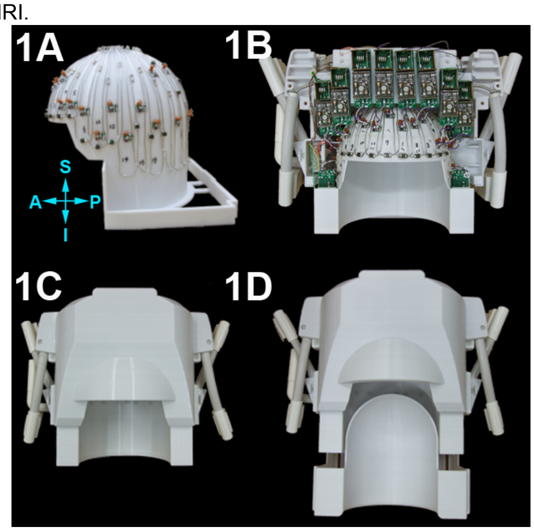
A 32-channel head coil array with circularly symmetric geometry for 2D accelerated 3D human brain imaging

Ying-Hua Chu[1], Boris Keil[2], Wei-Chao Chen[3], Wen-Jui Kuo[4], Fa-Hsuan Lin[1,2], and Fa-Hsuan Lin[5]

[1]Institute of Biomedical Engineering, National Taiwan University, Taipei, Taiwan, [2]Athinoula A. Martinos Center for Biomedical Imaging, Massachusetts General Hospital, Harvard Medical School, Charlestown, MA, United States, [3]SDI corporation, Chang-Hua, Taiwan, [4]Institute of Neuroscience, National Yang Ming University, Taipei, Taiwan, [5]Department of Biomedical Engineering and Computational Science (BECS), Aalto University, Espoo, Finland

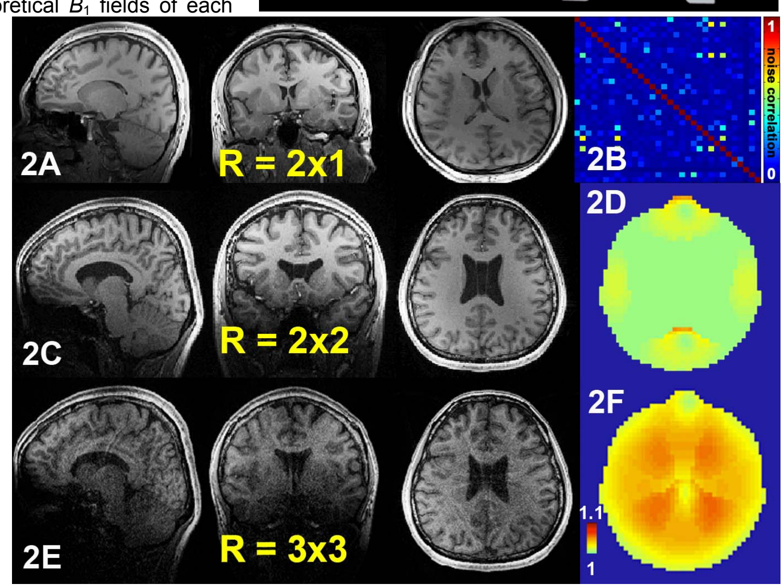
INTRODUCTION Array coils have been introduced as a way to increase signal-to-noise ratio (SNR) by simultaneous acquisition from multiple smaller surface coils with the ability for large field-of-view (FOV) coverage [1]. In parallel MRI (pMRI) reconstruction methods, the spatial modulation of the signal intensity of array elements is utilized to accelerate image encoding [2, 3]. Recently, 32-channel head arrays have become the standard type in use for clinical imaging [4]. Higher element brain arrays with up to 96 channels have been proposed for research studies [5]. The flexibility of trading-off the spatiotemporal resolution in pMRI critically depends on amount of independent spatial information in a coil array in order to un-alias folded images (SENSE [2]) or synthesize spatial harmonics (SMASH [3]). Considering a 3D acquisition with two phase encoding directions, we hypothesize that pMRI acquisitions using a coil array with a circular symmetry (CS) geometry in the plane of two phase encoding directions can improve the SNR efficiency in accelerated imaging. We developed a 3T 32-channel CS receive-only head coil array with slim curved trapezoidal shaped elements. Theoretical and empirical results demonstrate the advantage of this coil array design in 2D accelerated parallel MRI.

METHOD 32 trapezoidal shaped elements were tiled over the coil former in a longitudinal arrangement (Fig.1A), similar to the 16-channel head coil array reported previously [6]. The length of the elements were adjusted inividually to ensure simulatenously maximal head coverage and largest visual field. Each element has a witdh of approxximately 30 mm. Figures 1B, 1C, and 1D show the top view of the array together with the mechanical cover. All housing parts were 3D printed using polycarboneate (PC-ISO) plastic (FORTUS, Eden Prairie, MN, USA). Each coil element was tuned to Larmor frequency at 123.25 MHz with two distriubted capacitors. A capacitive matching network transforms the coil imedance to 50 Ohm under load.



For sufficient decoupling between the individual elements, all neighboring coils were critically overlapped and empircally optimized under a S21 measure. Decoupling of next nearest neighbors utilzed low-impedance preamplifiers (Stark Contrast, Erlangen, Germany), where the preamplifier's input impedance was transformed to a high-series impedance within the loop. An active detuning circuit was formed across the match capacitor using a variable inductor and a PIN diode. In addition to active detuning during Tx, series fuses with where added to each loop as additional protection during transmit.

Imaging experiments were performed on a 3T scanner (Tim Trio, Siemens, Erlangen, Germany). The MPRAGE pulse was used test the coil array (TR: 2530 ms, TE: 3.03 ms, TI : 1100 ms, Flip angle = 7^o, slices = 192, slice thickness: 1 mm, FOV: 256 mm). The noise covariance matrix was calculated using data collected without RF excitation. Given the coil array geometry, we used the Biot-Savart's law to calculate the theoretical B_1 fields of each channel. The empirical noise covariance matrix and the simulated coil sensitivity maps were used to calculated the *g*-factor maps [3] to estimate the noise amplification in 2D acccelerated imaging.Using a fully gradient encoded data set, 3D images with 3x3 2D accelerations was calculated by setting part of the k-space data to zero. Accelerated images from the MRI console were also obtained using the GRAPPA algorithm [7].



RESULTS The constructed coil shows S21 coupling between adjacent loops < -12dB and a preamp decoupling of -25dB. When isolated, the ratio between the unloaded quality factor (Q) to the loaded Q was 161/49=3.29. Figure 2A shows the sagittal, coronal, and transverse slice sum-of-square images using 2-fold GRAPPA acceleration. The noise covariance matrix of the CS array was shown in Figure 2B. The average and the maximal off-diagonal entries are 4.5% and 0.6, respectively. Figures 2C and 2E shows the 2x2 4-fold and 3x3 9-fold accelerated MPRAGE images. While 9-fold image become noisier, we can still discern gray and white matter clearly. The 2x2 and 3x3 fold g-factor maps is shown in Figures 2D and 2F. At 4-fold and 9-fold acceleration, the maximal g-factors are was less than 1.01 and 1.1 respectively.

DISCUSSION We designed and constructed a 32-channel head array coil with curved and tapered trapezoidal loop elements, which may be considered inefficient in detecting brain MRI signals in un-accelerated cases. However, in accelerated imaging the developed coil shows low additional noise amplifications during pMRI image reconstruction. Because of the geometry of the CS array designed, this symmetric arrangement makes construction of the coil to be more efficient and 2D accelerated images exploit the spatial information among channels evenly. We expect this coil may complement other 32-channel head arrays in neuroscientific and clinical applications to provide highly accelerated image SNR.

REFERENCES

1. Roemer, P.B., et al., Magn Reson Med, 1990. **16**(2): p. 192-225.
2. Sodickson, D.K. and W.J. Manning, Magn Reson Med, 1997. **38**(4): p. 591-603.
3. Pruessmann, K.P., et al., Magn Reson Med, 1999. **42**(5): p. 952-62.
4. Wiggins, G.C., et al., Magn Reson Med, 2006. **56**(1): p. 216-23.
5. Wiggins, G.C., et al., Magn Reson Med, 2009. **62**(3): p. 754-62.
6. de Zwart, J.A., et al., Magn Reson Med, 2002. **47**(6): p. 1218-27.
7. Griswold, M.A., et al., Magn Reson Med, 2002. **47**(6): p. 1202-10.