Original

System Stability of a 3T-MRI During Continuous EPI Scan

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Introduction

The high field magnetic resonance imaging (MRI) system has become widespread in the clinical field due to its intrinsically higher signal-to-noise ratio (SNR), driven by recent improvements in the RF pulse^{1, 2)} and receiver coil system³⁻⁷⁾ (e.g. multi-channel phased array coil). In addition, it has also been intensively used in the non-invasive brain functional imaging community. Thus, it is important to know the general characteristics of high field MRI.

At the ATR-Brain Activity Imaging Center (ATR-BAIC), functional-MRI (f-MRI) experiments using an echo planar imaging (EPI) sequence have been routinely performed, and we are especially concerned about the stability of EPI. The EPI sequence has been widely used for f-MRI. Although this sequence can detect a subtle magnetic field change resulting from activations of the local brain, it is easily affected by physiological alterations⁸⁻¹⁰⁾ such as respiration, pulsation, and subject motion

in addition to system instability due to a main magnetic field (B₀) change¹¹⁻¹⁵⁾ throughout data acquisition.

The experimental paradigms of f-MRI are diverse. In many cases, the EPI sequence of several minutes is repeated several times, but long duration (more than 30 min) scanning is sometimes required.¹⁶⁾ In such a situation, the signal stability over time series is inevitably important and, if not maintained, the results of the statistical analysis might degenerate because of large noise components.

In our previous study,^{17, 18)} we reported large signal fluctuations in EPI time series due to B₀ instability of the 1.5T MRI at ATR-BAIC and proposed some solutions, including system warm-up and/or image shift corrections (along the phase direction). The implementation of these procedures succeeded in preventing the under- and over-estimation of brain activation. Since the characteristics of an MRI system highly depend on B₀ strength and even the manufacturer,¹⁹⁻²³⁾ we have to

Summary

The purpose of this study is to evaluate the general stability and image properties of a 3T MRI system newly installed at the ATR-Brain Activity Imaging Center (ATR-BAIC), in addition to a conventional 1.5T system. In this study, we focused on the echo planar imaging (EPI) sequence since continuous EPI with a relatively long duration of up to 30 min is routinely used, and the stabilization of EPI is always a concern. The following five results were obtained: (1) Significant image shifts along the phase direction were observed in the 1.5T data but not in the 3T data, although B₀ shifts in both the 1.5T and 3T systems were the same level (1.3 Hz/min); (2) The signal fluctuations were 1/2-1/3 smaller in the 3T system compared with the 1.5T system; (3) The temporal signal-to-noise ratio (TSNR) of the 3T system was 1.7-2.0 (CP-coil) and 2.5-4.0 (12ch-coil) greater than the 1.5T system; (4) We found a low frequency periodic fluctuation (cycles of approximately 30-40 sec), and an increase in noise in the latter half of the long term series, which might originate from the 3T MRI scanner; and (5) Spatial non-uniformity of TSNR and voxels with a linear-trend were observed in the 3T data.

Key words: 3T-MRI, EPI, spatio-temporal stability

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Table 1 Imaging parameters

	1.5T	3T
Phantom		
TR	5000 msec	3000 msec
TE	49 msec	30 msec
FA	90°	80°
FOV	192 mm	192 mm
Matrix	64×64	64×64
Slice thickness	4 mm	4 mm
Number of slice/volume	50 (no gap, interleave)	50 (no gap, interleave)
BW (phase direction)	23.15 Hz/pixel	37 Hz/pixel
BW (frequency direction)	1482 Hz/pixel	2368 Hz/pixel
Volunteer		
TR	1000 msec	1000 msec
TE	45 msec	30 msec
FA	60°	60°
FOV	192 mm	192 mm
Matrix	64×64	64×64
Slice thickness	3 mm	5 mm
Number of slice/volume	10 (no gap, interleave)	16 (no gap, interleave)
BW (phase direction)	23.15 Hz/pixel	37 Hz/pixel
BW (frequency direction)	1482 Hz/pixel	2368 Hz/pixel

validate the system stability of a newly installed 3T-MRI.

To achieve high-quality data acquisition, we evaluate our 3T-MRI system in terms of the spatio-temporal characteristics of the image time series, including temporal signal change, periodic fluctuation, temporal SNR (TSNR) and trend, during a long duration (30 min) EPI scan.

1. Method

Comparative studies were performed at different field strengths on a 1.5T whole body MRI system (Gradient coils strength of 27 mT/m and a maximum slew rate of 72 T/m/s), the Magnex ECLIPSE (Shimadzu-Marconi, Kyoto, Japan) and a 3T whole body MRI system (Gradient coils strength of 45 mT/m and a maximum slew rate of 200 T/m/s), the Magnetom Trio a Tim (Siemens, Erlangen, Germany). A spherical phantom (17 cm in diameter and doped with 1.25 g $\text{NiSO}_4 \times 6\text{H}_2\text{O}$ per 1000 g H_2O) provided by the MRI vendor and three healthy volunteers (one volunteer for the 1.5T and two volunteers for the 3T) were scanned in a long-duration (up to 30 min) 2D EPI acquisition. The room temperature had been set to 23°C all day. The

phantom was placed in the scanner room overnight or longer before the experiment. Informed consent was obtained from every volunteer before participation in this study.

A receive-only quadrature detection (QD) head coil for 1.5T and two types of head coils including a transmit/receive cross polarization (CP) head coil and a receive-only 12ch head matrix coil for 3T were used. The 12ch head matrix coil supports three types of imaging modes (4ch CP-mode, 8ch dual-mode and 12ch triple-mode), and we used triple-mode in this study because of higher image homogeneity and SNR in the preparatory experiment.

Since the purpose of this study is to evaluate under actual f-MRI conditions, selected imaging parameters were set to nearly maximum system load conditions. Consequently, imaging parameters were not identical between the systems depending on system limitations. Imaging parameters are summarized in Table 1.

1-1 B0 change and image shift

To investigate the effects of the change in the magnetic field with increased scanning time, we measured the central frequency and image shift. From

April 25 to May 22, 2002, we measured the center frequency before and after system warm-up for more than 20 min on the 1.5T system and calculated the centroid position (i.e. image shift) of the phantom image obtained by EPI (BW 23.15 Hz/pixel). From August 27 to October 24, 2007, we also measured the center frequency before and after a 30 min EPI with BW 37Hz/pixel on the 3T system three times in the phantom and once in the volunteers, and image shifts were estimated using realign functions in the SPM99 software (Wellcome Department of Cognitive Neurology, London, UK) for phantom data only. The data from volunteers (dash mark “-” at 3T data 2 on Table 2) were excluded from this analysis to avoid the confounding effects of body motion.

1-2 Signal fluctuation

To investigate signal fluctuations over time series, percent signal change (PSC_i in eq. 1) was evaluated. We first calculated the mean signal intensity of each slice for all volumes. To eliminate the spatially varying fluctuations due to image shifts, we used the mean value of all voxels (pixels) within a slice instead of ROI placed in the specific area. The sum of the total pixel values through the plane were divided by the number of pixels: $64 \times 64 = 4096$ pixels/slice, giving $S_{mean,i}$ in eq. 1, where the suffix mean represents the slice-based mean and the number of volumes is indexed by i . The time series of a particular slice was then normalized by the first observation ($S_{mean,1}$ in eq. 1). The resulting time series, hence, indicated the percentage of signal fluctuation relative to the start of scanning.

$$PSC_i = (S_{mean,i} / S_{mean,1}) \times 100 \quad \dots\dots\dots (1)$$

1-3 Periodic fluctuation

In order to evaluate the periodic fluctuation, we estimated the auto correlation function (ACF) and cross correlation function (CCF) at the middle slice of PSC (c.f. eq. 1). These functions are suitable for finding the periodical structures in a noisy time series. Before ACF and CCF were calculated, the low frequency trend was removed by 5th order orthogonal polynomial and centered by the grand mean of time series (R).

The ACF was defined as:

$$ACF(j) = \frac{1}{N} \sum_{i=1}^N R(i) \cdot R(i+j) \quad j = 0, 1, 2, \dots, N \quad \dots (2)$$

where j is lag and N is the total number of acquisitions. Similarly, the CCF was defined as:

$$CCF(j) = \frac{1}{N} \sum_{i=1}^N Rb(i) \cdot Rc(i+j) \quad j = 0, 1, 2, \dots, N \quad \dots (3)$$

where Rb and Rc are the de-trended time series from volunteers B and C.

1-4 Temporal Signal-to-noise ratio (TSNR)

It is obvious that the SNR is greater in the 3T MRI system than in the 1.5T, since SNR is theoretically proportional to magnetic field strength.²⁴⁾ However, in the temporal aspect, it actually decreases by system fluctuation, physiological noise,²¹⁾ and apparent image shift due to the B0 shift.^{12, 14)} Thus, we used the TSNR²⁵⁾ to estimate the pixel signal fluctuation. The TSNR of each pixel was calculated every 10 min. TSNR was defined as:

$$TSNR_{(x,y)} = \frac{\mu_{(x,y)}}{\sigma_{(x,y)}} = \frac{\mu_{(x,y)}}{\sqrt{\frac{1}{N} \sum_{i=1}^N (S_{(x,y,i)} - \mu_{(x,y)})^2}} \quad \dots\dots\dots (4)$$

where $\mu_{(x,y)}$ is temporal mean of the pixel time series at coordinates (x, y) , $\sigma_{(x,y)}$ is the standard deviation of this and $S_{(x,y,i)}$ is a pixel value at coordinates (x, y) on i th acquisition.

1-5 Trend

If the signal fluctuation shows the same tendencies over all pixels through the plane, adjustments may be quite simple. To evaluate the low frequency trend at each pixel, we calculated the correlation between the pixel time series and monotonically increasing function as $Y(i) = x_i$ ($x_1=1, x_2=2, \dots, x_n=N$).

The TC was defined as:

$$TC_{(x,y)} = \frac{\sum_{i=1}^N S_{(x,y,i)} Y(i) - N \mu_{(x,y)} \bar{Y}}{\sqrt{\left[\sum_{i=1}^N (S_{(x,y,i)} - \mu_{(x,y)})^2 \right] \left[\sum_{i=1}^N (Y(i) - \bar{Y})^2 \right]}} \quad \dots (5)$$

where $S_{(x,y,i)}$ is a pixel value at coordinates (x, y) on i th acquisition, $\mu_{(x,y)}$ is the temporal mean of a pixel at coordinates (x, y) , and $\bar{Y}$ is a mean value of monotonically increasing function. Note that TC was calculated every 10 min.

Table 2 B0 fluctuation and image shift to the phase direction (Y-axis) during system load

1.5T				
data #	Δf (Hz)	ppm	Image shift (pixel)	Scan duration (min)
1	+41	+0.64	1.60	25
2	+35	+0.55	1.30	20
3	+18	+0.28	0.65	20
4	+25	+0.39	0.96	21
5	+39	+0.61	1.39	23
6	+27	+0.42	1.00	22
7	+33	+0.52	1.31	23
8	+14	+0.22	0.50	23
9	+12	+0.19	0.48	24
10	+20	+0.31	0.73	25
11	+23	+0.36	0.83	25
Average	+26	+0.41	0.98	23

3T				
data #	Δf (Hz)	ppm	Image shift (pixel)	Scan duration (min)
1	+40	+0.32	0.023	30
2	+36	+0.29	—	30
3	+41	+0.33	0.014	30
4	+43	+0.35	0.002	30
Average	+40	+0.32	0.013	30

2. Result

2-1 B0 change and image shift

The B0 change during system load and image shift along the phase direction are summarized in Table 2. The increase in central frequency is about 1 Hz/min, and the image shift was about 1 pixel, which was roughly in accordance with the theory (23.15 Hz/pixel). Likewise, the central frequency of the 3T system increased approximately 40 Hz/30 min (i.e. 1.3 Hz/min). This is equivalent to more than 1 pixel displacement along the phase direction, however, the actual image shift we observed was 0.013 pixel, which was less than 1/80 of the theoretical value compared to the theoretical image shift.

Fig. 1 shows the realignment parameters (a, b: phantom; c, d: volunteer; a, c: CP-coil; and b, d: 12ch-coil) in the 3T data. The letters X, Y, Z in this figure denote the direction of frequency, phase, and slice, respectively. Note that both image shift patterns of a vs. b and c vs. d were similar and therefore independent of the head coil type.

In the phantom's data (Fig. 1a, b), we did not observe a significant image shift in the Y-direction, which should be obvious in theory. Instead, we

observed large image shifts along the X-direction (CP-coil: 0.07 mm=0.023 pixels; 12ch-coil: 0.1 mm=0.033 pixels) and Z-direction (CP-coil: 0.14 mm; 12ch-coil: 0.18 mm).

2-2 PSC

Fig. 2 shows the time course of PSC of all slices, and the typical time course from the middle slice is also shown in the small graph in Fig. 2. Signal fluctuations in the 3T system were relatively stable, and it was actually 1/2-1/3 of the minimum to maximum width of the 1.5T system. The ranges of these fluctuations were $\pm 0.5\%$ in the phantom and $\pm 2\%$ in the volunteer. Moreover, we observed an increased variance in the data from the volunteers in the 3T system as the number of scans grew. When we compared the noise variance between the first and later half of the de-trended time series (e.g. the time course of the middle slice as shown in the small graph in Fig. 2), it was found that noise variance increased by three times or more in the later part.

2-3 ACF and CCF

Fig. 3 shows the ACF and CCF, where the horizon-

Fig. 1 The realignment parameters (a, b: phantom; c, d: volunteer, a, c: CP-coil; and b, d: 12ch-coil) in the 3T data. The letters X, Y, Z in the figure denote the direction of frequency, phase, and slice, respectively.

a	b
c	d

Fig. 2 Time course of PSC of all slices. Typical time series of the PSC was shown in the small graph. The upper row (a, b, c) shows phantom data and the bottom row (d, e, f) shows volunteer data. Left column (a, d), middle column (b, e), and right column (c, f) show 1.5T, 3T using CP-coil and 3T using 12ch-coil, respectively.

a	b	c
d	e	f

tal axis represents the lag (or sec, in this case) and the vertical axis represents the ACF and/or CCF. Periodical fluctuation of the cycle of 30-40 sec, which is indicated by small triangles in the graph, was observed. This periodical fluctuation was also observed in the other volunteer's data (Fig. 3c, gray line), and more importantly, the period of fluctuation was similar between different volunteers (B and C) as shown in CCF (Fig. 3c, black line).

2-4 TSNR

Fig. 4 shows TSNR images of every 10 min in sequence. The pixel TSNRs in the 3T system (Fig. 4b, c, e, f) were approximately 1.7-2.0 times greater at the CP-coil (i.e. Fig. 4 a vs. b and d vs. e) and approximately 2.5-4.0 times greater at the 12ch-coil (i.e. Fig. 4 a vs. c and d vs. f) than that of the 1.5T system (Fig. 4a, b). A torus-shaped higher SNR region was observed in the 12ch-coil when we used the phantom (Fig. 4c; see first 10 min and last 10 min of TSNR images).

2-5 TC

Fig. 5 shows the signal trend of pixel time series of every 10 min in sequence. The lower graph shows the PSC of the center slice as presented in Fig. 2. Distinctive spatial non-uniformities of signal trend were observed on the 3T data (Fig. 5b, c, e, f).

3. Discussion

The B0 fluctuation of MRI during the system load has often been reported,¹¹⁻¹⁵⁾ and this causes image shifts, where the degree of shift and direction depend on the imaging sequence. In the EPI sequence, the B0 change can affect the image position in both phase and frequency directions simultaneously. In this study, the image shift along the phase direction (Y-axis) is expected to be 64 times greater than the frequency direction (X-axis) in theory.²⁶⁾ However, the image shift along the phase direction was quite small, but that along the frequency direction remained in the theoretical level as shown in Fig. 1a, b (i.e. phantom data). Therefore, we suppose that the phase shift correction during the image reconstruction may be applied and actually functioned well.

Fig. 2 suggested that the PSC and the TSNR of the 3T system were stable compared to that of the 1.5T. In the volunteer's PSC by the 3T system, however, the variance progressively increased in the latter half of the

Fig. 3 The top (a) and middle (b) figures show the ACF by phantom and volunteer, respectively. Bottom (c) shows the ACF by volunteers B, C and the CCF between volunteers B and C.

The horizontal axis represents the lag (or sec) and the vertical axis represents the ACF and/or CCF. Triangles in Fig. 3c indicate the peak of periodical fluctuation.

time course (see small graph in Fig. 2e, f). This change was specific to the volunteer's data by the 3T. According to the references,²⁷⁻³⁰⁾ body temperature increased by the heating effect of radio-frequency (RF) pulse²⁷⁻²⁹⁾ and become higher over time.²⁹⁾ Moreover, the relationship between cerebral blood flow (CBF) and artificial

Fig. 4 TSNR images of every 10 min in sequence. The top row (a, d) is 1.5T data. The middle (b, e) and bottom rows (c, f) are 3T data by CP-coil and 12ch-coil, respectively.

a	d
b	e
c	f

Fig. 5 The linear trends of signal fluctuations (i.e. TC) of every 10 min in sequence. The lower graph shows the PSC time course of the middle slice as presented in Fig. 2. The middle (b, e) and bottom rows (c, f) are 3T data by CP-coil and 12ch-coil, respectively. The TC ranges from -1.0(black) to 1.0(white). The value represents the degrees of correlation with the monotonically increasing function.

a	d
b	e
c	f

elevation of brain temperature are linear (10% per 1°C).³⁰⁾ Hence, the variance fluctuation we observed might be attributed to heat accumulation from continuous exposure to RF pulse and/or increased CBF, and we would emphasize careful attention to volunteers during prolonged scans or experiments with the 3T.

We observed the periodical fluctuation with a cycle of about 30-40 sec in ACF (Fig. 3) at the 3T. This may cause a serious problem in the f-MRI experiment because the periodic fluctuation we observed could synchronize with the experimental paradigm, which has a cycle of 40-60 sec as usual. Since it was commonly observed among different volunteers as shown in the CCF and, furthermore, the period was too long to believe that this phenomenon was of physiological origin, we speculate that it might be attributed to some sort of system fluctuation with our 3T system. However, we unexpectedly could not observe such a fluctuation in the phantom and suppose that this may result from the phantom characteristics, such as high fluidity of the liquid phantom. Further study will need to clarify the origin of this phenomenon, and the design of the new phantom whose fluidity can simulate biological tissues is required.

We also observed spatial non-uniformity in the TSNR (Fig. 4) and TC (Fig. 5) of the phantom data at 3T. This is also a serious problem because this may cause positional bias in the f-MRI data analysis. Therefore, we have to conduct further investigations to confirm whether this phenomenon is specific to the phantom or not by using a special liquid phantom with controlled fluidity and persistence.

Finally, we would emphasize that our observations

in this study were not the general features of the 3T MRI system. It is quite natural that image qualities and characteristics are different across B0 strength, system set-ups^{19, 20)} and even MRI manufacturers. Therefore, it is important for each MRI center to know its own system characteristics and incorporate them, for example, in creating experimental design and conducting analysis. In particular, it is important to carefully evaluate the image time series in term of temporal stabilities after system installation. We believe that the measurement method we used in this study, for example, image shift and signal fluctuation are useful. Besides work at the center level, we report that multi-center and/or multi-scanner comparative studies^{19, 20, 31)} have already begun in the United States to standardize the MRI system. Such an attempt will also be needed in Japan.

4. Conclusion

We evaluated the spatio-temporal characteristics of MRI systems (1.5T and 3T) installed at ATR-BAIC by using relatively long duration EPI (up to 30 min). As a result, we confirmed that the PSC and the TSNR in the 3T were excellent compared to the 1.5T and that the 3T system was applicable to similar f-MRI experiments that have been performed with the 1.5T. On the other hand, we found some problems, including low frequency periodic fluctuations in the volunteers but not in the phantom, time-dependent increase of noise variance, positional dependence of signal trends, and TSNR. These results should be incorporated in experimental design and analysis.

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図表の説明

- Fig. 1 3Tにおけるrealignment parameter (a, bはphantom, c, dは被験者, a, cはCP-coil, b, dは12ch-coilである). 図中のX, Y, Zはそれぞれ周波数, 位相, スライス方向を示す.
- Fig. 2 各スライス平均信号のPSCを示す. 代表的なPSCを小グラフに示した. 上段の三つ(a, b, c)はphantomデータ, 下段の三つ(d, e, f)は被験者データである. 左列(a, d)は1.5Tのデータ, 中央列(b, e)はCP-coilを用いた 3Tデータ, 右列(c, f)は12ch-coilを用いた 3Tデータである.
- Fig. 3 上段(a)と中央(b)はそれぞれphantomと被験者のACFである. 下段(c)には被験者B, Cの各ACFおよびこの両被験者間データにおけるCCFを示した. 横軸はラグ(sec), 縦軸はACF, CCFを示す. 周期性変動のピークを図中に△印で示した.
- Fig. 4 1.5T (a, b)と 3T (b, c, e, f)データの各10分ごとのTSNR画像を順に示す.
- Fig. 5 1.5T (a, b)と 3T (b, c, e, f)データの各10分ごとのTC画像を順に示す. 図下グラフはFig. 2でも示した代表的なスライス信号変動のグラフである. TCの表示幅は-1.0(黒)から1.0(白)となる. マイナスは単純増加関数に対して逆方向に信号が変動していることを示し, プラスは順方向を示す.

Table 1 撮像パラメータ

Table 2 システム稼働中のB0変動量と位相方向への画像シフト