

## Optimization of scanning parameters for MR elastography at 3.0 T clinical unit: volunteer study

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Received: 15 January 2014 / Accepted: 10 April 2014 / Published online: 28 April 2014  
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### Abstract

**Purpose** We sought to optimize scanning parameters for MR elastography at 3.0 T clinical unit.

**Materials and methods** 10 volunteers were scanned with various magnetization encoding gradient (MEG) frequencies from 60 to 120 Hz at every 10 Hz, with otherwise fixed parameters (external driver frequency/amplitude = 60 Hz/50 %, 10 mm slice thickness, etc.). Images were qualitatively assessed for the degree of image defects,

and also quantitatively for the areas without cross-hatching. After determining optimal MEG frequency, external driver amplitudes of 70 % (vs 50 %) and slice thickness of 8 mm (vs 10 mm) were also tested. With the optimized parameters, scans were repeated 1 week after the initial scan, and the repeatability of the liver stiffness measurement was validated.

**Results** 80 or 90 Hz was shown to be the best MEG frequency. There were no significant differences in the qualitative and quantitative assessment between the two amplitudes and two slice thicknesses; however, 70 % amplitude resulted in discomfort at the chest wall beneath the external acoustic driver. Thus, MEG 80 (or 90) Hz, amplitude 50 %, and thickness 10 (or 8) mm were considered optimal. Repeatability of the liver stiffness measurement was  $\pm 10$  % (95 % confidence interval).

**Conclusions** With the optimized parameters, repeatability of  $\pm 10$  % in liver stiffness measurement was obtained.

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**Keywords** MR elastography · 3.0 T · Parameter optimization · Magnetization encoding gradient · Repeatability

### Introduction

The usefulness of MR elastography (MRE), with its high accuracy and repeatability, in assessing the degree of liver fibrosis has been reported [1–13]. Most of these data were obtained at 1.5 T units and little clinical data obtained at 3.0 T unit have been published, to the best of our knowledge. MRE at 3.0 T clinical unit was installed in our institute in 2012, and we started its clinical application; however, we found that MRE images were substantially degraded for clinical use when default scanning parameters

were used. We therefore conducted this volunteer study to elucidate the optimal scanning parameters for a 3.0 T clinical unit, with particular interest in motion encoding gradient (MEG) frequency, and confirmed its usefulness by validating the repeatability of the liver stiffness measurement with the optimized parameters.

## Materials and methods

This study was approved by our institutional review board, and informed consent was obtained from all volunteers.

### Volunteers

Ten volunteers (nine men and one woman) with no known significant liver disease were recruited. Their ages ranged from 27 to 63 years (mean 36.4); body weights ranged from 48 to 100 kg (mean 63.8), and body mass indices ranged from 17.8 to 31.9 (mean 22.4).

### MR techniques

The MR equipment used was a 3.0 T clinical unit (Discovery 750 W, GE, Milwaukee, WI, USA) along with a 32-element phased-array coil. At least MRE and three-dimensional (3D) T1-weighted images (T1WI) were obtained for all of the volunteers. 3D T1WI was obtained with the two-point Dixon method, with repetition time (TR)/echo time (TE)/flip angle = 5.4 ms/1.2 and 2.3 ms/15 degrees, 224 × 224 matrix, 68 slices/slab (slice thickness = 3 mm), and one excitation.

For MRE, a 19-cm-diameter passive pneumatic driver was positioned over the right rib cage at the level of the xiphoid process and attached to an acoustic waveform generator. A 60 Hz waveform was applied to the driver. A 2D spin-echo echo-planar MRE sequence acquired magnitude and unwrapped phase difference wave images using a 42 cm field-of-view. Four slices were obtained including the level of the porta hepatis under 16 s breath-holding. Wave images and MRE images (stiffness map) with cross-hatching were automatically generated on the operating console. The initial default parameters for MRE were magnetization encoding gradients (MEG) = 60 Hz, TR/TE = 1000 ms/80 ms, 96 × 64 matrix (296 × 296 reconstruction matrix), and 5 mm slice thickness. When scanned with these default parameters, significant image degradation of the wave images and the stiffness map of MRE was occasionally observed, which was considered to be due to insufficient signal-to-noise ratio of the images. We therefore decided to increase the signal by decreasing the matrix to 64 × 64 (96 × 296 reconstruction matrix) and increase the slice thickness to 10 mm. Then, various

frequencies of MEG ranging from 60 to 120 Hz at every 10 Hz step were applied. TE were automatically set at their minimal values according to each MEG.

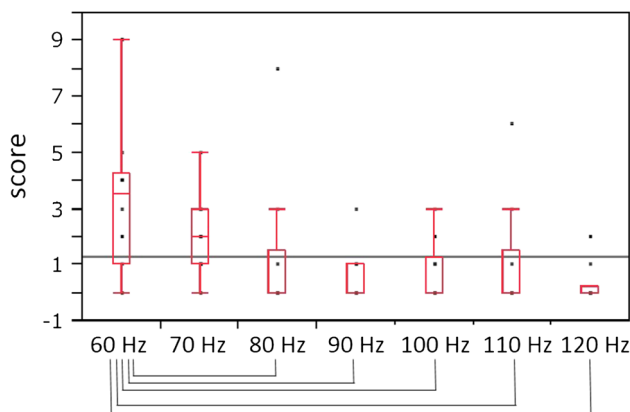
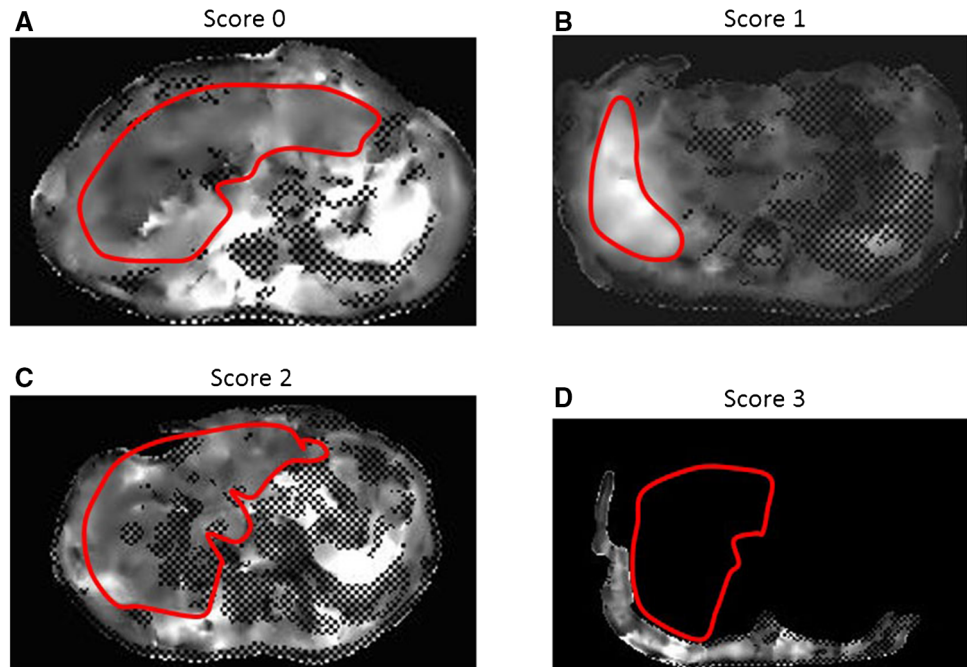
### Assessments

For the assessment of the image quality of MRE, we utilized three indices: the degree of the defects of MRE images (semi-quantitative index), the absolute values of the areas adequate for liver stiffness measurement, and its ratio to the whole liver area (quantitative indices). The degrees of image defects were visually graded for each slice by two radiologists (both with more than 10 years of experience as abdominal radiologists) in consensus, according to the definition as follows: scores 0 = no defect, score 1 = defect present but not hampering stiffness measurement of the liver, score 2 = defect partially hampering liver stiffness measurement, and score 3 = defect disabling liver stiffness measurement. The sum of the scores for four slices was regarded as the score of the patient, ranging from 0 to 12. Representative images for each score are shown in Fig. 1. The areas which were considered adequate for stiffness measurement were defined as those within the contour of the liver and without cross-hatching marks, using the fusion technique as reported previously [14]. Cross-hatching marks were introduced to the current version of MRE, indicating inadequate areas for measurement on the stiffness map, based on the amplitude and configuration on the wave images and signal to noise ratio (SNR) on the magnitude images [14]. Two radiologists independently traced the contours of the areas without cross-hatching marks within the liver and also those of the whole organ on the stiffness map fused with anatomical images. The measured values by the two radiologists were averaged, and adequate areas for measurement (AAM) in mm<sup>2</sup> and its ratio to the whole liver area (R-AAM) in percentage were used as indices for the evaluation. Images with significant defects (corresponding to scores 2 and 3, according to the previous assessment) were excluded for this quantitative evaluation. We also assessed the values of liver stiffness as measured by various MEGs, to see whether stiffness values are influenced by different MEGs.

After determining the optimal MEG, we further examined other factors, including external driver amplitude (50 vs 70 %), and slice thickness (8 vs 10 mm), and the measured liver stiffness, utilizing the same indices as the one used for the assessment of MEG frequency.

Using these optimized parameters, repeatability of measuring liver stiffness was validated by scanning the same volunteers in one week using Bland–Altman analysis and paired *t* test. For other comparisons among multiple groups, Tukey–Kramer's HSD test, which is corrected for type 1 error, was applied. Statistical software used was JMP (SAS institute Inc., version 9.0).

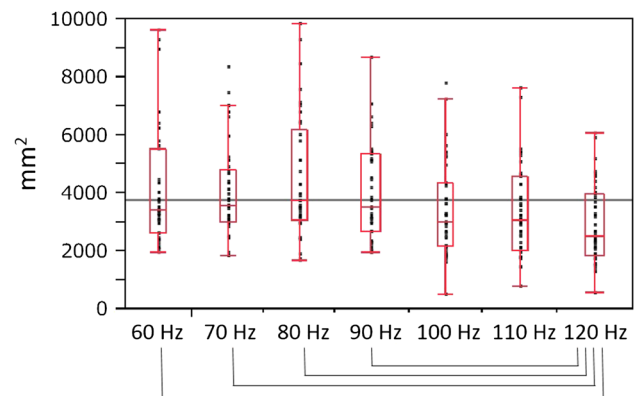
**Fig. 1** Examples of four scores in assessing the degree of image defects on MR elastography. The contour of the liver is shown in red lines. **a** Score 0, showing no image defect on stiffness map. **b** Score 1, showing image defect (around the region of the midline towards the left side of the anterior abdominal wall, but outside of the liver contour), not hampering liver stiffness measurement. **c** Score 2, showing small image defect (around the region of the right anterior abdominal wall within the contour of the liver), partially hampering liver stiffness measurement. **d** Score 3, showing substantial image defect disabling liver stiffness measurement (no signal is present from the liver)



**Fig. 2** Correlation between degree of image defects on stiffness map as expressed by the sum of the scores (Y-axis) and frequencies of magnetization encoding gradient (MEG) (X-axis). The most severe defects were observed at MEG of 60 Hz, and the least at MEG around 90 Hz and at 120 Hz. Statistically significant differences were present between 60 Hz and 80, 90, 100, 110, and 120 Hz, respectively ( $p < 0.05$ , Tukey–Kramer’s HSD test)

**Results**

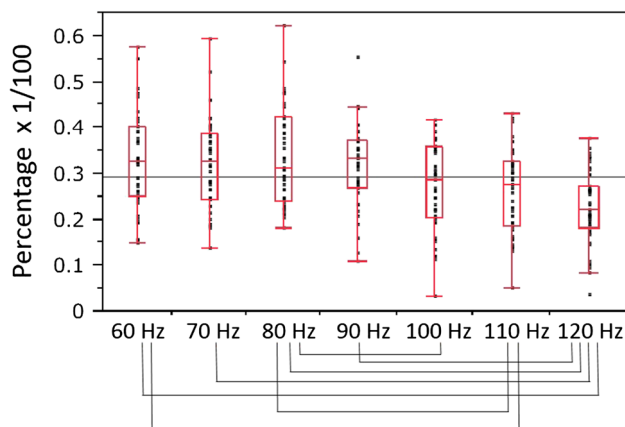
TEs for MEG frequencies 60, 70, 80, 90, 100, 110, and 120 Hz were 63, 60.7, 58.9, 57.5, 56.4, 55.5, and 54.7 ms, respectively. Regarding the degree of image defects on MRE according to the various frequencies of MEG, there were 280 images (4 slices per patient, 10 patients, 7 different MEGs) in total for review, among which scores 0, 1, 2, and 3 were given to 227, 32, 4, and 17 images, respectively. The most severe defects were observed at MEG of



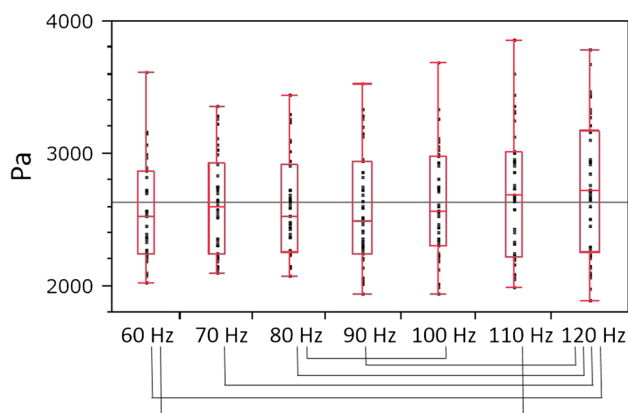
**Fig. 3** Correlation between areas adequate for stiffness measurement on stiffness map (in mm<sup>2</sup>, Y-axis) and frequencies of magnetization encoding gradient (MEG) (X-axis). MEG around 80 Hz show highest values, with statistical significance between 120 Hz and 60, 70, 80, and 90 Hz, respectively ( $p < 0.05$ , Tukey–Kramer’s HSD test)

60 Hz, and the least at MEG around 90 Hz and at 120 Hz. Statistically significant differences were present between 60 Hz and 80, 90, 100, 110, and 120 Hz, respectively. The detailed results are shown in Fig. 2.

As for AAM, 259 images excluding 21 images of scores 2 and 3 were reviewed. MEG around 80 Hz show highest values of AAM, with statistical significance between 120 Hz and 60, 70, 80, and 90 Hz, respectively (Fig. 3). As for R-AAM, similar trends were observed, with statistical significance between 120 Hz and 60, 70, 80, and 90 Hz, between 110 Hz and 60 and 80 Hz, and between 100 Hz and 80 Hz, respectively (Fig. 4).



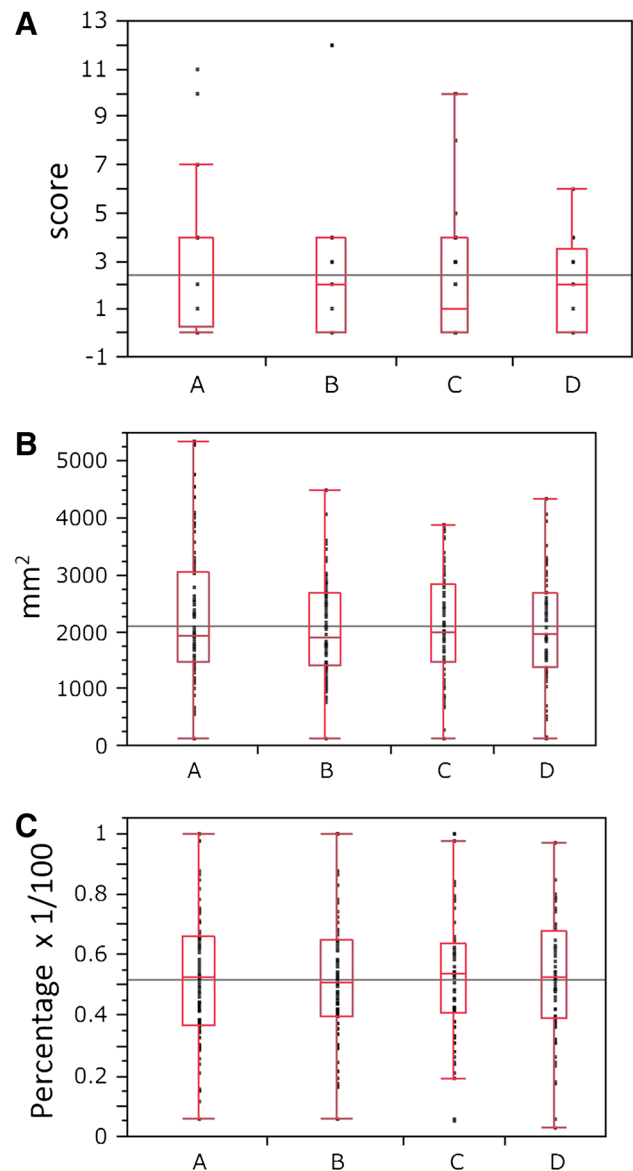
**Fig. 4** Correlation between the ratio of the areas adequate for stiffness measurement on stiffness map (Y-axis) and frequencies of magnetization encoding gradient (MEG) (X-axis). MEG around 80 Hz show highest values, with statistical significance between 120 Hz and 60, 70, 80, and 90 Hz, between 110 Hz and 60 and 80 Hz, and between 100 Hz and 80 Hz, respectively ( $p < 0.05$ , Tukey–Kramer’s HSD test)



**Fig. 5** Correlation between the measured values of the liver stiffness of volunteers (in Pa, Y-axis) and frequencies of magnetization encoding gradient (MEG) (X-axis). The measured liver stiffness was relatively consistent with MEG from 60 to 90 Hz; however, it was significantly higher with MEG of around 100–120 Hz ( $p < 0.05$ , Tukey–Kramer’s HSD test)

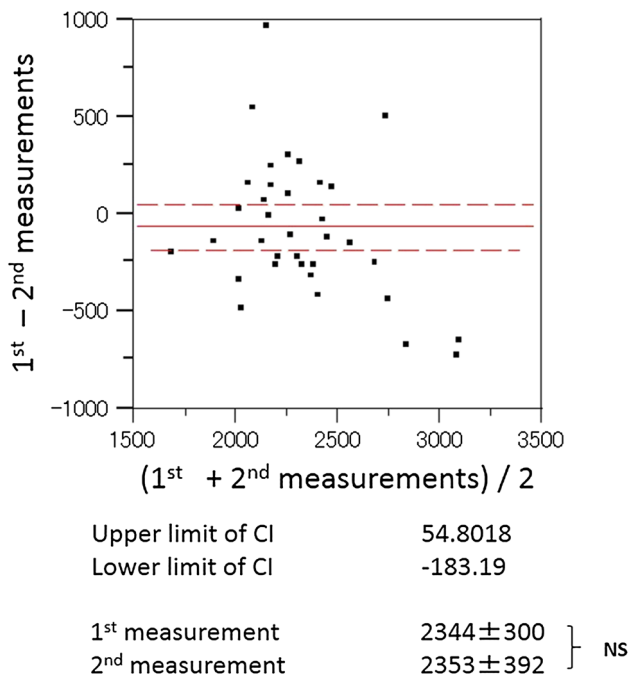
The measured liver stiffness was relatively consistent with MEG from 60 to 90 Hz; however, it was significantly higher with MEG of around 100 to 120 Hz (Fig. 5). From these results, 80 or 90 Hz was considered the optimal MEG.

With MEG fixed at 80 Hz, scans were repeated with MEG amplitudes/slice thickness = 70 %/8 mm, 50 %/8 mm, 70 %/10 mm, and 50 %/10 mm. There was no significant difference in any of the above-mentioned three indices among these four protocols (Fig. 6a–c). However, with MEG amplitude of 70 %, all volunteers felt some discomfort at the chest wall beneath the external acoustic driver, and therefore, 50 % amplitude was considered preferable.



**Fig. 6** Comparison among four protocols using various magnetization encoding gradient (MEG) amplitudes and slice thickness (thk) with MEG frequency fixed at 80 Hz. Protocols A, B, C, and D stand for MEG amplitudes/thk = 70 %/8 mm, 50 %/8 mm, 70 %/10 mm, and 50 %/10 mm, respectively. **a** Degree of image defects. There was no significant difference among the four protocols (Tukey–Kramer’s HSD test). **b** Areas adequate for stiffness measurement on stiffness map. There was no significant difference among the four protocols (Tukey–Kramer’s HSD test). **c** Ratio of the areas adequate for stiffness measurement on stiffness map. There was no significant difference among the four protocols (Tukey–Kramer’s HSD test)

With MEG frequency of 80 Hz, amplitude of 50 %, and slice thickness of 10 mm, MRE imaging was repeated in one week to validate the repeatability of MRE. Four pairs of images were excluded due to severe image defects, and 36 pairs of images were assessed. Between the scans, there was no significant difference in the liver stiffness values



**Fig. 7** Bland–Altman analysis to assess repeatability of liver stiffness measurement using optimized parameters ( $n = 36$ ). Bland–Altman analysis showed no significant trend between the initially and secondly measured values of liver stiffness, with 95 % confidence interval (CI) of within  $\pm 10$  % of the mean value. No significant difference was present between the initially and secondly measured values of the liver stiffness (paired  $t$  test). Stiffness values are shown in Pa (X- and Y-axes)

( $235 \pm 40$  vs  $234 \pm 30$  Pa, NS with paired  $t$  test), and 95 % confidence interval was  $-184.2$  to  $+54.3$  Pa (within  $\pm 10$  %) (Fig. 7, Bland–Altman analysis).

**Discussion**

Unlike the default parameters or the previously reported parameters for 1.5 T units, MEG frequency of 60 Hz did not work well for our 3.0 T clinical unit, and instead, 80 Hz was found to be the optimal MEG frequency. The precise reason why a frequency that is not synchronized to the external driver frequency (60 Hz) works better than a synchronized one is unclear, but it may at least be partially explained as follows: because TEs were automatically set at the minimum values at the current clinical 3.0 T unit, the higher the MEG frequencies, the shorter the TEs, resulting in a better signal-to-noise ratio and fewer defects on the original echo-planar magnitude images. Actually, in most of the 55 images with defects of scores 2 and 3, areas of insufficient signal were observed on the original echo-planar magnitude images corresponding to the areas of defects on the stiffness map. On the other hand, the degree of phase shift caused by the MEG implement should be

maximized at the frequency synchronized to the external acoustic vibration, namely 60 Hz, and as the MEG frequency grows further away from 60 Hz, the degree of phase shift becomes smaller, which would make it difficult to detect phase-shift on echo-planar phase images, resulting in increased cross-hatched areas on the stiffness map as shown in Figs. 4 and 5. In other words, 80 Hz can be said to be the best balanced point of MEG frequency from the viewpoint of TE-shortening effects and phase-shift maximizing effects. Although we did not have a gold standard for the actual value of the liver stiffness of our volunteers, the consistency of liver stiffness as measured with MEGs of 60–100 Hz may also justify the use of 80 Hz as an optimal MEG frequency. The reason why stiffness values were higher with MEG of 110 and 120 Hz is unclear and requires further investigation.

As for the repeatability of liver stiffness measurement, a 95 % confidence interval less than  $\pm 10$  % may be sufficient or at least acceptable for clinical use. Of course, our data are validated only for volunteers, whose liver stiffness measurements are much lower than in patients with chronic liver disease who are the actual target population of this new technology; thus, our data, namely 95 % confidence interval of  $\pm 10$  %, cannot simply be applied to the patient population in a strict sense. Repeatability in the chronic liver disease population should be validated in future investigation.

There are several limitations in the current study. First, we only tested a limited range of MEG frequencies, namely from 60 to 120 Hz, but judging from the obtained data and using degree of image defects and areas without cross-hatching as indices, it would be expected that images with MEG lower than 60 Hz and those with MEG higher than 120 Hz may suffer from more severe image defects and too much cross-hatching, respectively. Second, although we could successfully optimize parameters for MRE in our clinical 3.0 T unit, the precise mechanism or rationale behind them is still unknown; therefore our optimized parameters could be applicable only to our version of the 3.0 T MRE system, and further optimization may be required for another or different version of an MRE system.

In conclusion, parameters for MRE at our 3.0 T clinical unit were optimized as follows: external acoustic vibration frequency and amplitude 60 Hz and 50 %, respectively, MEG frequency 80 Hz, TR/TE = 10000/59 (minimum) ms, slice thickness 8–10 mm, and 64 matrix. With these optimized parameters, repeatability of  $\pm 10$  % (95 % confidence interval) in liver stiffness measurement was obtained in volunteers, which may be acceptable for clinical application.

**Acknowledgments** We greatly thank Dr. Richard C. Ehman, Department of Radiology, Mayo Clinic, for his invaluable advice and

comments on our discussion, and also Mr. Kazuyuki Uchiumi and Mr. Hiroyuki Kabasawa, GE Health Care Japan, for their expertise.

**Conflict of interest** The authors declare that they have no conflict of interest.

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