Journal of Computer Assisted Tomography
9(4):835–838, July/August
© 1985 Raven Press, New York

Technical Note

Respiratory Ordered Phase Encoding (ROPE): A Method for Reducing Respiratory Motion Artefacts in MR Imaging

D. R. Bailes, D. J. Gilderdale, G. M. Bydder, A. G. Collins, and D. N. Firmin

Abstract: A method of reducing respiratory artefact when using the spin warp technique of magnetic resonance imaging is described. A respiratory signal is used to determine the order in which the rows of the data matrix are measured. The aim is to make the respiratory signal at the time of each phase encoding gradient a slowly varying function of the time integral of the gradient and hence of the row number. Data are collected during each cycle of the imaging sequence; thus the scanning time is not increased. **Index Terms:** Artifacts—Nuclear magnetic resonance, physics—Nuclear magnetic resonance, techniques—Nuclear magnetic resonance.

The artefacts due to respiratory motion when using the spin warp method of magnetic resonance (MR) imaging are well known (1,2). The artefacts often take the form of one or more ghost images of the moving structures of the patient. These ghost images are displaced by various amounts along the phase encoding direction with respect to the real image. The problem with the artefacts is that they degrade the parts of the image depicting structures that are not moving. It would be preferable if this could be avoided and only the sections of the image displaying the moving structures were degraded.

One method of reducing the artefacts is to collect the data only during the end expiratory phase of the respiratory cycle. This has been shown to work well, but it can lead to long scan times (1–3). Another method, which is described in this paper, is to use the respiratory signal to determine the order in which the data are collected.

THEORY

The data set collected using the spin warp method is the two-dimensional spatial frequency spectrum of the image (4–6) and hence the image is obtained by taking the two-dimensional Fourier transform.

Let the coordinates of the image matrix be x and y, those of the data matrix k_x and k_y, and the size of both matrices N_x by N_y. The integers x, y, k_x, and k_y are all dimensionless with the ranges $x = -N_x/2 \ldots (N_x/2 - 1)$, $y = -N_y/2 \ldots (N_y/2 - 1)$, $k_x = -N_x/2 \ldots (N_x/2 - 1)$, and $k_y = -N_y/2 \ldots (N_y/2 - 1)$. Let a row of the data matrix be the points with constant k_y and varying k_x. To measure a given row, an x gradient is applied during the data collection that is the same for all rows; this gradient is called the frequency encoding gradient. In addition, a y gradient with a time integral proportional to k_y is applied before the data collection; this gradient is referred to as the phase encoding gradient. The time between measuring rows is the repetition time of the sequence, but each row is measured in a time that is short compared with the respiratory cycle. Thus there is appreciable respiratory motion in the y direction between the rows of the data matrix, but relatively little in the x direction between columns of the matrix. Hence the motion artefact is much greater in the phase encoding direction than it is in the frequency encoding direction.

Consider a point in the body with a y coordinate $y(t)$ that varies in time due to respiration. Let y at the time when the phase encoding gradient is applied to measure row k_y of the spectrum be denoted by $y(k_y)$. If the rows of the spectrum are measured in numerical order, $y(k_y)$ is a periodic function. This periodicity tends to produce ghosts of the moving

From Picker International, Hirst Research Centre, Wembley, Middlesex, England (D. R. Bailes, D. J. Gilderdale, and A. G. Collins); the Department of Diagnostic Radiology, Royal Postgraduate Medical School, Hammersmith Hospital, London, England (G. M. Bydder); and the National Heart Hospital, London, England (D. N. Firmin). Address correspondence and reprint requests to D. R. Bailes at Picker International, Hirst Research Centre, East Lane, Wembley, Middlesex HA9 7PP, England.

structures of the body in the image. Alternatively, if the rows are measured in a random order, $y(k_y)$ is random, and this smears out the moving structures of the body in the y direction. Our goal is to confine the effect of the motion to the sections of the image depicting moving parts. To achieve this, it seems reasonable that we should aim to make $y(k_y)$ a slowly varying function—for example, monotonic. In practice we cannot measure $y(t)$ for all moving points in the body. However, if we measure some quantity $r(t)$ that is a single valued function of all $y(t)$, then if $r(k_y)$ is monotonic, all $y(k_y)$ will be monotonic. Because of the complex movements associated with respiration, it is probably impossible to find a quantity that is a single valued function of all $y(t)$. However, even if the strict single valued relationship does not hold, all $y(k_y)$ are still slowly varying functions. A method for making $r(k_y)$ as near to monotonic as possible is described below, but, because of the variations in the respiratory cycle, $r(k_y)$ probably will not be exactly monotonic.

During the scan, and just before each phase encoding, r is measured. Since r is recorded digitally, it is considered a discrete variable. On the basis of this measurement the most appropriate row of the spectrum to be measured is selected, and the corresponding phase encoding gradient is used. To make this choice, some information about the respiratory signal is needed. In this work the probability distribution $p(r)$ of the signal is used, and this is measured before the scan. The spatial frequency spectrum has to be measured over a range of k_y. Let each value of r be "responsible" for a range of k_y. Since we want $r(k_y)$ to be monotonic, the ith value of r, r_i, should be responsible for the ith range of k_y, δk_{yi} (see Fig. 1). Since we want to measure all the rows of the spectrum during the scan, the probability of each value of k_y being chosen should be the same and therefore equal to $1/N_y$. That is $p(r_i)/\delta k_{yi} = 1/N_y$, or $\delta k_{yi} = p(r_i)N_y$. If r is measured to be r_i, the first choice of k_y is the nearest integer to the centre of δk_{yi}. For every value of r, we have

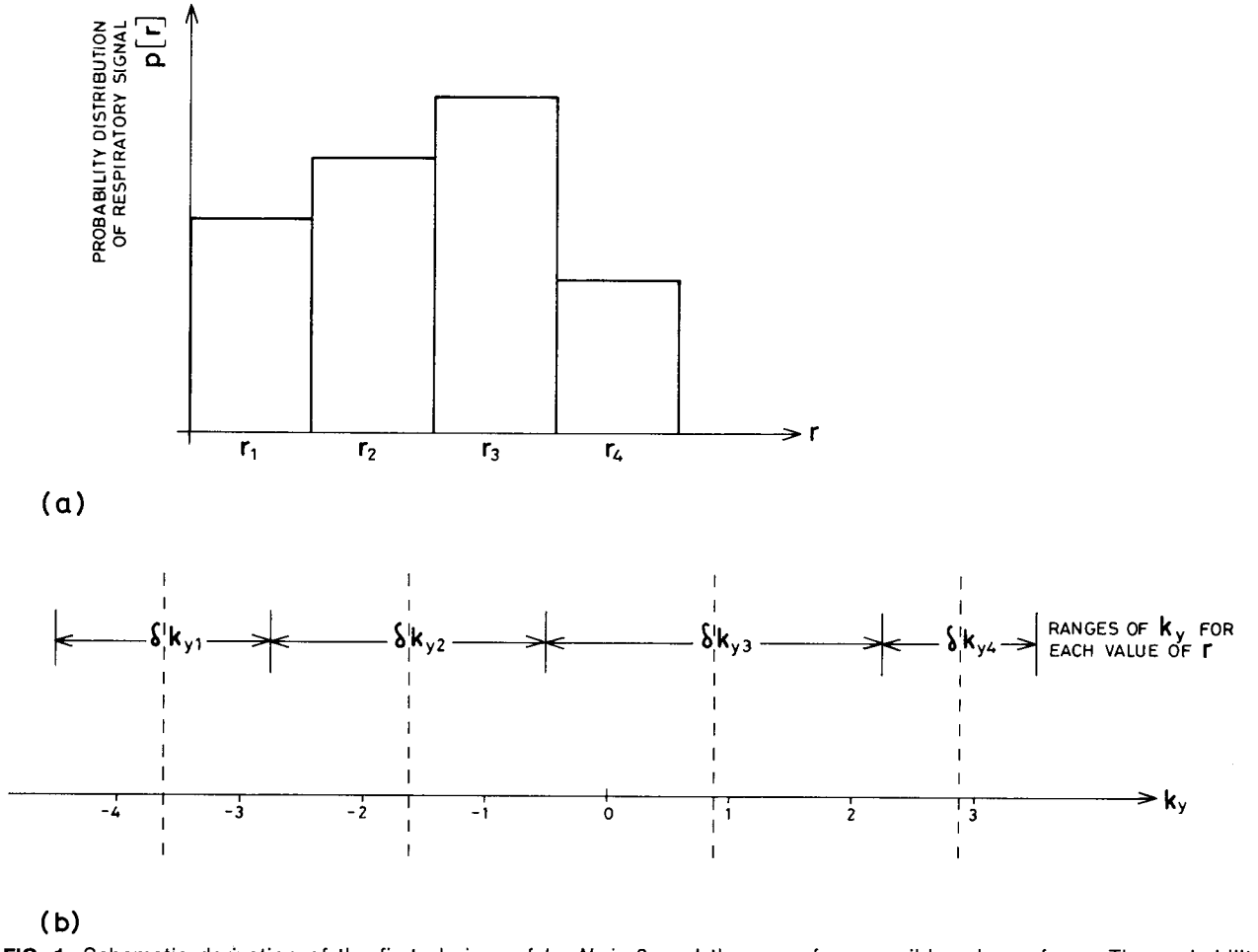

FIG. 1. Schematic derivation of the first choices of k_y. N_y is 8, and there are four possible values of r. **a:** The probability distribution of the respiratory signal. **b:** Derivation of the first choices of k_y. $\delta k_{yi} = p(r_i)N_y$. It can be seen that the first choices of k_y are -4, -2, 1, and 3.

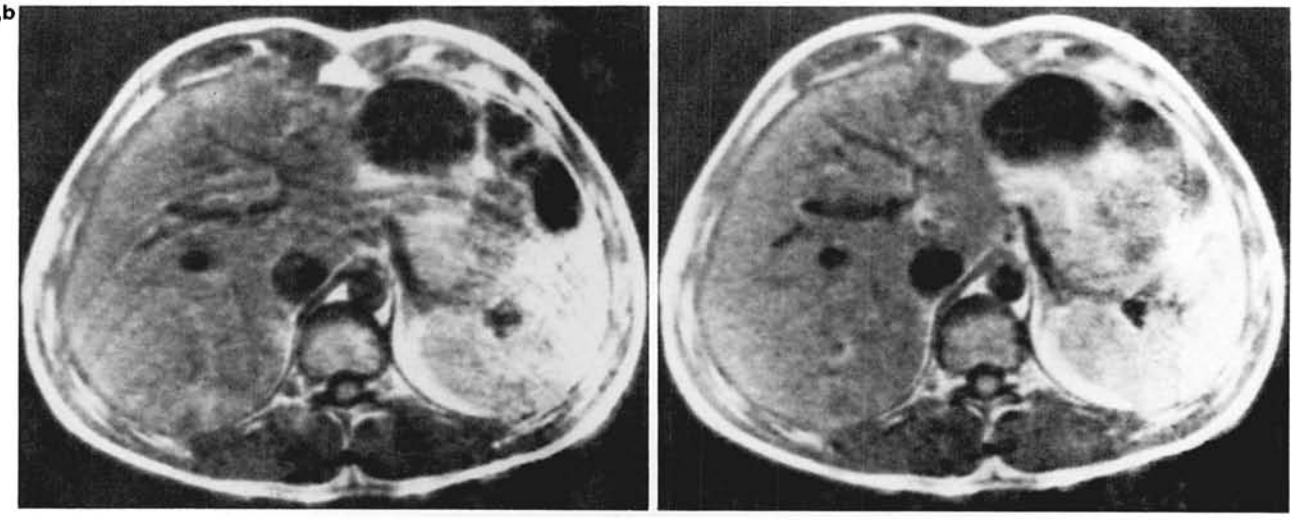

FIG. 2. Normal volunteer imaged with a spin echo (SE: 544–44) sequence. In **(a)** the rows of the spatial frequency spectrum are measured in numerical order. In **(b)** the respiratory ordered phase encoding technique is used. There is considerable reduction in the respiratory artefacts in (b).

a first choice of k_y. If this row has already been measured, the nearest row that has not been previously measured is taken. If the spatial frequency spectrum is measured a number of times, the spectrum is measured at all values of k_y before proceeding to the next measurement of the spectrum.

The above description has only considered the spin warp method. However, the technique could be applied to other two-dimensional data collection schemes, for example the "reconstruction from projections" method.

IMPLEMENTATION

The quantity $r(t)$ is obtained as follows. A light vertical rod is placed on the patient's anterior thoracic or abdominal wall near the slice being measured. The rod is connected to a transducer that is supplied with a constant intensity light source via a fibre-optic cable. The intensity of the light source is modulated by the y coordinate of the rod to produce a light intensity that varies monotonically with the y coordinate of the rod. Since this transducer contains no metal, it can be placed in the MR scanner without any adverse effects on the image. The output light intensity is fed via a fibre-optic cable to another transducer that produces a voltage that is a monotonic function of the light intensity. This voltage is used as the quantity $r(t)$ and it is digitised using an 8 bit analogue to digital converter. The probability distribution $p(r)$ of the respiratory signal is determined by measuring the respiratory signal at 10 ms intervals for 40 s, and calculating $p(r)$ by equating each $p(r_i)$ to the number of times r was measured to be r_i divided by the total number of measurements.

An example of the efficacy of the technique is shown in Fig. 2. Both spin echo images (SE: 544–44) are of a normal volunteer. For Fig. 2a the rows of the spatial frequency spectrum were measured in numerical order; for Fig. 2b the above technique of respiratory ordered phase encoding (ROPE) was used. There is considerable reduction in the respiratory artefact in Fig. 2b.

DISCUSSION

The advantage of the ROPE technique over methods of respiratory gating is that the actual scan time is not increased. A problem with the present implementation of the technique is that the probability distribution $p(r)$ of the respiratory signal, and hence the first choices of k_y, are determined only before the scan. However, $p(r)$ may well change during the scan, and this will degrade the efficacy of the technique. Therefore, it would be better to continue making frequent measurements of r during the scan and modifying the first choices of k_y if necessary.

Acknowledgment: We are grateful for the support given to us by the MRC and the DHSS and in particular by Mr. G. R. Higson and Mr. J. L. Williams.

REFERENCES

1. Runge VM, Clanton JA, Partain CL, James AE. Respiratory gating in magnetic resonance imaging at 0.5 Tesla. *Radiology* 1984;151:521–3.
2. Schultz CL, Alfidi RJ, Nelson AD, Kopiwoda SY, Clampitt

ME. The effect of motion on two-dimensional Fourier transformation magnetic resonance images. *Radiology* 1984;152:117–21.

3. Ehman RL, McNamara MT, Pollack M, Higgins CB. Respiratory gated MRI: evaluation of technical approaches. In: *Scientific program and book of abstracts. Society of Magnetic Resonance in Medicine. Third annual meeting August 13–17, 1984, New York, NY.* Berkeley CA: Society of Magnetic Resonance in Medicine, 1984:210–1.
4. Kumar A, Welti D, Ernst RR. NMR Fourier zeugmatography. *J Mag Res* 1975;18:69–83.
5. Edelstein WA, Hutchison JMS, Johnson G, Redpath T. Spin warp NMR imaging and applications to human whole-body imaging. *Phys Med Biol* 1980;25:751–6.
6. Twieg DB. The k-trajectory formulation of the NMR imaging process with applications in analysis and synthesis of imaging methods. *Med Phys* 1983;10:610–21.