

Investigating the stability of thermal and physiological noise in phase domain using NORDIC for low-intensity MR-Acoustic Radiation Force Imaging (ARFI) in humans

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Introduction

The acoustic pressure waves generated by transcranial focused ultrasound (FUS) interact with cell membranes and can be used for neuromodulation of brain function¹. The acoustic force generated by FUS results in tissue displacements that have a local maximum at the focal point². MR acoustic radiation force imaging (MR-ARFI) uses a PGSE experiment to detect bulk displacement of the tissue². The sonication is performed during motion encoding gradient (MEG), and the displacements are encoded as phase changes (Fig 1). It has been demonstrated that FUS can map the focal point in animals^{3,4}.

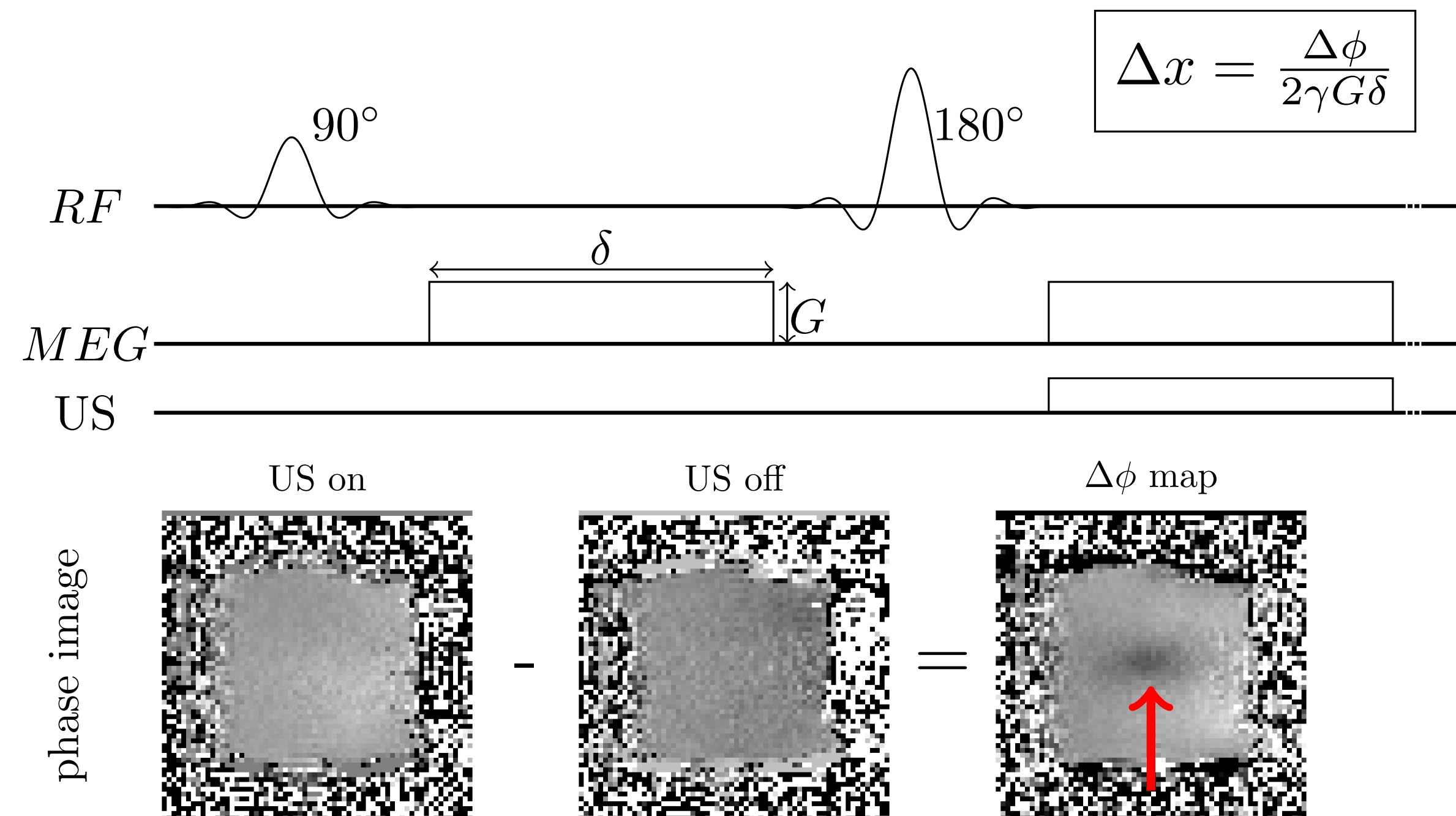


Figure 1: An example of MR-ARFI sequence diagram and difference maps acquired from phantom experiment. Ultrasound (US) is fired while one of motion encoding gradient (MEG) is on. The induced mechanical displacement (Δx) is not refocused, and results as phase changes pointed by a red arrow.

Only a limited number of centres currently apply FUS in an MRI system for neuromodulation. The application of MR-ARFI requires stringent safety control as both modalities have to be applied simultaneously giving a cumulative SAR effect. MR-ARFI is important to determine the location of the focal point, and the intensity of the delivered ultrasound. It forms an essential component of any system for closed-loop modulation. It was previously shown that low-intensity FUS can be used for MR-ARFI in humans⁵, however the expected phase changes are small enough to be easily buried by noise floor. To make low-intensity FUS MR-ARFI in humans feasible within the shortest possible measurement time, it is essential to understand both physiological and thermal noise contributions to MR phase images. In this study, we simulated MR-ARFI in a human subject with no sonication to assess how physiological and thermal noise impede MR-ARFI experiment.

Methods

A healthy volunteer was scanned, having provided written informed consent, at 3T MAGNETON Prisma with a 64-Ch head coil (Siemens Healthineers, Germany). For the MR-ARFI sequence, CMRR 2D diffusion weighted SE-EPI sequence was used with b -value = 200 s/mm^2 in readout direction (see protocol in Fig. 2 caption) with the MEG polarity alternated between successive volumes, resulting in the displacement being encoded as positive and negative phase respectively - increasing sensitivity in respect to standard MEG on off. Eddy current and motion correction were first performed on magnitude data, then the estimated correction parameters were applied to the phase data. Within a session, 100 phase difference maps were acquired.

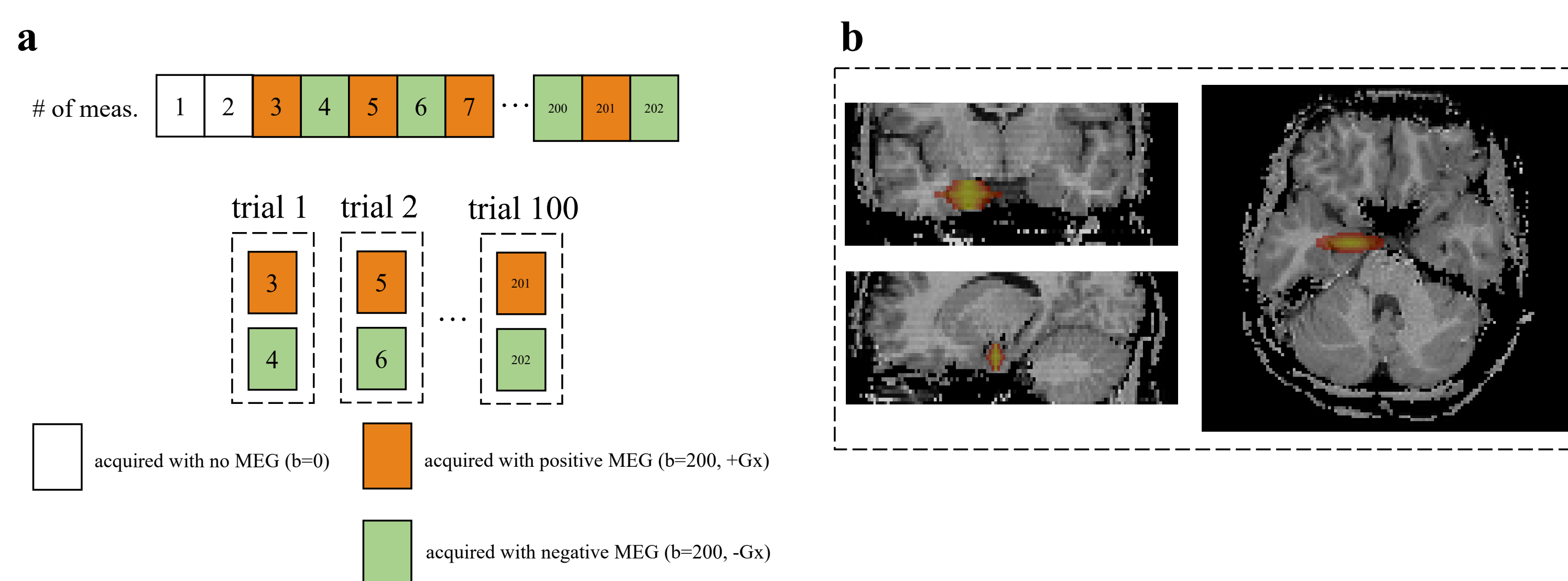


Figure 2: (a) Experimental paradigm and sequence parameters. After 2 measurements, diffusion with $b=200 \text{ s/mm}^2$ was applied in readout direction. Every other volume, the polarity of MEG was altered. Phase difference maps were computed trial by trial. Following parameters were used for DW-SE-EPI sequence; TR=1.5s, TE=52.8ms, 16 slices, slice thickness=3mm, $1.5 \times 1.5 \text{ mm}^2$ inplane, in plane GRAPPA=3, no PF, echo spacing = 0.77ms(b) Artificial phase changes displaced on T1-weighted anatomical image.

To investigate the number of volumes needed to robustly map the location of the displacement, the standard deviation in the phase difference map was computed as a function of number of trials. TR of 1.5 s is not fast enough to robustly sample cardiac pulsation that is a major source of artifacts in DWI data. We computed standard deviations from 50 random permutations of trials as a function of number of available trials, and took the mean to estimate the phase standard deviations across number of trials. The standard deviation was computed voxel by voxel. NORDIC⁶ denoising on the complex data associated to each of those permutations was used to improve sensitivity.

To mimic a low-intensity MR-ARFI dataset, an artificial induced phase change having a peak ± 0.025 , ± 0.05 , and ± 0.1 radian amplitude changes (Fig 2b). The data was added to the acquired phase data. Both original and the synthesized data were analyzed in the same way.

Results

Figure 3a shows that NORDIC clearly suppressed thermal noise in phase images, however strong phase instability remained across volumes and slices independent of MEG polarity. When phase difference maps were averaged over 100 trials, the instability was smoothed out (Fig 3b), although a background phase gradient along the MEG direction remained, presumably caused by uncanceled pulsation artifacts (if the origin was eddy currents, successive slices would show the same level of phase variation).

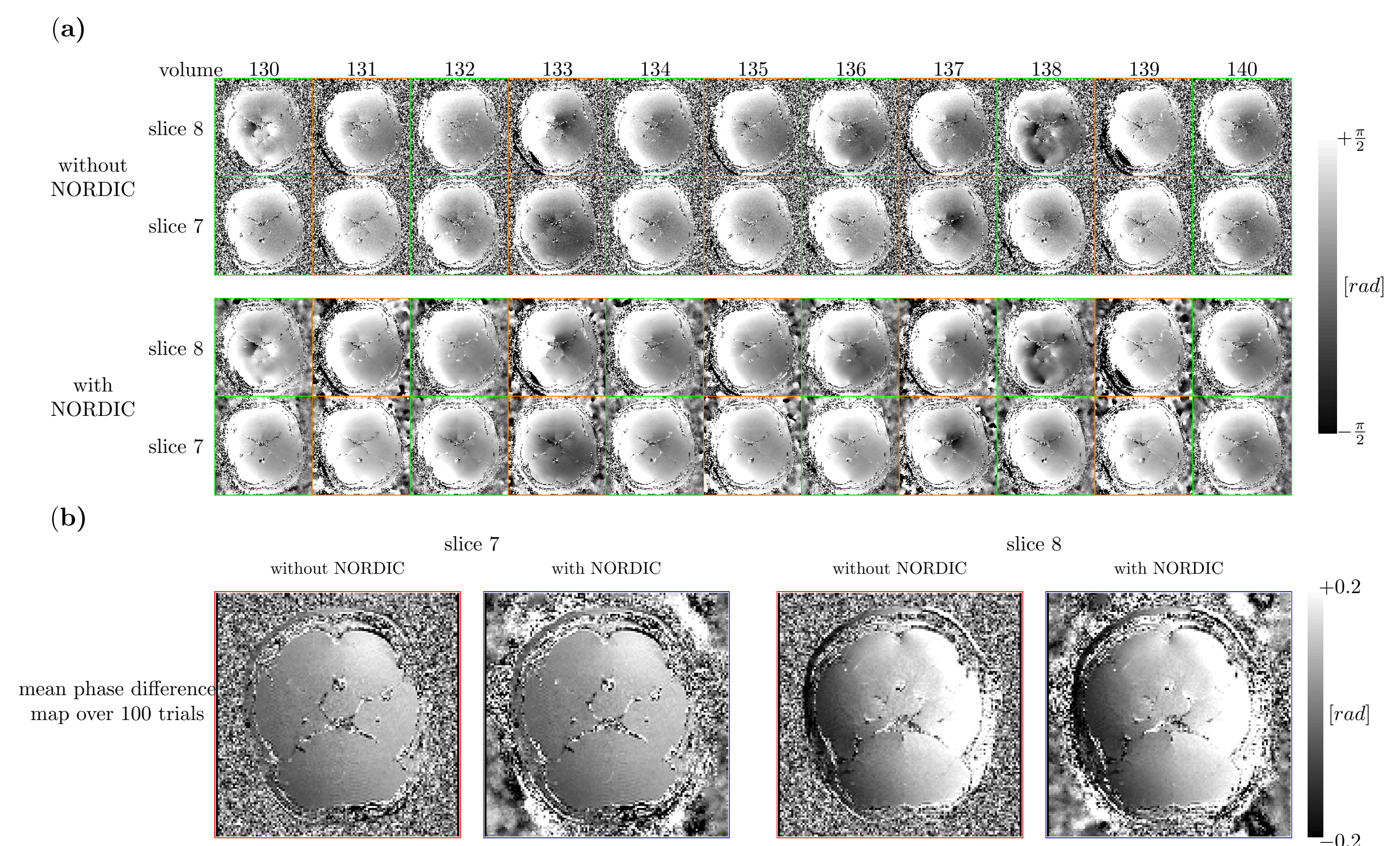


Figure 3: (a) Phase images from volume 130 to 140. (b) Phase difference maps averaged over 100 trials.

In Figure 4a, computed phase difference values without and with NORDIC denoising on original dataset are plotted. The shaded areas represent the standard deviations. The standard deviations in phase difference map decreased with increasing number of volumes. With less than 60 trials, NORDIC could suppress thermal noise effectively.

Figure 4b shows phase difference map from synthesized datasets computed. With 10 trials, when artificial phase amplitude was larger than noise standard deviation, NORDIC improved detectability (red arrows in Fig 4b.)

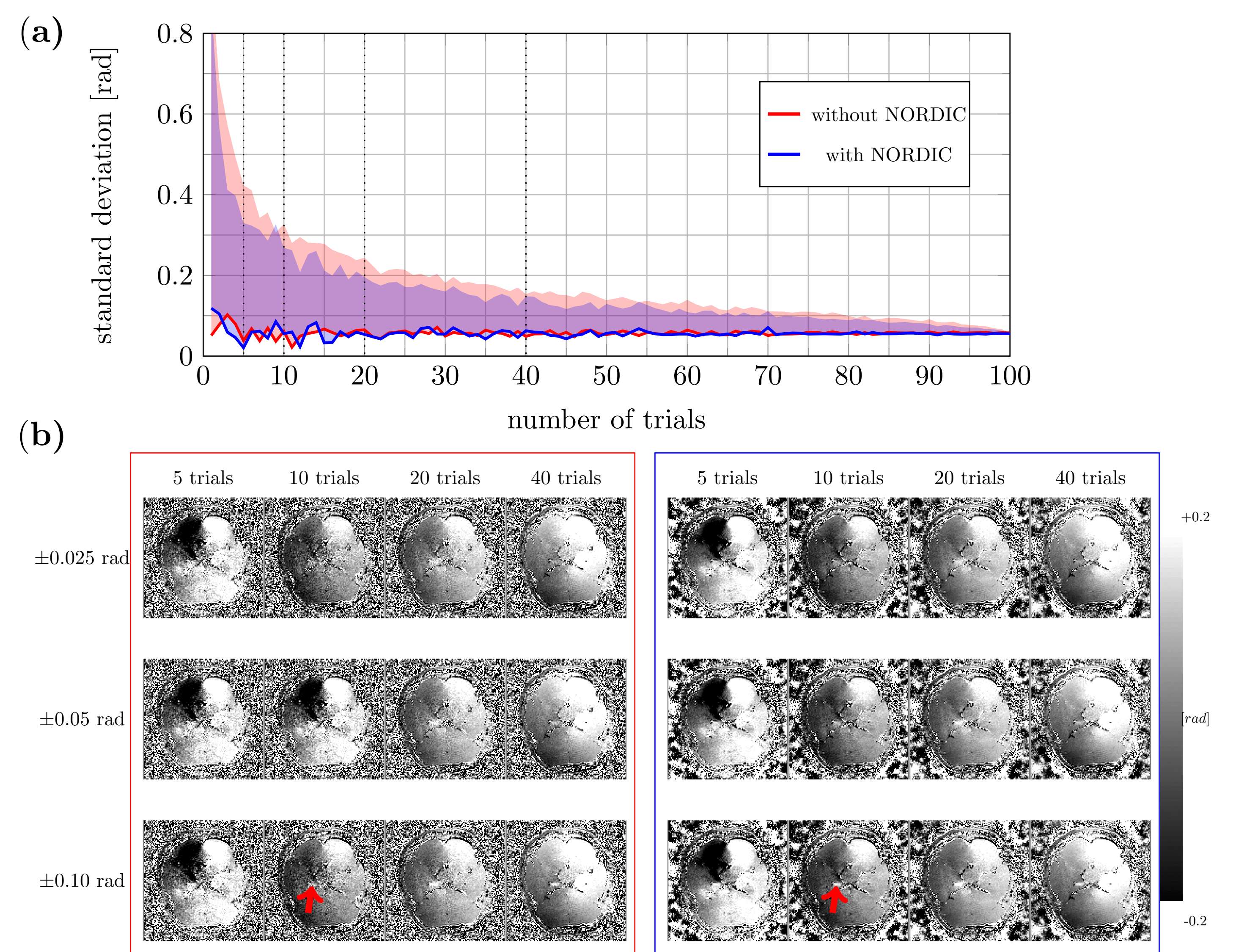


Figure 4: (a) Mean value and standard deviation in phase difference map with and without NORDIC. Solid lines and shaded areas represent mean the standard deviation. (b) Mean phase difference maps from synthesized dataset. Red arrow points where the synthesized phase changes were added.

Discussion and Conclusion

Physiological noise was more dominant than thermal noise. When phase change amplitude is under the physiological noise floor, NORDIC cannot help improving the detectability ($> \sim 60$ trials, 3 minutes of data). With physiological monitoring, a shorter TR acquisition could allow to sample physiological noise and its induced phase artifacts without temporal aliasing may allow us to robustly separate cardiac induced phase changes from displacement ones, although at the cost of reduced SNR and volume coverage.

NORDIC thermal noise denoising is effective with smaller number of trials where thermal noise dominates. Therefore, to further suppress noise floor with small number of trials (< 60 trials), either a technique to characterize and suppress physiological noise in phase images or a robust sequence for physiological noise is required for low-intensity FUS MR-ARFI in humans.