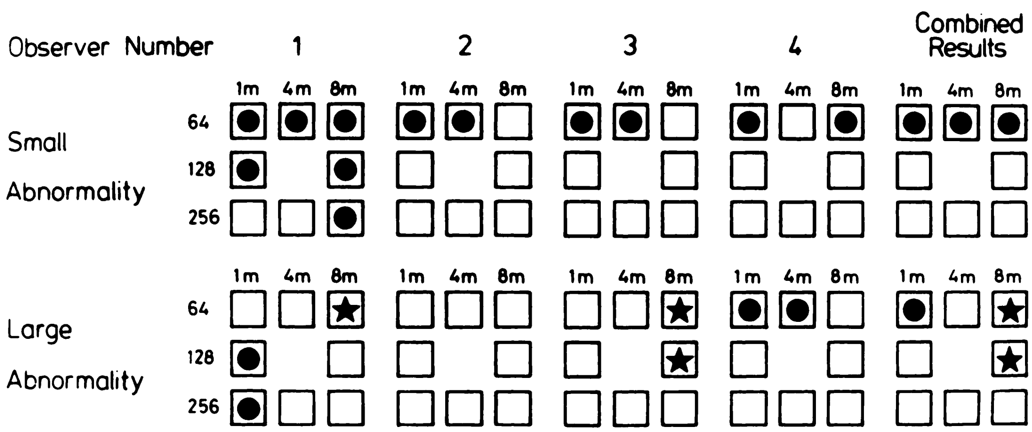

FIG. 9. Comparison of results from viewing 128 × 128 matrix from 4 m with other combinations of viewing distance and pixel size. Patterns were clinical bone images, as shown in Fig. 5. Symbols are as in Fig. 6.

It can be seen from Fig. 9 that the 128 × 128 matrix viewed from 4 m was adequate for the detection of small abnormalities. The large vertebral lesions, however, were perceived better by Observers 1 and 3 on the coarse 64 × 64 matrix at 8 m, and this is also the conclusion when the observers' results are pooled.

There is no reason to suppose that one particular viewing distance will be optimum for all tasks. When the comparisons of Figs. 8 and 9 are taken together, however, it appears that the results from the 128 × 128 matrix viewed from both 4 and 8 m could not be bettered by using a 256 × 256 matrix with any combination of viewing distances.

CONCLUSIONS

The principal advantage of using digitized images is the ability of observers to modify the display interactively. The disadvantage, however, is the possible loss of information caused by representing a continuously varying image by a discrete array of square pixels. So, whereas consideration of the physical factors limiting performance of a gamma camera suggests that a 128 × 128 digitization matrix is sufficiently fine for quantitative analysis of clinical radionuclide images, this size of matrix is thought to be too coarse for visual interpretation.

The results of this series of experiments show that reducing the angle subtended by the image at an observer's eye overcomes his subjective dislike of coarsely digitized images. Furthermore, an image 32 cm square digitized onto a 128 × 128 matrix and viewed from both 4 and 8 m has been found to be adequate for the detection of metastatic deposits even in images as finely detailed as those of the skeleton. In practice this arrangement is equivalent to looking at an image 8 cm square from 1 m, and then taking one pace backwards and viewing it from 2 m. By varying the visual angle of the image in this simple manner, a clinician is more likely to detect local areas of increased count density.

The results reported here, of course, apply only to unprocessed images, and whether or not image-filtering affects these conclusions remains to be seen. In the light of these experiments, however, digitization of unfiltered

TABLE 4. RELATIVE RATES OF DETECTION OF LESIONS IN CLINICAL BONE IMAGES (FIG. 5) AT VIEWING DISTANCES OF 1, 4, AND 8 m

Small lesions Matrix size	Observer number 1	2	3	4
		8		
			8	8 = 4
64	8 = 4 = 1	4		
			4 = 1	1
		1		
	4	8 = 4		
128			8 = 4 = 1	8 = 4 = 1
	8 = 1	1 = 4		
	4 = 1	8 = 1	1 = 8	
256				8 = 4 = 1
	8 = 1	4 = 1	4 = 8	

Large lesions Matrix size	Observer number 1	2	3	4
	8 = 1	8 = 4	8	8
64				
	4 = 1	1	4 = 1	4 = 1
	8 = 4		8 = 1	
128		8 = 4 = 1		8 = 4 = 1
	1		4 = 1	
	8 = 4			
256		8 = 4 = 1	8 = 4 = 1	8 = 4 = 1
	1 = 4			

Arrangement is explained in Table 3.

images onto matrices finer than 128 × 128 appears unnecessary for visual interpretation provided appropriate viewing conditions are adopted. The use of a 256 × 256 matrix does not produce better results, but does increase the complexity and cost of the imaging process.

ACKNOWLEDGMENTS

The authors are grateful for the advice of Dr. F. W. Campbell, at the University of Cambridge, and for the encouragement of Professor J. R. Mallard. Also, we appreciate willing participation of the following observers: Drs. H. G. Gemmell and M. Y. Norton, Ms. J. Davidson and H. Oliver, and Messrs. C. S. McGeechan and D. H. Temperton. One of the authors (WRP) is in receipt of a postgraduate research studentship from the Medical Research Council.

REFERENCES

1. POTTER R: Display systems for nuclear medicine images. In *Physical Aspects of Medical Imaging*. Moores BM, Parker RP, Pullan BR, eds. Chichester, John Wiley & Sons, 1981, 193–202
2. ROLLO FD, HARRIS CC: Factors affecting image formation. In *Nuclear Medicine Physics, Instrumentation, and Agents*. Rollo FD, ed. St. Louis, Mosby, 1977, 387–435
3. TODD-POKROPEK A: Image processing in nuclear medicine. *IEEE Trans Nucl Sci NS-27:* 1080–1094, 1980
4. DENDY PP, SHARP PF, KEYES WI, et al: Radionuclide emission imaging: Single-photon techniques including radiopharmaceutical developments. *Br Med Bull* 36:223–230, 1980
5. SHARP PF, CHESSER RB, MALLARD JR: The influence of picture element size on the quality of clinical radionuclide images. *Phys Med Biol* 27:913–926, 1982
6. HARMON LD, JULESZ B: Masking in visual recognition: Effects of two-dimensional filtered noise. *Science* 180: 1194–1197, 1973
7. CAMPBELL FW, ROBSON JG: Application of Fourier analysis to the visibility of gratings. *J Physiol* 197:551–566, 1968
8. CAMPBELL FW: The physics of visual perception. *Phil Trans R Soc Lond* [*B*] 290:5–9, 1980
9. PITT WR: The influence of digitisation on the quality of radionuclide images. In *Technical Advances in Biomedical Physics: NATO ASI Series*. Dendy PP, Ernst DW, Sengun A, eds. The Hague, Martinus Nijhoff: in press
10. SELTZER RA: Letter to the editor. *J Nucl Med* 5:76–77, 1964
11. MISHKIN FS, REESE IC, DOWELL JW: Advantages of producing a minified scan image. *Am J Roentgenol* 109: 682–685, 1970
12. BRAUNSTEIN P, HERNBERG JG, CHANDRA R: A practical compromise in bone scanning. *J Nucl Med* 12:639–640, 1971
13. BARBER DC: Reduction in scan time with minification. *J Nucl Med* 14:361–362, 1973
14. RAUF CG, SILBERSTEIN EB, KEREIAKES JG, et al: Bone scintigraph minification without equipment modification. *Int J Nucl Med Biol* 3:52–53, 1976
15. SMITH DB, UNDRILL PE, BOLTON MP: Optimising the performance of medical diagnostic displays by microprocessor assistance. In *Changes in Health Care Instrumentation due to Microprocessor Technology*. Pinciroli F, Anderson J, eds. Amsterdam, North-Holland Publishing Company, 1981, 119–126
16. ARMITAGE P: *Statistical Methods in Medical Research*. Oxford, Blackwell, 1971
17. WARREN RC, PANDYA YV: Effect of window width and viewing distance in CT display. *Br J Radiol* 55:72–74, 1982