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ALIASING ARTIFACTS IN MR IMAGING

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Abstract—Aliasing artifacts occur in the phase encoding direction when the dimensions of the imaged object exceeds the field of view. Signal generated from outside the field of view appears as a superimposed object at the opposite edge of the image. Increasing the field of view, changing the gradient axes relative to the patient, or use of surface coils can reduce aliasing and are parameters which are controlled by the radiologist/technologist. The manufacturer may provide software packages which exploit two additional strategies, either limiting the volume of the patient from which the MR signal is acquired as in Inner Volume Imaging or display of only the central portion of the acquisition field of view. Increasing the field of view will decrease spatial resolution unless the number of phase encoding steps is increased at a cost of increased acquisition time. The radiologist may in some clinical situations choose to tolerate aliasing in favor of improved resolution in the area of interest and decreased acquisition time.

Image artifacts   Image quality   MR technology   MR image processing

INTRODUCTION

Each new imaging modality introduces a new set of artifacts which need to be recognized. Motion artifacts [6, 10], truncation artifacts [4, 11], and chemical shift artifacts in MR imaging [1, 7, 9] have been examined in recent publications. Aliasing artifact appears only in the phase encoding axis of current MR imaging systems and can be directly influenced by the radiologist/technologist's choice of image parameters. Prior studies of aliasing have dealt with minimizing artifacts in the frequency encoding axis [5]. Aliasing artifacts in the phase encoding axis have not previously been examined in the literature. Aliasing artifact appears as a wrap-around effect. Structures at one edge of the imaged plane along the phase encoded axis are superimposed over structures at the opposite edge in the final image. A typical example occurs when the patient's nose extends outside the field of view and is superimposed over posterior structures (Fig. 1). Aliasing artifact is normally easily recognized and only in unusual cases may be confused with pathology. However, its significance lies in the limitations it imposes on imaging parameters. In this study, we discuss the mechanism of aliasing, its behavior with varied fields of view, and methods to minimize the artifact.

METHODS

Patient and phantom studies were obtained on a 0.5 T superconducting imager (Picker International, Cleveland, Ohio). Spin echo pulse sequences were used for all studies. The MR images were acquired and reconstructed with a 2 D fourier transform technique and produced planar images. A nonuniform phantom consisting of a plexiglass cylinder filled with 0.2 mM NiCl with a diameter of 20 cm was imaged with a spin echo pulse sequence of TE 40 ms and TR 500 ms. Images were obtained with varying number of phase encoding steps (64, 128, 256), fields of view (25, 20, 17, 15 cm), and signal averages (1, 2). Low pass digital filters to remove high spatial frequency information currently used in most MR systems were left in place.

RESULTS

In the phantom studies, aliasing artifacts occurred only in the phase encoding direction when the diameter of the object exceeded the field of view (FOV). Signal originating from outside the FOV was

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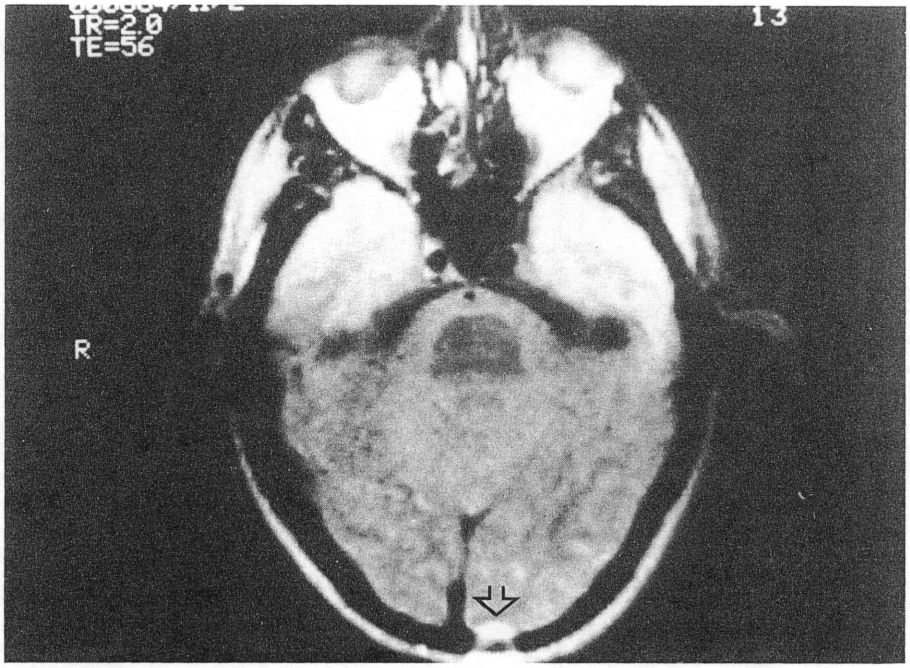


Fig. 1. Axial images of the brain in which the patient's nose extends outside the field of view and is superimposed on the posterior edge of the image (open arrow).

superimposed over the image at the opposite edge of the field. When the FOV encompassed the entire phantom of 20 cm, aliasing artifact did not occur (Fig. 2A). Reduction of the viewing field to 17 cm caused aliasing to occur along the phase encoding axis (Fig. 2B). Smaller FOV caused increasing degree of aliasing artifacts in the image (Fig. 2C).

Changing the number of applications of the phase encoding gradient used in the 2D-FT reconstruction did not affect the degree of artifact in the final image (Fig. 3) when the FOV remained constant. Increasing the number of phase encoding steps improved the spatial resolution of both the image and the artifact. An increase in the FOV necessary to remove aliasing without a concomitant increase in the number of applications of the phase encoding gradient decreased spatial resolution.

The effect of aliasing artifact on imaging of the spine was examined. A desire to limit the superimposed images caused by aliasing artifact usually limits the choice of phase axis to the anterior–posterior direction. The full diameter of the patient in this plane may then be included in the FOV. It is well recognized that respiratory motion causes image harmonics in the phase encoding direction [11] and causes the most image degradation when the phase axis is in the direction of motion. Hence, motion artifacts may significantly degrade the diagnostic quality of a sagittal image of the lumbar spine. Aliasing artifacts are eliminated at the cost of decreased image quality in the area of interest due to motion artifacts. We compared the diagnostic quality of sagittal views of the lumbar spine with the phase encoding axis in the cranio-caudad and in the anterior-posterior direction (Fig. 4). Images obtained with the cranio-caudad phase axis showed improved visualization of nerve roots within the neuroforamina and improved image quality in the area of the posterior border of the lumbar discs despite aliasing artifacts in other parts of the image. Respiratory motion artifact remained primarily confined to the portion of the image anterior to the spinal column.

The use of surface coils allows the volume of the patient producing the MR signal to be reduced. Signal and noise generated deep to the surface coils is not received and does not impinge on the final image. A narrow FOV may then be used without causing aliasing artifact to allow greater spatial resolution (Fig. 5).

DISCUSSION

Explanations of aliasing artifacts as previously described in signal processing theory have focused on the mechanism by which high frequency information may masquerade as low frequency informa-

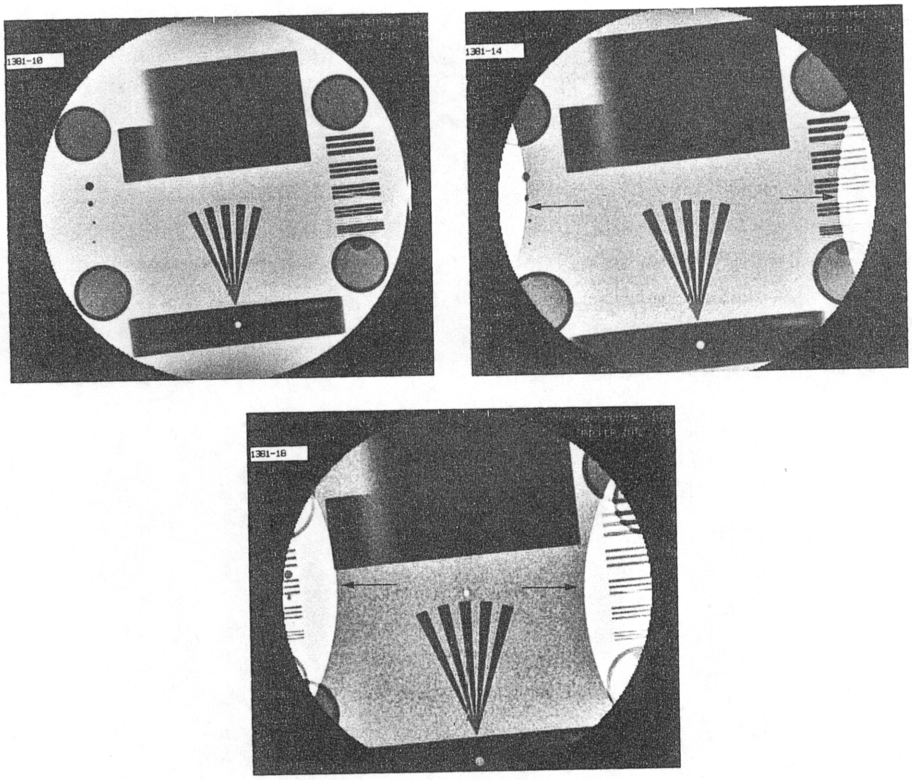


Fig. 2. A 20 cm diameter phantom is imaged with varying fields of view. Aliasing artifact does not occur when the field of view, 20 cm (A) is equal to or greater than the diameter of the phantom. Increasing amount of image wraparound or aliasing is demonstrated in the phase axis when the diameter of the phantom exceeds the field of view, 17 cm (B) and 15 cm (C).

tion due to a low data sampling rate [2]. In the example of a sinusoidal wave a high sampling rate would be required to adequately describe this continuous wave by a finite set of numbers. If the sampling interval were too long the peaks and troughs of the wave would not be accurately described. At a sampling rate less than twice the frequency of the sine wave, the wave function would be incorrectly assigned to a lower frequency. The nyquist limit is defined as the highest frequency wave that can be unambiguously represented by a given sample rate. Therefore, the nyguist limit is the highest frequency at which each wave is sampled twice. If we arbitrarily designate this limit to be N, then any wave with frequency greater than N, e.g., $N + f$, will be considered by the fourier transform

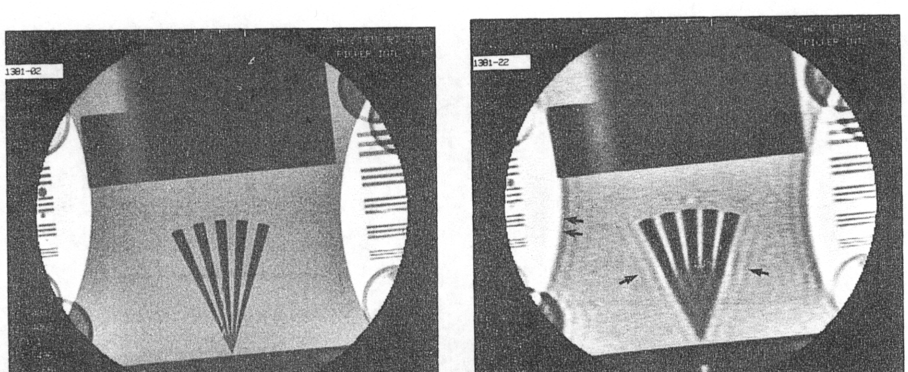
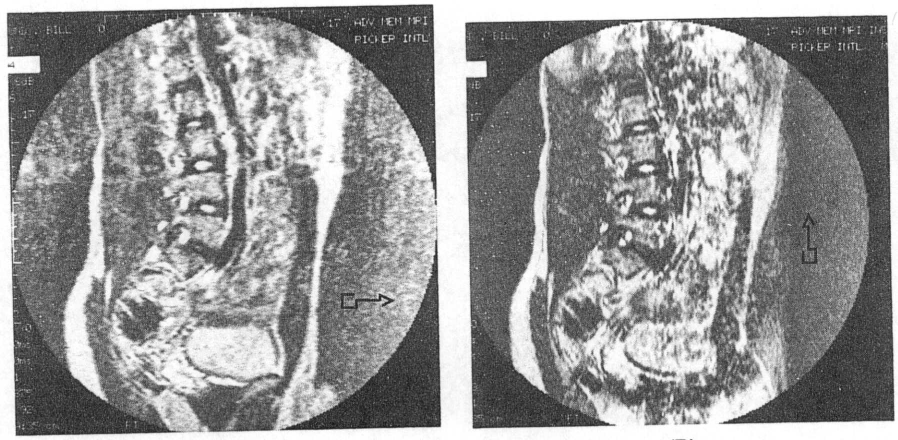


Fig. 3. Aliasing artifact (double arrows) is unchanged by change in number of phase encoding steps, 128 levels (A) and 256 levels (B) when FOV is constant. Note prominent boundary artifacts due to truncation errors (ref. 3, 4) are more prominent with decreased number of phase encoding steps (single arrows).



(A) (B)

Fig. 4. Sagittal images of the spine (TE 80 ms, TR 3000 ms). images obtained with phase axis in the antero-posterior direction or cranio-caudad direction as indicated by arrows. (A) Respiratory motion artifact degrades image quality; however, aliasing artifact does not occur. (B) Image obtained at same level as (A). Improved visualization of the nerve roots within the neuroforamina. Respiratory motion artifact is confined to the compartment anterior to the spinal column. Aliasing artifact occurs which degrades image of pelvic structures but does not interfere with evaluation of the lumbar spine.

to have a frequency of $N - f$. In MR spectroscopy, aliasing occurs when sampling includes components of the signal that have a higher frequency than the nyquist limit. This part of the signal is then folded back into the spectrum at a lower frequency. Appropriate filters must be used to eliminate this high frequency data. In discrete fourier transforms of a finite number of data points, as in MR image reconstruction, the transform is periodic with a period of $X = 1/\Delta t$, where Δt is the sampling interval,

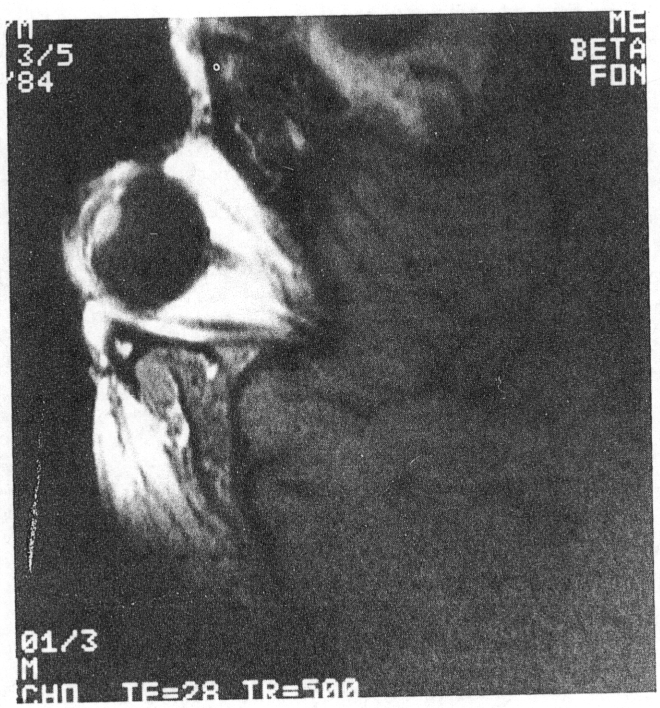


Fig. 5. Image of orbit obtained with a surface coil. Application of a surface coil allows a smaller field of view to be used without causing aliasing artifact.

and $X/2$ is the nyquist limit. When the sampling interval is large the period, X, is decreased and overlapping repetition of the tranform occurs. If the sampling interval is too large, or the filter pass band is too wide, the final image will include information from the overlapping copies of the image. In the frequency encoding direction, this problem can be dealt with by a method suggested by McVeigh, et al. [5]. The signal is oversampled in the frequency encoding direction by a factor of 2 and a temporal filter is used that excludes frequencies $df = (2X$ nyquist) − (image bandwidth/2). This does not cause appreciable increase in scanning time since the increase in sampling is achieved by decreasing the time interval between samples. This or similiar methods are used to eliminate aliasing in the frequency encoding axis and the artifact does not appear in this axis on current MR scanners. In the phase encoding direction [8], Δt is the magnitude of change in the intensity with each succeeding application of the phase encoding gradient. Where Δt is small, the repetitions of the transform will be separated and aliasing does not occur. Decrease in Δt will increase FOV. If Δt is held constant to prevent aliasing and the number of phase encoding steps is decreased there will be a decrease in range of gradient intensity spanned by the gradient pulses and resolution will decrease. In phantom images (Fig. 3) changing the number of phase encoding steps at constant FOV changed image resolution but did not affect aliasing artifact. If the range of gradient intensity is held constant so that resolution will be unchanged but the number of phase encoding steps is decreased Δt will increase causing decrease in FOV and increase in the period of repetition of the transform and aliasing will occur. In the phase encoding axis if the intensity increment between phase encoding steps, Δt, is decreased to eliminate aliasing, then the number of phase encoding steps must be increased to keep resolution constant at a cost of increased acquisition time. For this reason aliasing artifact remains a problem in the phase encoding axis.

Note that because of the periodicity of the transform, aliasing artifacts arise at the edges of the viewing field. In cases where the interesting pathology can be localized in the central portion of the MR image, aliasing will not hamper visualization of any lesions. Aliasing can be tolerated in such instances, and no measures need to be taken which may affect spatial resolution or scanning time. Furthermore, since aliasing occurs only in the phase encoding axis, the gradient axes may be reoriented so that aliasing occurs in portions of the image which are relatively unimportant in the image evaluation. A more appealing image in these cases may be produced by displaying only the central portion of the image. The limitations of increased scanning time to preserve image resolution remain although the displayed FOV is smaller. The radiologist needs to be aware when this method is used by manufacturers as the displayed and actual FOV used in the reconstruction process may not be the same.

If an object can be imaged using surface coils, then aliasing may be avoided simply by using these coils. Surface coils however are limited to superficial tissues. Inner volume imaging as described by Feinberg, et al. [3] allows restriction of the imaged volume to the deep tissues without producing aliasing artifact. The 180° pulse in a spin echo pulsing sequence is applied perpendicular to the plane of the initial 90° pulse. Only the volume of tissue in the intersection of these two planes receives both the 90° and 180° pulses and produces a signal which is rephased at the time of signal acquisition TE. As in the case of the surface coils, only the MR signal from this small volume of tissue is used in image reconstruction, and aliasing does not occur despite a narrow FOV.

Hence, a number of methods are available to reduce the impact of aliasing artifacts on image evaluation. The choice of method will vary from subject to subject and among radiologists/technologists monitoring MR studies.

CONCLUSION

Aliasing artifacts in MR imaging occur in the phase encoding direction when the dimension of the object exceeds the image FOV. Signal generated from outside the FOV is superimposed on the opposite edge of the field. Aliasing artifact may be reduced or eliminated by increasing the FOV, changing the gradient axes relative to the patient, displaying only the central portion of the actual FOV, use of surface coils, or inner volume imaging techniques. Tolerating aliasing artifacts may be prefereable in some cases when accompanied by improved image quality within the area of clinical interest.

REFERENCES

1. E. E. Babcock, L. Brateman, J. C. Weinreb, S. D. Horner and R. L. Nunnally, Edge artifacts in MR images: chemical shift effect, *J. Comput. Assist. Tomogr.* **9,** 252 (1985).
2. R. N. Bracewell, The fourier transformation and its applications. McGraw-Hill, New York (1978).
3. D. A. Feinberg, J. C. Hoenninger, L. E. Crooks, L. Kaufman, J. C. Watts and M. Arakawa, Inner volume MR imaging: technical concepts and their application, *Radiology* **156,** 743 (1985).
4. R. Lufkin, E. Pusey, D. Stark, R. Brown, B. Leikind and W. Hanafee, Boundary artifact due to truncation errors in magnetic resonance imaging, *Am. J. Roentgenol.* **147,** 1283 (1986).
5. E. R. McVeigh, R. M. Henkelman and M. J. Bronskill, Noise and filtration in magnetic resonance imaging, *Med. Physics* **12,** 586 (1985).
6. C. L. Schultz, R. J. Alfidi, D. Nelson, S. Y. Kopiwoda and M. E. Clampitt, The effect of motion on two dimensional fourier transformation magnetic resonance images, *Radiology* **152,** 117 (1984).
7. K. P. Soila, M. Viamonte Jr. and P. M. Starewiez, Chemical shift misregistration effect in magnetic resonance imaging, *Radiology* **153,** 153 (1984).
8. V. J. Wedeen and Y. S. Chao, Rapid three-dimensional angiography with undersampled MR imaging, *J. Comput. Assist. Tomgr.* **11,** 24 (1987).
9. J. C. Weinreb, L. Brateman, E. E. Babcock, K. R. Maravilla, J. M. Cohen and S. D. Horner, Chemical shift artifact in clinical magnetic resonance images at 0.35 T, *Am. J. Roentgenol.* **145,** 183 (1985).
10. M. L. Wood and R. M. Henkelman, MR image artifacts from periodic motion, *Med. Phys.* **12,** 143 (1985).
11. M. L. Wood and R. M. Henkelman, Truncation artifacts in MR imaging, *J. Mag. Res. Med.* (in press).

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