Automated analysis of multiple performance characteristics in magnetic resonance imaging systems

Michelle M. Covell,[a] David O. Hearshen,[b] Paul L. Carson,[c] Thomas P. Chenevert, Paul Shreve, Alex M. Aisen, Fred L. Bookstein,[d] Brian W. Murphy, and William Martel
Department of Radiology, School of Medicine, The University of Michigan, Ann Arbor, Michigan 48109

(Received 26 August 1985; accepted for publication 30 July 1986)

As with other digital imaging systems in heavy medical use, it is desirable with magnetic resonance imaging (MRI) to obtain extensive, rigorous system performance measures from a small set of images of one or two relatively simple test objects. Digital analysis of the parallel square rod (PSR) test object introduces digital image system self-evaluation to MRI and extends automated image evaluation to include rigorous measures throughout the imaging volume rather than just average measures over the image. Precise comparisons with theory and between systems can be performed as well as quality control and corrections for nonuniformities. The PSR test object consists of an $18 \times 18 \times 36$ cm rectangular acrylic container enclosing 60 parallel square acrylic rods running the entire length. The inter-rod space is filled with a liquid or gel that produces strong, tissuelike signals in MRI and high contrast relative to the rods for computed tomography (CT). For profiles of slice thickness and separation, the rods are tilted in the test object to intersect the image plane at a 45° angle when the test object sides are parallel and perpendicular to the image plane. The test object itself is rotated 6–12° about its major axis so that the sides of the rods make a small angle to the rows and columns of pixels. This allows digital sampling at finer spacing than the pixels for determination of edge response functions. Over the 25–49 blocks in each slice of the imaged volume, maxima, minima, mean values, variances, and ratios currently are reported for the following variables: signal-to-noise ratio and sensitivity, linear and nonlinear image distortion, full width at half maximum (FWHM) resolution of the point spread function (PSF), slice separation, and slice thickness. These performance values at each rod or edge are displayed as gray scale functional images. Individual rod values are recorded and plotted as histograms and profiles. Results of the automated analysis for MRI system examples are in good agreement with expectations from theory and more manual tests.

INTRODUCTION

System performance of magnetic resonance (MR) scanners depends on many factors. The performance parameters can be classified into two groups: (1) those that depend upon the digital nature of the acquired signals and thus are similar to other imaging modalities such as computed tomography (CT), and (2) those which are dominated by magnetic resonance effects. Since MR images depend jointly upon these parameters in a complicated way, it is necessary to develop a set of standards and rigorous spatial measurements that will characterize image quality. These measurements should also allow for direct comparison of results from different MRI or CT scanners.

Quantification of MR images involves calculation of tissue specific characteristics: T_1, T_2, and density of the imaged nuclei. The signal intensities used to calculate these quantities depend upon the characteristics of the machine as well as the choice of a model for relaxation. It is important to separate machine dependence from inadequacies of the relaxation model when comparing relaxation times within the imaging volume and between machines. Prior investigations[1–8] have shown that variation in T_1 and T_2 measurements is influenced by inaccurate flip angles, motion and/or flow, and inadequacies of the models used to obtain the values. Several authors have shown that imperfect slice definition can lead to errors in T_1 of up to 100%.[2,3] Others have shown that errors of similar magnitude can occur when the rf field is nonuniform across the image plane or from slice to slice.[5,8] Diffusion during long echo delay times is also a source of error, which is dependent upon field gradients as well as echo time.[9] In addition to these effects, imperfect selective 180° pulses can lead to errors in intensity in multiple-echo techniques.[10,11] While analysis of the test object does not result in a direct measure of the accuracy of T_1 and T_2 calculated values, the analysis does provide information about parameters such as rf nonuniformity and slice definition which indirectly affect the accuracy of relaxation time calculations or any other quantitative measures.

Since the acrylic rods are much denser than the filling material, the parallel square rod (PSR) test object works well for CT as well as MRI, although the analysis program has not yet been run on CT images.

Automated analysis

Earlier system performance measurements in MRI have been accomplished with a variety of test objects using methods that were laborious and subjective.[1,12–14] Further complicating standardization of system parameters is lack of consistency in the definition of signal-to-noise ratio (SNR),[15–17] and normalization of signal when there are inaccuracies in slice thickness.[2,3] More importantly, in most approaches a complete set of quality control (QC) tests re-

 0094-2405/86/060815-09$01.20

TABLE I. Summary of performance evaluation measurements; 18×36 cm cross section.

Measurement	Slice 30		Slice 31[a]		Slice 32		Slice 33	
	Mean	*s*	Mean	*s*	Mean	*s*	Mean	*s*
HSB[b]	762.1	45.0	752.1	42.5	752.3	36.0	749.5	39.0
LSB[c]	47.0	2.5	44.7	4.2	47.2	2.7	49.3	3.7
Anisotropy[d]	1.02	0.009	1.03	0.018	1.02	0.011	1.02	0.010
Resolution[e]	1.62	0.20	1.62	0.18	1.60	0.20	1.63	0.27
Slice[e] thickness	5.71	0.36	5.78	0.32	5.73	0.26	5.72	0.32
Slice[e] separation			10.07	0.18	10.16	0.21	10.28	0.24

[a] Slice 31 is the center slice of 20.
[b] High signal block SNR (ratio of the mean signal to the standard deviation of pixel values in a block).
[c] Low signal block SNR.
[d] Ratio of principal axes (see text).
[e] In millimeters.

automation include lateral displacement of images relative to others in the same sequence, Gibbs phenomenon and other ringing artifacts, spatial variation of noise in the LSB's or standard deviation in the HSB's (these are computed but not displayed separately at present), and centering and orientation of the various system components relative to each other.

Because this program uses a small amount of image display software and hardware that is system dependent available in the Electrical Engineering and Computer Science Department, it is not easily transportable to other systems. In addition, the edge detection algorithm may be inadequate when there are large distortions or low SNR in portions of the image. These two problems are currently being addressed and a more portable version is expected shortly.

CONCLUSIONS

Once quality control and higher performance evaluation techniques of sophistication are developed, they rarely are utilized routinely in medical imaging with the possible exceptions of scintillation imaging and film processing. In most radiologic imaging, technologists and service personnel do not perform and interpret adequate tests. Neither they nor the physicists usually have time for laborious complete performance tests or less complete routine tests and when these are performed there is neither standardization nor objectivity in the interpretation. Quality control of film processing is one example where there is an immediate return for the department, and this has been strongly supported by film suppliers.

Automated performance evaluation of images with the maximum number of relevant tests per image sequence would solve most of these problems. With improvements in computer technology it will soon be practical to perform complete analyses of test images routinely, and to automatically flag deviations from the norm. The test object and software presented here are directed toward that goal. With the described digital analysis of the edges and statistical analysis of multiple performance characteristics on a single test pattern, it is possible for a single-image sequence to replace the multiplicity of image acquisitions with various test sections or inserts as required in most CT and MR phantoms. Certainly test images should be acquired in a variety of conditions, such as coronal or oblique planes to verify proper system functioning, but the variety of tests would be much larger with most other phantom configurations, and not all tests would need to be performed at the maximum frequency.

Measurements also are made at many points throughout

TABLE II. Summary of performance evaluation measurements; 18×36 cm cross section.

Measurement	Slice 20		Slice 21[a]		Slice 22		Slice 23	
	Mean	*s*	Mean	*s*	Mean	*s*	Mean	*s*
HSB[b]	303.4	76.1	294.2	66.8	290.9	61.5	296.2	69.0
LSB[c]	18.0	2.3	21.3	3.2	16.1	1.4	18.5	2.3
Anisotropy[d]	1.07	0.048	1.06	0.044	1.06	0.031	1.07	0.060
Resolution[e]	1.61	0.20	1.59	0.19	1.60	0.17	1.61	0.19
Slice[e] thickness	6.35	0.45	6.21	0.37	6.28	0.36	6.23	0.38
Slice[e] separation			10.12	1.62	9.53	3.22	9.85	3.62

[a] Slice 21 is the center slice of 20.
[b] High signal block SNR (ratio of the mean signal to the standard deviation of pixels in a block).
[c] Low signal block SNR.
[d] Ratio of principal axes (see text).
[e] In millimeters.

the imaging volume. While the utility of this feature is obvious for establishing the useful field of view for coils, via signal levels or SNR's, and for in-plane image distortion, it is also useful for more complex distortion such as image plane tilting and thickness variation. Other than statistical precision, the value of multiple resolution measurements is yet to be established. Another important feature of this general approach is that it provides a more objective, reproducible measurement than does visual analysis. Only one test pattern is available in the phantom for most measures, thus increasing the likelihood that readily comparable measurements will become available on numerous and various scanners.

Perhaps the main value of the test methods, in addition to providing the otherwise laborious measurements of spatial distortion and spatial variation in signal or relaxation times, is that they provide an objective, reproducible measurement of resolution, slice thickness, and separation, at each location and averaged for each plane and with adequate statistics for high precision.

ACKNOWLEDGMENTS

Helpful information and feedback were provided by Dr. Howard Guenther, Dr. Joel Blank, and Ron Schilling of the Diasonics Corporation. Initial phantom tests and data acquisition were performed at the Huntington Medical Research Institute with the kind assistance of Dr. William Bradley. Dr. Charles Meyer provided guidance and many helpful ideas. Mr. Quentin Westrick very skillfully constructed the first few models of the PSR test object. This work was supported in part by the Diasonics Corporation and Grant No. PDT 242 from the National Cancer Society.

[a] Present address: Department of Electrical Engineering, Massachusetts Institute of Technology, Cambridge, MA.

[b] Present address: Department of Radiology, Henry Ford Hospital, Detroit, MI.

[c] Address correspondence to Paul L. Carson, Ph.D., University of Michigan Medical Center, 1500 E. Medical Center Drive, Radiology Department, Box 0030, Room XS2315, Ann Arbor, MI 48109, tel. (313) 763-5884.

[d] Present address: Center for Human Growth and Development, The University of Michigan, Ann Arbor, MI.

[1] L. Brateman, L. W. Jennings, and R. L. Nunnally, in Scientific Program, Society of Magnetic Resonance in Medicine, Third Annual Meeting, New York, 1984, pp. 98–99.

[2] D. J. Bryant, J. A. Payne, D. R. Bailes, and I. R. Young in Scientific Program, Society of Magnetic Resonance in Medicine, Third Annual Meeting, New York, 1984, pp. 109–110.

[3] I. L. Pykett, B. R. Rosen, F. S. Bunonamo, and T. J. Brady, Phys. Med. Biol. **6**, 723 (1983).

[4] R. L. Ehman, R. O. Kjos, R. C. Brasch, M. Brant-Zawadzki, and C. B. Higgins, in Scientific Program, Society of Magnetic Resonance in Medicine, Third Annual Meeting, New York, 1984, pp. 206–207.

[5] G. B. Matson, J. Magn. Reson. **56**, 200 (1984).

[6] W. T. Sobol, I. G. Cameron, J. C. MacTavish, L. J. Schreiner, and M. M. Pintor, in Scientific Program, Society of Magnetic Resonance in Medicine, Third Annual Meeting, New York, 1984, pp. 695–696.

[7] A. M. Aisen, G. M. Glazer, P. L. Carson, and D. O. Hearshen, Magn. Reson. Imag. **4**, 207 (1986).

[8] J. Granot, J. Magn. Reson. **53**, 386 (1983).

[9] E. O. Stejskal and J. E. Tanner, J. Chem. Phys. **42**, 288 (1964).

[10] J. Duijn, J. Creyghton, and J. Smidt, in Scientific Program, Society of Magnetic Resonance in Medicine, Third Annual Meeting, New York, 1984, pp. 197–198.

[11] E. L. Hahn, Phys. Rev. **80**, 580 (1950).

[12] E. L. Madsen and G. D. Fullerton, Magn. Reson. Imag. **1**, 135 (1982).

[13] V. M. Runge, G. Johnson, F. W. Smith, J. J. Erickson, R. R. Price, C. L. Partain, and A. E. James, Noninv. Med. Imag. **1**, 49 (1984).

[14] T. W. Redpath, Phys. Med. Biol. **27**, 1057 (1982).

[15] W. A. Edelstein, P. A. Bottomley, and L. M. Pfeifer, Med. Phys. **11**, 180 (1983).

[16] L. E. Crooks, M. Arakawa, J. Hoenninger, B. McCarten, J. Watts, and L. Kaufmann, Radiology **152**, 127 (1984).

[17] H. R. Hart, P. A. Bottomley, W. A. Edelstein, S. G. Karr, W. M. Leuer, O. Mueller, R. W. Reddington, J. S. Schenck, L. S. Smith, and D. Vatis, Am. J. Roentgenol. **141**, 1195 (1983).

[18] B. W. Kernighan and D. M. Ritchie, *The C Programming Language* (Prentice–Hall, Englewood Cliffs, 1978).

[19] P. F. Judy, Med. Phys. **3**, 233 (1976).

[20] S. M. Bentzen, Med. Phys. **10**, 579 (1983).

[21] F. L. Bookstein, J. Theor. Biol. **107**, 475 (1984).

[22] F. L. Bookstein, *The Measurement of Biological Shape and Shape Change* (Springer, New York, 1978).