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Original

MTF Measurement in MRI Using a Complex Subtraction Method

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Introduction

In evaluating a quality of images obtained by Magnetic Resonance Imaging(MR)system, one of important index is a resolution property. The resolution property is needed not only for a quality control, but also for developing pulse sequence and hardware or for performance evaluation in replacement of them and in the verification of simulation. Especially, nowadays the pulse sequences to perform k-space trajectory by various methods are widely used for shortening the imaging time. The assessment of the resolution property, therefore, is now very important.[1-11)]

Regarding the evaluation of the MRI resolution property, there are many reports and standards[12-33)], e.g. a method to just evaluate it visually[17)] and a report on the development of pulse sequence creating directly the point spread function(PSF).[27)] Visual evaluation produces only vague results and the pulse sequence directly creating PSF involves a difficulty in making with ordinary MRI systems. Under such circumstances, for assessing the resolution property of MRI, it is still desirable to measure a modulation transfer function(MTF).[17,20)]

Steckner et al raised various kinds of problem in measuring MTF of MRI[19,21,23)], namely 1)Loss of system linearity due to the absolute value calculation, 2) Modulation of low spatial frequency is generated by the effect of non-uniformity, 3)By the effect of artifact (chemical shift, T_2 relaxation, ringing, partial volume, etc.)the measured value is underestimated and so on. Above all the process of above 1)Absolute value calculation causes an important problem in evaluating the resolution property of MRI. Theoretically a real number component satisfies the linearity of MR signal, but in general images are output after calculating the absolute value with following equation.

$$M = \left(R^2 + I^2\right)^{\frac{1}{2}} \geq 0 \quad \cdots\cdots(1)$$

where; M=Absolute value
R=Real
I=Imaginary

By the above equation the negative value on a base line of line spread function(LSF)is inverted so that the linearity of signal value is not satisfied.

The absolute value operation is expressed by a phase function $p(x)$ ($|p(x)|=1$)which converts functions of all complex domain only into positive real domain. The absolute value $LSF_m(x)$of $LSF(x)$ in the complex domain is expressed by following equation(2).

$$LSF_m(x) = p(x) \cdot LSF(x) \quad \cdots\cdots(2)$$

MTF($MTF_m(f)$)after calculating the absolute value can be expressed by the equation(3).

$$MTF_m(f) = FT\{p(x)\} * MTF(f) \quad \cdots\cdots(3)$$

where; $*$=Convolution
FT=Fourier transform
$MTF(f)$=MTF before calculating the absolute value

This shows that an error occurs in MTF because of not constant phase response. $p(x)$causes serious distortion, since there is no bandwidth limitation in $FT\{p(x)\}$(Fig. 1). MTF of the modulus image obtained by the above said process does not express accurately the resolution property of MRI system itself, because a new spatial frequency characteristics is added by the absolute value operation which does not satisfy the linearity.

In order to avoid the process error in the absolute value reconstruction, there are several methods as shown below; a)a method in which a filter of triangle function is used before reconstruction, then the deconvolution is

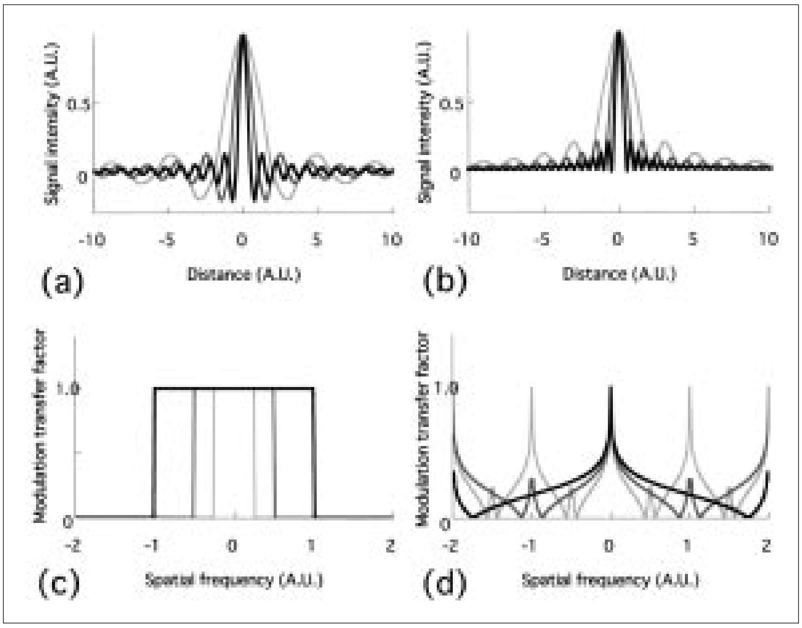

Fig. 1 Simulation of the erroneous MTF in MRI.
LSF profiles of (a) the ideal MRI signal [sinc (x)] and (b) its magnitude operator (|sinc (x)|) for altering the acquisition time of the raw data. The ratio was 1 (black line): 2 (dark gray line): 4 (light gray line). MTFs (c and b) were calculated from the respective LSF profiles (a and b). Note that (c), the MTF values without magnitude operator, fall dramatically at each of the theoretically predicted cutoff frequencies. (d) A non-linear magnitude operator causes distortions. A.U.: arbitrary units.

performed using $\mathrm{sinc}^2(x)$ after calculating MTF[19]), b) a method to raise an edge spread function (ESF) so as to satisfy the linearity by utilizing partial volume effect of signal generating part and no-signal part[23]), c) a method to measure MTF of positive and negative spatial frequencies (pre-display MTF or two-sided MTF) from data of complex numbers before calculating the absolute value.[23]) In the method a), an multiplication of large numbers is required for filter compensating process, which amplifies MTF noise level significantly and adds limitations to accuracy and reliability. Furthermore, in both a) and b) methods the phase information in image area and information on negative spatial frequency are all lost due to the above said phase function $p(x)$. In the method c) it is possible to measure MTF of MRI system most accurately, but its measurement and processing are difficult and involves troublesome works.[25])

To improve these problems, we developed a method (complex subtraction method) to measure the resolution properties of MRI system more simply and accurately. With this method, we could obtain the pre-display MTF similar with the result in Steckner's report. In addition, it becomes possible not only to save the compensation of image non-uniformity and discrete differentiation, but also to eliminate unnecessary artifact. In this report, we describe procedures of complex subtraction method and as well the results of the MTF measurements under various conditions in fast spin-echo (SE) sequences.

1. Methods

1-1 Phantom

As shown in Fig. 2, a phantom used for the complex subtraction method has a structure as mentioned below. In a hemispherical container (170mm dia.) made of polymethyl-methacrylate resin (plexi-glass), we installed a guide frame (50mm high) made of plexi-glass in which thin polyethylene terephthalate (PET) sheet having 70mm high could be put in and pull out. As pixel diameter used in present test is about 1mm, the thickness of PET sheet was set to 400μm. It is necessary, however, to compensate the thickness of PET sheet as mentioned later. If the thickness is too thin, SNR of slit image becomes very poor, on the contrary, if it is too thick, an error in deconvolution increases.

According to the purpose of measurement, we sealed a discretional solution. As the guide frame and PET sheet are fit via anti-leakage agent, the solution does not leak out even if the container is laid. In present study, the effect of T_2 in the fast SE to the resolution properties was also measured. Thus, phantoms for long T_2 value and short T_2 value were prepared. In the phantom for long T_2, copper sulfate solution ($CuSO_4 \cdot 5H_2O$: 0.77g/l) was sealed. The solution has characteristics; T_2 365ms and T_1 423ms under static magnetic field strength of 1.5T. In the short T_1 phantom, manganese chloride ($MnCl_2 \cdot 4H_2O$: 0.05mg/l) was sealed. It has characteristics, T_2 52ms and T_1 481ms under static magnetic field strength of 1.5T. Since data sampling time (4ms) was sufficiently short, the error due to T_2 relaxation in the frequency encoding direction was ignored.[34] Even with the phantom for short T_2 (T_2=52ms), an amplitude fluctuation[34] at high spatial frequency was 1.002.

1-2 Measurement and analysis of MTF

Since in some cases the resolution properties of MRI is not isotropy, it is necessary to measure the MTFs both in the frequency encoding direction and the phase encoding direction. At first, place the phantom in an RF coil and fix it tightly. At this time, a space for pulling out PET sheet by 3 cm at least was secured. As shown in Fig. 2a and b, the slice plane should be accurately perpendicular to the PET sheet at the position not crossing with the guide frame. Also, the line object must be set so that it passes through the same part of each pixel. Since the shape of phantom is hemispherical ringing artifact[23] from surrounding edge is attenuated by partial volume effect.

Perform the first imaging under a discretional condition that SNR of LSF is sufficiently high to calculate MTF and obtain the complex image (real and imaginary images) (complex image A). In normal MRI system, it displays absolute images after phase-detection and Fourier transform of the digital data. Since MTF measurement is applicable only to linear system, LSF obtained from the modulus image of non-linearity can not be theoretically used.[19,23] However, as its pre-step, there is not any problem in calculating MTF from LSF in the complex domain. Field of view (FOV) should be set larger than the diameter of phantom at imaging plane in order to avoid to generate the aliasing artifact. Fathermore, the echo-peak should be located at the center of raw data[23], otherwise linear phase term is included in it and eventually the shift invariant can not be satisfied. To begin with, the image uniformity of MRI is low.[20] Even with the head coil which gives relatively high uniformity, the slit image is distorted as shown in Fig. 3a.

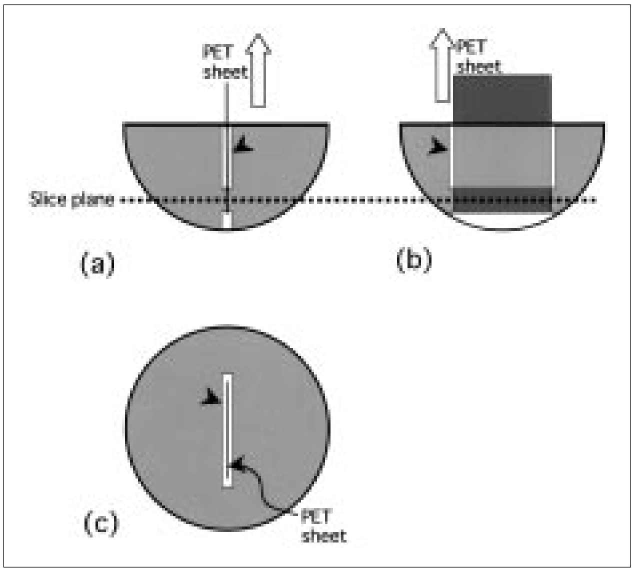

Fig. 2 Hemispherical phantom (170 mmϕ) used for the complex subtraction method.
(a and b) Two orthogonal side views and (c) top view. A thin PET sheet (400 μm) (a, b, and curved arrow in c) can be pulled out along the guiding frame (arrowheads in a, b, and c). The horizontal dotted lines in (a) and (b) show slice planes.

Therefore, immediately after the finish of first imaging, pull out PET sheet quietly about 2 cm while keeping the phantom unmoved. At this time, PET sheet and slice plane should not cross over. After passing 2 minutes in that situation, perform the imaging again under the same condition as that for complex image A (complex image B). As the PET sheet is thin and its moving distance is very short, the phantom configuration changes little. When the complex image A is subtracted from the complex image B, it is possible to obtain uniform slit image as shown in Fig. 3b and at the same time the artifact can be eliminated.

Sufficiently long complex LSFs (128 pixels in present study) obtained from the subtracted image is added more than 100 lines and they are averaged in order to improve SNR. If SNR is not high enough, the foot of LSF (low amplitude part of side-lobe) is buried in noises, resulting in the error in MTF.[21] SNR of LSF obtained in this study is almost equal to that of Steckner's report.[23] To this complex LSF (Fig. 3b), Fourier transform is applied, then the absolute value is calculated.

Finally, MTF ($MTF(f)$) in positive and negative spatial frequencies is calculated after compensating the thickness of PET sheet with following equation (4).

$$MTF(f)=\begin{cases} MTF'(f)\cdot \pi fd/\sin(\pi fd), & |f|>0 \\ 1 & , \quad f=0 \end{cases} \quad \cdots(4)$$

where; $MTF'(f)$=MTF before compensation

f=Spatial frequency

d=Thickness of PET sheet

In measuring MTF of MRI, it is not needed to take into consideration of the aliasing error, since a sampling theorem is met.[21,23] On the assumption that a relaxation time can be ignored, all signals $S(t)$ of 1-dimension detected in time t can be obtained by integrating spatially each spin signal.

$$S(t)=\int \rho(\mathbf{r})\exp\left(-i\mathbf{r}\,2\pi\gamma\int_0^t G(t')dt'\right)d\mathbf{r} \quad \cdots\cdots\cdots\cdots(5)$$

where; $\rho(\mathbf{r})$=Spin density at location $\mathbf{r}$,

γ=Gyromagnetic ratio

$G(t)$=Gradient strength

Let $k(t)$ be as follows.

$$k(t)=\gamma\int_0^t G(t')dt \quad \cdots\cdots\cdots\cdots(6)$$

Following equation (7) is induced.

$$S(t)=\int \rho(\mathbf{r})\exp\left(-i2\pi k(t)\mathbf{r}\right)d\mathbf{r} \quad \cdots\cdots\cdots\cdots(7)$$

Since above equation (7) can be deemed as Fourier transform of density function $\rho(\mathbf{r})$ in the location $\mathbf{r}$, it is found that $S(t)$ corresponds to the spatial frequency domain.[11,35] It is possible to consider that MR images are produced by inverted Fourier transform of spatial frequency data in k-space. Therefore, at the time when data sampling in the time axis was finished it was also sharply cut off against the spatial frequency. Inevitably frequency component more than Nyquist frequency does not exist at this stage. Therefore, pre-display MTF correspondents to digital MTF in which the aliasing is not generated and shows the spatial frequency characteristics in pre-stage of the absolute value process and image display.

1-3 Experimental condition

For the test we used Gyroscan ACS II of 1.5T static magnetic field made by Philips. In all the tests, the head coil was used and the imaging was performed by adding a loading device to the above phantom. MTF was measured in a conventional SE (C-SE) and in a fast SE (Turbo-SE) in which an effective echo time corresponds to the first echo.

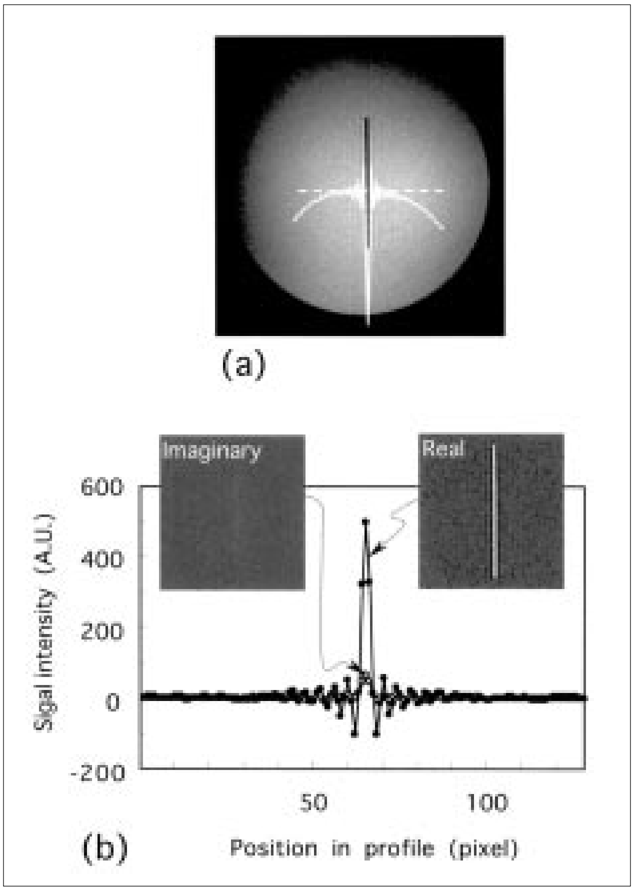

Fig. 3 Phantom images, profile of the slit, and LSF in C-SE.
(a) Slit image (real part) and its profile of the first scan.
(b) Subtracted complex images and the complex LSFs. Open and closed circles in the graph are real (right image in b) and imaginary (left image in b) parts, respectively. A.U.: arbitrary units.

k-space arrangement in the phase encoding direction of the Turbo-SE is【0, $+S$, $-S$, $+2S$, $-2S$, $3S$, $-3S$, ……$nS/2$, $-nS/2$】and the arrangement is set as one segment. The measurement was repeated from $(-S+1)/2$ to $(S-1)/2$ until data acquisition was finished. [S=Segment number, n=Echo Train Length (ETL)].[9,10] By altering ETL in the order of 1 (=C-SE), 3, 7, 15, and T_2 in the order of 52ms and 365ms, MTFs to the frequency encoding and phase encoding directions was measured. In case ETL was 1, 3, 7 and 15, the phase encoding line numbers were set as 256 (256 segments), 255 (85 segments) 252 (36 segments) and 255 (17 segments) respectively. The frequency encoding numbers were set to 256, and in order to identify the location of cut-off frequency 0 filling was done and raw data were set to 512×512. However, the window function to the raw data was not given, since MTF was modulated. Every imaging was conducted under below mentioned conditions;

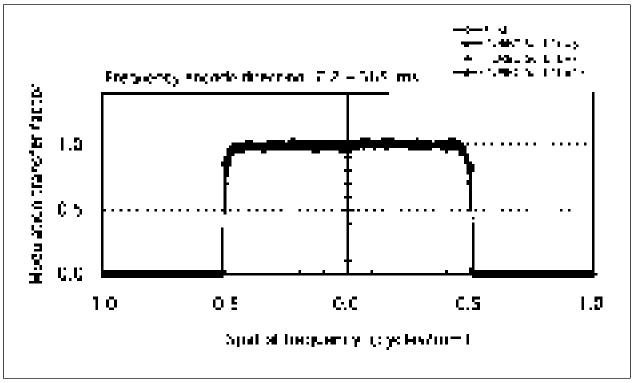

Fig. 4 MTFs of C-SE and Turbo SE (ETL of 3, 7, and 15) at long T_2 phantom (365 ms) in the frequency-encoding direction using the complex subtraction method.

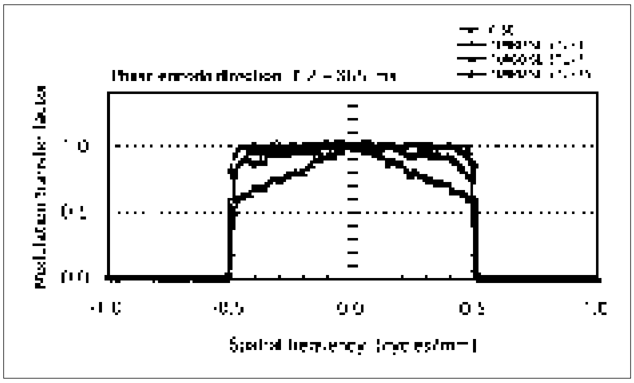

Fig. 5 MTFs of C-SE and Turbo SE (ETL of 3, 7, and 15) at long T_2 phantom (365 ms) in the phase-encoding direction using the complex subtraction method.

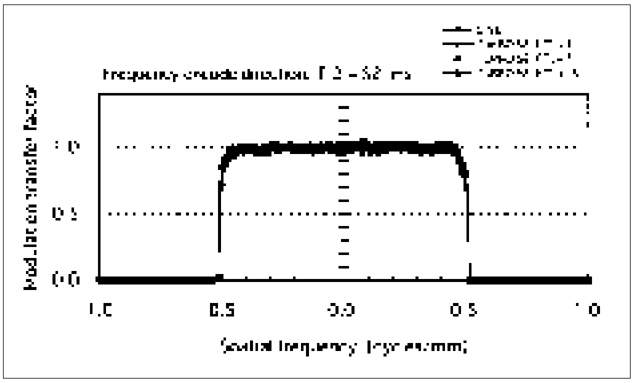

Fig. 6 MTFs of C-SE and Turbo SE (ETL of 3, 7, and 15) at short T_2 phantom (52 ms) in the frequency-encoding direction using the complex subtraction method.

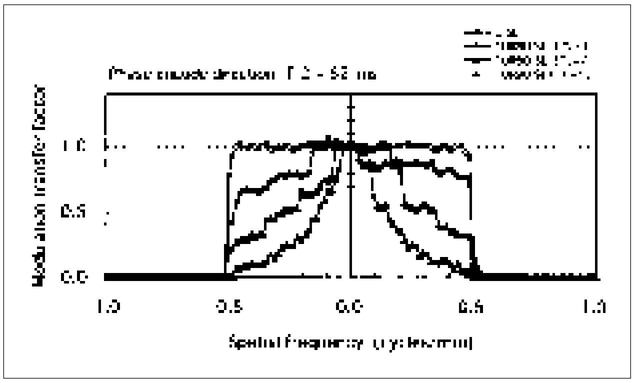

Fig. 7 MTFs of C-SE and Turbo SE (ETL of 3, 7, and 15) at short T_2 phantom (52 ms) in the phase-encoding direction using the complex subtraction method.

Repetition time: 1500ms, Effective echo time: 11ms, Echo interval: 11ms, Slice thickness: 3mm, FOV: 256×256mm. Reconstruction matrix: 512×512, number of signal averaged: 4, Band width: 250Hz/pixel. To avoid the artifact due to interference by multiple echo signal (spin echo and inhomogeneity echo) a shimming (linear gradient component) process was applied[3], after that the first imaging was performed.

2. Results

MTF measurement results obtained by the complex subtraction method are shown in Figs. 4〜7. MTFs in C-SE were well correspond to that of Steckner's report[23] (Fig. 4〜7 —○—). MTFs of C-SE were sharply dropped by theoretically obtainable cut-off frequency (0.5 cycles/mm) both in frequency encoding direction and phase encoding direction so that the shape of MTF is correspond very well to positive and negative spatial frequencies.

Similar with C-SE, MTFs of Turbo-SE were sharply fallen like a rectangular pulse wave by the cut-off frequency (Figs. 4 and 6). For Turbo-SE in the phase encoding direction, MTFs fell according to increasing ETL. Falling ratio of MTFs in the phase encoding direction were so large as T_2 becomes shorter (Figs. 5 and 7). Also, MTFs of Turbo-SE in the phase encoding direction were not symmetrical in positive and negative frequencies.

3. Discussion

In MRI which uses 2-dimensional (or 3-dimensional) Fourier transformation raw data themselves correspond to spatial frequency.[11,35] Therefore, it is not unable to analyze MTF directly from raw data. It is, however, very difficult to obtain SNR sufficient for calculating MTF so that for ordinary MRI systems it is difficult to acquire raw data. Therefore, at first we tried to average multiple LSFs obtained in image area in order to raise SNR, which made the calculation of MTF simpler and accurate. However, even if LSF or PSF was simply sought and also as much as we tried to input an ideal delta function, MR signal was lowered, which made it impossible to obtain sufficient SNR. Thus, in the complex subtraction method, as stated in [1. Methodology], the thickness of PET sheet was set to be within 1 pixel, namely, in the range of 2/5～3/5 pixel which could satisfy SNR for LSF, then the thickness PET sheet convoluted by LSF was compensated after Fourier transform. In actual, in C-SE we could obtain a rectangular MTF[21,23] which represents nearly constant value (within coefficient variation of 2%) up to the cut-off frequency. This proves that SNR is satisfied under conditions of this study. Also, by utilizing the complex subtraction method, it was possible to get LSF directly without calculating LSF from ESF and as well without affection by image non-uniformity.[20] Furthermore, ringing artifact from unwanted phantom edge which was a problem in measuring MTF[23] could be eliminated due to the effect of partial volume and subtraction process.

In actual, in C-SE and Turbo-SE obtained by the use of complex subtraction method we were able to obtain characteristics different respectively in encoding direction, ETL and T_2 value of phantom (Figs. 4～7). In all the measured results of MTFs, they did not have the frequency component more than Nyquist frequency of sampling system, since, as stated in [1-2 Measurement and analysis], the resolution properties as it is are dependent to limited sampling. Also, the reason why MTF of Turbo-SE in the phase encoding direction did not lower continually can be explained that the trajectory (ref. to [1-3 Test condition]) of k-space itself was not continuous.[9] Namely, in present experimental condition, positive and negative phase encoding lines in k-space were buried outward alternately. So, signals lowered stepwise according to T_2 relaxation while showing the same amplitude in every segment. Similarly, the reason why MTFs of Turbo-SE were not symmetrical in positive and negative spatial frequencies was due to the k-space trajectory. From the above, it is possible to consider that MTF in positive and negative spatial frequencies measured by the complex subtraction method shows faithfully an way to perform the trajectory in k-space. This information can not be obtained by MTF of the absolute value image. Of course, if a method for trajectory of k-space in a pulse sequence is well known,

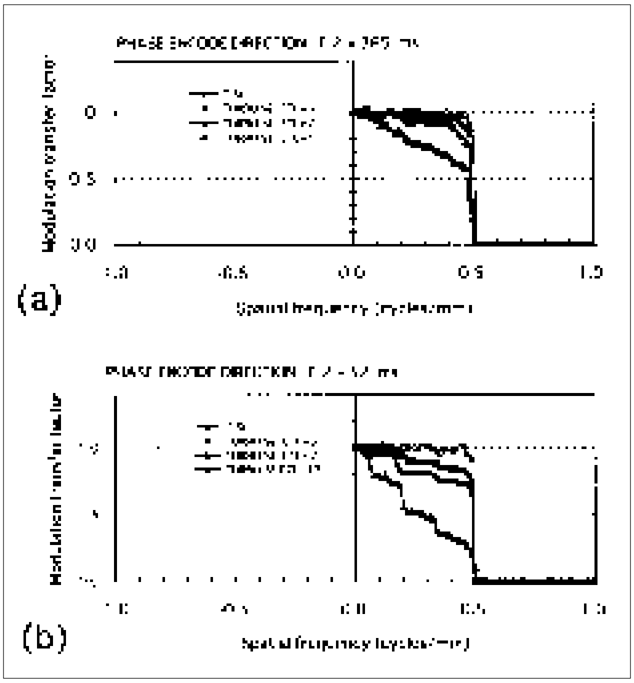

Fig. 8 MTFs of C-SE and turbo SE (ETL of 3, 7, and 15) at (a) long T_2 (365 ms) and (b) short T_2 (52 ms) phantoms in the phase-encoding direction using the magnitude subtraction method.

it is possible to simulate the resolution properties. In confirming actually, however, the detail of simulation and also in case the detail of sequence is not made public, it is necessary to measure MTF. For this reason, the evaluation of MTF measurement by the complex subtraction method we made herein is useful.

However, if any system fluctuation happens due to some unstable factors, it is impossible to perform a perfect subtraction among complex images obtained continuously, consequently MTF values of positive and negative spatial frequencies are distorted and become vague. In this case, as a alternate method, perform the subtracting process among absolute images in place of complex images. After that compensate a base line by polynomial fitting process, then it is possible to calculate MTF following the same procedure for the complex subtracting method (absolute subtraction method) (Fig. 8). With this method the problem of linearity as stated in [Introduction] does not arise, since the signal level of original line object is always higher than the background level. However, the absolute value subtracting method does not always represent information on all resolution properties of MRI. It is because that the phase information in image area and the negative spatial frequency information are all lost by the absolute processing[23] (Fig.

8). In actual, as seen in our experiments, unsymmetrical echo, half-Fourier method[23] and the like, there are cases that MTF of positive spatial frequency and negative MTF of negative spatial frequency are different. At this moment, there is no appropriate criteria to judge MTF from a non-linear system so that it is difficult to specify the condition applicable to the absolute subtraction method. We consider it is necessary to investigate hereafter the problem from multiple points of view.

4. Conclusion

We have developed a method to evaluate MTF of positive and negative frequencies from subtracted complex images. We evaluated actually MTFs of C-SE and Turbo-SE in which an effective echo is the first echo while altering T_2 values of phantom in the frequency encoding direction and phase encoding direction. MTF represents a feature of trajectory of k-space in each sequence. It becomes possible to eliminate the effect of artifact and image non-uniformity by the complex subtraction method.

The measurement of MTF by the complex subtraction method enables us to analyze more in detail the resolution properties of MRI system.

In ending the report we wish to express our sincere thanks to Mr. Steckner, Lawson Research Lab.(St. Joseph's Health Center of London)and related personnel in Nagoya City Hospital for proper advice in pursuing the subject.

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