A 64-Channel Brain Array Coil for 3T Imaging

Boris Keil[1], James N Blau[1], Philipp Hoecht[2], Stephan Biber[3], Michael Hamm[3], and Lawrence L Wald[1,4]
[1]*A.A. Martinos Center for Biomedical Imaging, Department of Radiology, Massachusetts General Hospital, Harvard Medical School, Charlestown, MA, United States,* [2]*Siemens, Medical Solutions USA, Charlestown, MA, United States,* [3]*Siemens Healthcare, Erlangen, Germany,* [4]*Harvard-MIT Division of Health Sciences and Technology, Cambridge, MA, United States*

Introduction: In brain imaging, parallel acquisition has impacted clinical applications such that nearly every brain examination is performed with an array comprising multiple smaller surface coil elements. In this study, we design, construct, and evaluate a 64-channel brain array coil and compare it to a 32-channel coil constructed with the same coil former geometry in order to precisely isolate the benefit of the two-fold increase in array coil elements. The constructed coils were developed for a standard clinical MRI scanner in a way that addresses patient comfort concerns and can be disseminated for robust daily use. The coil is validated through SNR and G-factor map comparisons as well as highly accelerated in vivo brain imaging.

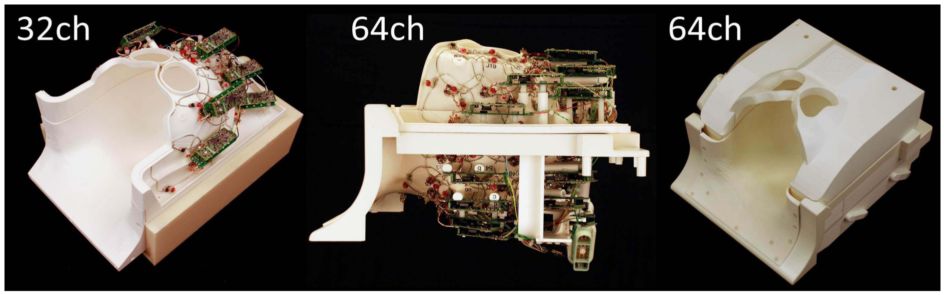

Fig.1: Constructed array coils for 3T brain imaging

Material and Methods: Both constructed coils (32ch and 64ch) used an overlapping, approximately circular element design on an anatomically shaped former consisting of a large posterior head part and an overlapping anterior head portion (Fig.1). The former is 20cm wide at the widest spot and utilizes a rounded posterior section and slightly tapers in at the neck. The 64ch array used 42 elements on the posterior part and 22 on the anterior. For the 32ch array, these numbers are 18 and 14, respectively. The larger posterior section is designed so the subject can lie down into the coil (►Fig.1), rather than a helmet design which must be pulled down over the head. The anterior head segment closes the helmet via a snap-in mechanism and overlaps with the posterior portion to allow the loops on the two halves to be geometrically decoupled. Large eye cut-outs facilitate visual stimulation for functional studies. We used hexagonal and pentagonal tiling patterns [1] to accommodate the 3D circuitry of the element arrangement. All neighboring elements are decoupled by critical overlap except the two eye loops which were decoupled using a shared conductor design. Each loop inductance was divided symmetrically with discrete components joining the semi-circles (a variable tuning capacitor at the top and the output circuit with matching capacitors and active detuning trap at the bottom). While nearest neighbor decoupling was addressed with critical overlap, next-nearest neighboring coil elements were decoupled using preamplifier decoupling [2] by transforming the low impedance of the preamplifier to an open circuit in the loop using a 5.5cm long coax cable and a series capacitor. Pairs of coils were attached to a preamp pair sharing a circuit board (Siemens Healthcare), which also served to mix the detected signal to an intermediate frequency and frequency domain multiplex the two channels onto a single coaxial output [3]. Data were acquired on a 3T Siemens Skyra MRI system with 64 receive channels. SNR and G-factor maps used PD-weighted spin echo brain images (TR/TE/α=300ms/15ms/20°, 5mm slice, matrix=256×256, FOV=240x240mm^2). Finally, the array performance was tested in highly accelerated anatomical brain MPRAGE image (TI/TR/TE= 1100ms/2530ms/3.49ms, 1mm isotropic, matrix=256×256, FOV=256x256mm^2).

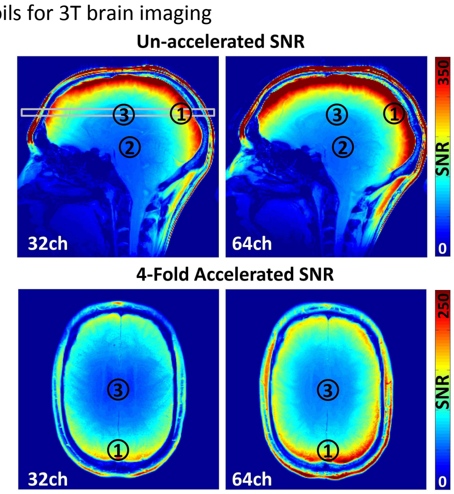

Fig.2: SNR map comparison between both coils in un-accelerated and 4-fold accelerated cases.

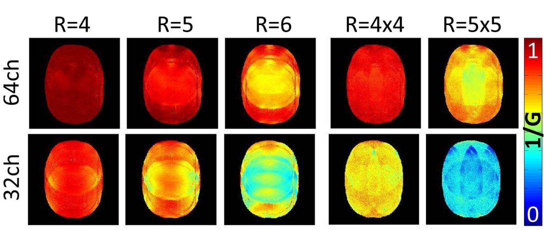

Fig.3: 1/G maps of transversal slice for 2D and 3D acceleration

Results: The coil shows *S*21 decoupling between nearest neighbors of -17dB and a preamplifier decoupling of -21dB. The Q_U/Q_L-ratio was 263/45=5.9 and 252/23=10.9 for typical elements in the 64ch and 32ch coil, respectively. The SNR increased at the location of the brain cortex by 1.7-fold compared to the 32ch coil (►Fig.2, ROI1). The SNR in the brain center was modestly improved by 5% using the 64ch coil (ROI2). ROI3 measured a 1.1-fold SNR increase. In 4-fold accelerated imaging, the 64ch coil shows larger SNR improvements; a 1.26 fold increase in the central brain area (ROI3) and 1.8-fold increase in the cortex (ROI1) compared to the 32ch array. The higher gains derive from the improved G-factors of the 64ch array. ►Figure 3 shows that the accelerated images obtained from the 64ch brain array provides the ability to accelerate at approximately one unit higher at a given noise amplification compared with the 32ch array.

Conclusion: A 64-channel brain array coil was constructed, tested and compared to a sized matched 32ch array coil. Compared to the 32ch coil, the 64ch coil provides significant improvements in peripheral SNR and overall SNR improvements in highly accelerated imaging acquisitions. Robust and compact design considerations were implemented to facilitate clinical studies. Thus, the highly parallel brain array is well-suited for brain studies with improved sensitivity.

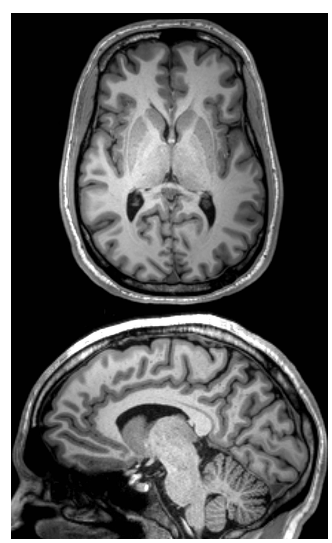

Fig.4: 64ch MPRAGE, 1mm iso, R=4, acquisition time: 3.39min.

References: [1], Wiggins GC, et al. MRM 2006;56;216-223. [2] Roemer PB, et al. MRM 1990,16;192-225. [3] Keil B, et al. 19th ISMRM 2011, p.160.