

A 64-Channel Array Coil for 3T Head/Neck/C-spine Imaging

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Introduction: 32-channel brain arrays have become increasingly used for clinical brain imaging and larger arrays (up to 128ch) have shown potential in research applications in abdominal [1], cardiac [2], and brain [3] imaging. In this study we extend the coverage of brain-only designs by placing 40 elements on the brain and adding 24 close-fitting elements along the neck, C-spine and face to support integrated head-neck-spine examinations. The array interfaces with a standard clinical scanner with workflow optimization allowing T and L-spine imaging thru incorporating the clinical spine array via software selection of array elements. For Head/C-spine studies, the superior row (4 elements) of the table-top spine array is integrated into the 60 elements of the head/neck array. The coil is validated through SNR and G-factor map comparisons as well as highly accelerated imaging of the brain/neck and C-spine showing improved SNR and the potential for 2D acceleration factors up to $R=3 \times 3$.

Material and Methods: The coil was constructed on an anatomically shaped former consisting of three parts (Fig.1): a large posterior head-neck part with 38 elements, an anterior head portion with 18 elements, and an anterior neck section with 4 elements. This 60ch coil is combined with the upper row of 4 elements of the spine coil to form a 64ch array covering the head, neck and C-spine. The coil formers are based on the surface 95% contours of aligned 3D MRI scans and adjusted to a standard adult head size and printed in ABS plastic using a rapid prototyping 3D printer (Dimension, Inc., MN, USA). We used hexagonal and pentagonal tiling patterns [4] to accommodate the 3D circuitry of the element arrangement. For sufficient decoupling between the individual elements, all neighboring coils are critically overlapped (except for the posterior neck part). Also, the two eye loops were decoupled using a shared conductor design. Each loop inductance was divided symmetrically with discrete components joining the semi-circles (a variable tuning capacitor and fuse 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 where possible, next-nearest neighboring coil elements were decoupled using preamplifier decoupling [5] performed 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 low noise converters (Siemens Healthcare) which also served to down convert the detected signal to an intermediate frequency and frequency domain multiplex the two channels onto a single coaxial output. The signals were later de-multiplexed back into individual channels in the receiver. Thus the 60ch coil required 30 output coaxes. Data were acquired on a 3T Magnetom Skyra MRI system (Siemens Healthcare, Erlangen, Germany) with 64 receive channels. For SNR and G-factor comparison, T1-weighted spin echo images ($TR/TE/\alpha=300\text{ms}/15\text{ms}/90^\circ$, $SL=5\text{mm}$, $matrix(m)=256 \times 256$) were obtained using a head-neck phantom. Brain data were compared to a 32ch adult 3T head coil. Head-neck data were compared with a 20ch head-neck coil. Finally, the array performance was tested in highly accelerated anatomical brain imaging (T2-SPACE, $R=3 \times 3=9$, $TR/TE/\alpha=3200\text{ms}/410\text{ms}/120^\circ$, 0.9mm iso, $m=256 \times 256$, $FOV=24 \times 24\text{cm}^2$, $\#SL=192$, $TA=1:05\text{min}$) and C-spine imaging (T2-TSE, $R=3$, $TR/TE/\alpha=3500\text{ms}/112\text{ms}/160^\circ$, $SL=3\text{mm}$, $m=320 \times 256$, $FOV=22 \times 22\text{cm}^2$, $\#SL=13$).

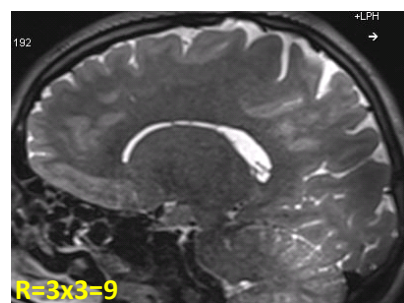


Fig.3: 3D T2 SPACE, 0.9mm isotropic, $R=3 \times 3=9$, acquisition time = 1.05min.

Results: The coil shows S21 decoupling between nearest neighbors of -17 dB and a preamplifier decoupling of -22dB. The Q_u/Q_L -ratio for the receive coils was $252/33=7.6$. 1/G-factor maps for head-neck images show improved imaging performance compared to the 20ch and the 32ch coil (Fig.2). The SNR increased at the location of the brain cortex by 2.4-fold and 1.2-fold compared to the larger 20ch head-neck and the 32ch brain array coil, respectively (measured in ROIs Fig.2). The SNR in the brain center was similar in all three coils with slightly improved (~3%) SNR using the 64ch coil. In the C-spine ROI, the 64ch array coil showed an 1.8-fold increase in SNR compared to the 20ch head/neck array and considerably more than the brain-only 32ch array. Volunteer tests have shown that acceleration factors up to $R=9$ (3×3) were feasible, enabling a 0.9mm isotropic resolution T2 scan with acquisition time of 1.05 min (Fig.3). Figure 4 shows an accelerated ($R=3$) *in vivo* C-spine T2 image.

Conclusion: A 64ch head-neck-C-spine array coil was constructed and tested. In the brain, the sensitivity matches the 32ch brain array with small improvements in the cortex and also modest improvements in accelerated imaging capabilities. More importantly, the array extends the pattern of high sensitivity into the face, neck and C-spine; regions not covered by the 32ch brain array. The 64ch coil provides further improvements in SNR and accelerated imaging performance in both the brain, neck and spine compared to the larger 20ch coil. Thus, the highly parallel coil is well-suited for either brain examinations or head-neck-spine studies.

References: [1] Hardy CJ *et al.*; JMRI (2008) 28:1,1219-25. [2] Schmitt M *et al.*; MRM (2008) 59:6,1431-9. [3] Wiggins GC *et al.*; MRM (2009) 116:7,1332-5. [4] Wiggins GC *et al.*; MRM (2006) 56:1,216-23, [5] Roemer PB *et al.* MRM (1990), 16:2,192-225.

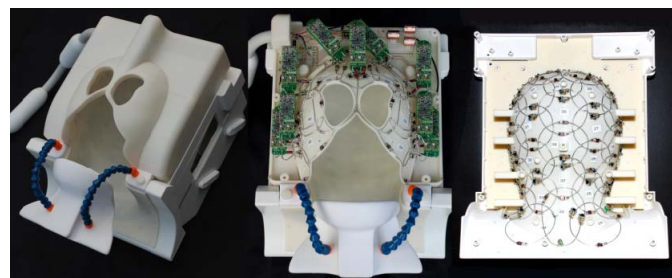


Fig.1: Constructed Head-Neck-C-spine array coil

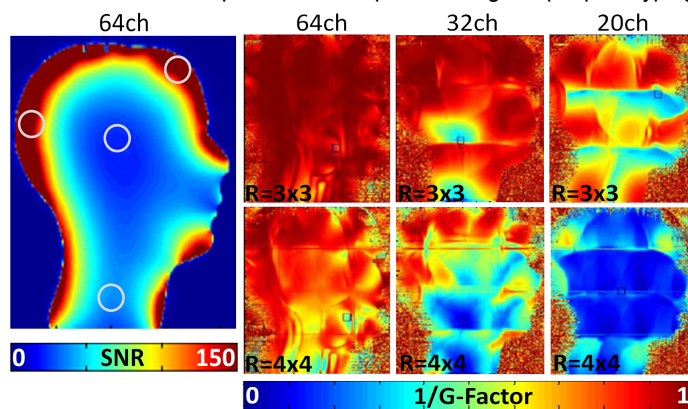


Fig.2: (Left) SNR map (left) of 64ch coil. (Right) 1/G-maps of sag. slice for 2D acceleration



Fig.4: T2 Cspine, $R=3$