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Exploring Functional Connectivity Networks with Multi-Channel Brain Array Coils

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Abbreviations

12Ch: 12 Channel

32Ch: 32 Channel

BOLD: Blood Oxygenation Level Dependent

aCompcor: anatomical Component based noise Correction method

CNR: Contrast-to-Noise Ratio

CSF: Cerebrospinal Fluid

DAN: Dorsal Attention Network

DMN: Default Mode Network

ECN: Executive Control Network

EPI: Echo Planar Imaging

fMRI: functional Magnetic Resonance Imaging

fcMRI: functional connectivity Magnetic Resonance Imaging

GM: Grey Matter

HCMN: Hippocampal Cortical Memory Network

N_{ch} : Number of array elements

RF: Radio Frequency

ROI: Region of Interest

SNR: Signal-to-Noise Ratio

SN: Saliience Network

tSNR: time-series Signal-to-Noise Ratio

WM: White Matter

ABSTRACT

Using multi-channel array head coils in functional and structural MRI provides increased Signal-to-Noise Ratio (SNR), higher sensitivity and parallel imaging capabilities. However, their benefits remain to be systematically explored in the context of resting state functional connectivity MRI (fcMRI). In this study, we compare signal detectability within and between commercially available multi-channel brain coils, 32-Channel (32Ch) and a 12-Channel (12Ch) at 3T, in a high-resolution regime to accurately map resting state networks. We investigate whether the 32Ch coil can extract and map fcMRI more efficiently and robustly than the 12Ch coil using seed-based and graph-theory based analyses. Our findings demonstrate that although the 12Ch coil can be used to reveal resting state connectivity maps, the 32Ch coil provides increased detailed functional connectivity maps (using seed-based analysis) as well as increased global and local efficiency, and cost (using graph-theory analysis), in a number of widely reported resting state networks. Exploration of subcortical networks, that are scarcely reported due to limitations in spatial-resolution and coil sensitivity, also proved beneficial with 32Ch coil. Furthermore, comparisons regarding the data acquisition time required to successfully map these networks indicated that scan time can be significantly reduced by 50% when a coil with increased number of channels (i.e. 32Ch) is used. Switching to multi-channel arrays in resting state fcMRI could therefore provide both detailed functional connectivity maps and acquisition time reductions, which could further benefit imaging special subject populations, such as patients or pediatrics who have less tolerance in lengthy imaging sessions.

INTRODUCTION

The improved sensitivity afforded by array coils was described in their first application (Roemer et al., 1990). The increased sensitivity (especially near the array) stems from the improved geometric coupling between small inductive elements and nearby spins compared to larger surface coils (Hayes et al., 1985). The array concept allows the sensitivity of the small surface coil to be extended over greater areas. The sensitivity benefit deep from the surface is smaller since the smaller diameter surface coils have steeper sensitivity drop-offs with depth (Hayes et al., 1985). But as the number of array elements, N_{ch} , used to tile a fixed area increases as the element size decreases, the added benefits of N_{ch} nearly independent measurements of the deep voxel exactly cancels the effect of the increased drop-off. The net effect is a sensitivity at the center of the brain that is comparable to the larger elements. Electro-magnetic simulation studies have been shown, that the best possible detection (ultimate SNR) in the center of a head-sized uniform spherical sample is already approachable with as few as 8 channels at 3T (Wiesinger et al., 2004). At the periphery, the sensitivity grows approximately linearly with the number of elements and therefore larger numbers of elements are required for approaching the theoretical SNR limit. Results from experimental coil array studies were found to be quantitatively in-line with the results obtained from the simulation studies (Keil et al., 2012; Wiggins et al., 2009).

The increased sensitivity from array coils in a time-series functional MRI (fMRI) data set translates directly to higher BOLD contrast sensitivity. The BOLD contrast-to-noise ratio (CNR) is proportional to the time series signal-to-noise ratio (tSNR): $CNR = tSNR TE \Delta R2^*$, where TE and $\Delta R2^*$ derives from the biological response of the tissue and is field dependent, but not

dependent on other acquisition sequence choices. Thus, the most acquisition parameters such as coil choice and voxel size affect BOLD CNR through tSNR.

Previous studies (Triantafyllou et al., 2011) have shown that in comparison to standard coils (single channel), multi-channel arrays offer improvements in fMRI time-series SNR when medium to small size voxel volumes are used, whereas in larger voxel sizes the improvements are modest, primarily because physiological noise (originating from fluctuations such as cardiac, respiratory, and hemodynamic induced signal modulations) increases with voxel size. For example, 32Ch coil improves the tSNR of the $1.5 \times 1.5 \times 3 \text{ mm}^3$ acquisition by 48% compared to 12Ch coil; the increase, however, is only 11% at low resolution ($5 \times 5 \times 3 \text{ mm}^3$) (Triantafyllou et al., 2011). Additionally, higher resolution fMRI acquisitions could potentially increase spatial specificity and localization of the resting state networks, while minimizing partial volume effects and thru-plane signal dropouts, due to thinner slices.

Although the acquisition parameter space in fcMRI have already been investigated thoroughly by Van Dijk and colleagues (Van Dijk et al., 2010), the additive sensitivity from advances in multi-channel array coils remains to be explored. In this study we therefore evaluate the detectability power of multi-channel arrays in resting state fcMRI at a high resolution EPI regime, where we are expecting to achieve the biggest benefits from the high-N arrays. We investigate some of the most widely reported networks including the default mode network (DMN) (Greicius et al., 2003), the hippocampal-cortical memory network (HCMN) (Vincent et al., 2008), the dorsal attention network (DAN) (Corbetta and Shulman, 2002), the executive control network (ECN), and the salience network (SN) (Seeley et al., 2007). In order to investigate the SNR improvements in deeper brain regions with the 32Ch array we also examine seed-based connectivity in sub-cortical regions (basal ganglia network) of the brain. In addition,

we evaluate the benefit of higher sensitivity of the 32Ch array coil by measuring the effective reduction in acquisition time to accurately map intrinsic correlations using fMRI.

METHODS

Data Acquisition

Data acquisition was performed on a Siemens 3T scanner, MAGNETOM Trio, a Tim System, (Siemens AG, Healthcare Sector, Erlangen, Germany), using two different commercially available Radio Frequency (RF) receive-only head coils; a 12Ch and a 32Ch brain array coils (Siemens AG, Healthcare Sector, Erlangen, Germany). The 32Ch array consists of 32 loop elements set in the soccer-ball geometry as described in the literature by Wiggins et al (Wiggins et. al 2006). The product version is a split-type design with the anterior part consisting of 12 elements and the posterior of 20 elements. The 12Ch coil is the vendor's "head matrix coil" product which is the default coil to the 3T Tim Trio system, Siemens. This coil combines 12 long elements in one ring. The whole-body transmit coil was used for excitation in both cases. The same 16 healthy volunteers (7 males), all right-handed, age range: 18-33 years, (mean age: 25 ± 5) were scanned on both RF coils using the same acquisition protocol. Written informed consent was obtained from all subjects for an experimental protocol approved by the institutional review board. Extra padding with foam cushions was used for head immobilization. To avoid any possible bias, the total number of subjects starting the experiment with any given coil was kept equal in the study. This was achieved by counterbalancing the type of coil that the subject starts and ends the session with, as data from both coils were acquired during the same imaging session. All subjects were asked to relax while in the scanner with their eyes closed, and instructed not to fall asleep. Automatic slice prescription, based on alignment of localizer scans

to a multi-subject atlas (van der Kouwe et al., 2005), was used to achieve a consistent slice prescription across the two imaging experiments with the different RF coils. However, given that the two coils vary in size, with the 32Ch being smaller and tighter fit, subject positioning could not be identical in the two coils, however we ensured consistent subject positioning within each coil. Specifically, we used similar under-head padding and foam cushions laterally and on top of the head (posterior of the coil) to minimize motion and to ensure each subject was positioned comfortably in the head coils. The mean obliquing parameters across subjects were $T > C - 19.6 \pm 6^\circ > S 1 \pm 1.9^\circ$ for the 12Ch coil, and $T > C - 18.3 \pm 6.5^\circ > S 1.6 \pm 1.5^\circ$ for the 32Ch coil, where T, C, and S denote Transverse, Coronal and Sagittal planes respectively; there was no significant difference in these parameters between the two coils.

Resting-state time-series were acquired using a single-shot gradient echo EPI sequence. At the beginning of each EPI acquisition, two “dummy” scans were acquired and discarded to allow longitudinal magnetization to reach equilibrium. Full head coverage was achieved with sixty-seven 2mm thick interleaved slices with orientation parallel to the anterior commissure – posterior commissure (AC-PC) plane. The imaging parameters were TR=6000 ms, TE=30 ms, flip angle=90°, in-plane spatial resolution of 2 mm x 2 mm and 62 time-points. Each resting scan lasted 6 minutes and 24 seconds. The TR was chosen to be 6 s in this study in order to achieve full-brain coverage at the given resolution of 2mm isotropic voxel size (without utilizing parallel imaging). Full brain coverage was essential in order to map global resting state networks, e.g. Default Mode Network. Array data was combined with the manufacturer’s Sum-of-Squares online reconstruction method. Additionally, a 3D high-resolution T₁-weighted structural scan was collected using an MP-RAGE sequence with voxel size = 1.3x1x1.3 mm³, other acquisition parameters were: TR/TE/TI/FA=2530 ms/3.39 ms/1100 ms/7°. Additionally, to demonstrate the

effect of coil geometry/design and the coil sensitivity profile on the various brain regions, proton density weighted gradient echo images were acquired from the same subject in both coils at the same scanning session. Acquisition parameters were: TR/TE/FA = 30ms/6ms/30°, Matrix: 192×192, Field-of-View: 170×170 mm², slice thickness: 7 mm, bandwidth = 200 Hz/Pixel. Noise data was also acquired with this acquisition scheme, but with 0V RF excitation.

Data Analysis

The resting state data were pre-processed with standard fMRI pre-processing steps using SPM8, (fil.ion.ucl.ac.uk/spm/software/spm8/) (Friston, 2007), including: *i*) a six parameter rigid body transformation to account for head motion, and perform image realignment, *ii*) slice-time correction to account for the interleaved slice acquisition, *iii*) normalization using a voxel size of 2x2x2 mm³ and the EPI template provided with SPM8 to allow comparison between subjects, and, *iv*) smoothing with 3-mm full-width-half-maximum (FWHM) Gaussian kernel. T₁-weighted structural images were segmented to grey matter (GM), white matter (WM) and CSF masks using the segmentation routine in SPM8 (Ashburner and Friston, 2005). The original structural image and the segmented images were also normalized using a voxel size of 1x1x1 mm³ and the T₁-weighted structural template provided with SPM8. Subject motion was evaluated with in-house custom software (nitrc.org/projects/artifact_detect/). At a motion threshold of 0.4mm, there were a total of 21 outliers in the 12Ch data set and 17 in the 32Ch data set (16 subjects per group). Since there were no significant differences in the mean number of outliers between 12Ch and 32Ch coils, nuisance regression of motion outliers was not carried out. Additionally, there was no significant difference (p=0.37) in the mean motion parameters between the 12Ch coil (mean motion = 0.49±0.41) and the 32Ch coil (mean motion = 0.38±0.23).

The SNR maps were calculated following the methodology from Kellman and McVeigh (Kellman and McVeigh, 2005). Noise correlations coefficients matrices were calculated from the noise only (RF=0V) acquisitions.

First-Level Connectivity Analyses

Functional connectivity analysis was performed using both seed-based and graph-theory approaches with MATLAB (MathWorks, Natick, MA) based custom software package: CONN (Whitfield-Gabrieli and Nieto Castanon, 2012). For seed-based analysis, sources will be defined as multiple seeds corresponding to the pre-defined seed regions for: (i) DMN and HCMN, (ii) DAN, (iii) ECN and (iv) SN. All seeds were independent of our data and were generated using WFU_PickAtlas, (nitrc.org/projects/wfu_pickatlas) (Maldjian et al., 2004; Maldjian et al., 2003). Seeds for DMN, DAN, ECN and SN were chosen to be 10-mm spheres centered on previously published foci (Zhang and Raichle, 2010), while HCMN seeds were chosen to be 12.5-mm spheres centered at coordinates provided by the literature (Vincent et al., 2008). Detailed description of the seed regions used are given in Table I. For the subcortical (basal ganglia) network, the sources were anatomical ROIs corresponding to: (i) thalamus, (ii) striatum (caudate and putamen), (iii) globus pallidus (medial and lateral), (iv) substantia nigra and (v) subthalamic nucleus, derived from WFU_PickAtlas. For graph-theory based analyses all 84 Brodmann areas, anatomically defined from the Talairach Daemon database atlas (Lancaster et al., 2000), were chosen as sources.

Table I around here

The seed time-series went through temporal band-pass filtering ($0.008 < f < 0.09$ Hz). Instead of removing the average signal over all voxels of the brain by global signal regression, contributions from non-neuronal sources, such as WM and CSF were considered as noise, the

principal components of which were estimated and removed using, aCompCor (anatomical component based noise correction method) (Behzadi et al., 2007). The optimal configuration of the aCompCor approach (Chai et al., 2011) as applied in the CONN toolbox (Whitfield-Gabrieli and Nieto Castanon, 2012) was followed. In addition, the six motion (3-rotation and 3-translation) parameters were also regressed out. For quality control purposes, it was ensured that the histogram plot of voxel-to-voxel connectivity (r value) appear approximately centered to the mean for each subject after confound removal. Correlation maps were generated by extracting the residual BOLD time-course from the seeds, followed by computing Pearson's correlation coefficients between the seed time-course and the time-courses of all other voxels. Correlation coefficients were converted to z- scores using Fisher's r-to-z transform to allow for second-level General Linear Model (GLM) analyses. Images from the first-level results (correlation maps and z-maps) provide the seed-to-voxel connectivity maps for each selected source for each subject and for each condition (one per subject/condition/source combination).

Second-Level Connectivity Analyses

For both seed-based and graph-theory-based methods, we first performed within- and between-group analysis of full data sets from 32Ch and 12Ch coils. 32Ch_{full} and 12Ch_{full} refer to “full-length” acquisitions of 6min and 24 s with 62 time-points. In addition, scan time reduction was evaluated by estimating the 32Ch_{half} vs. 12Ch_{full} contrast (within- and between-group analysis) to examine if sufficient signal power is held by the 32Ch data in the shorter run; 32Ch_{half} for “half-length” acquisition with 31 time-points. The outcome of GLM analyses performed at this level was the within-subjects linear combination of effects specified by the *sources* as contrasts, and applied to the first-level connectivity-measure volumes (for the seed-to-voxel analyses). For within-group comparisons of seed-based analyses, whole-brain False Discovery Rate (FDR)

corrected threshold of $p < 0.05$ ($p_{FDR-corr} < 0.05$) was used to identify areas of significant functional connectivity. For between-group comparisons of task-positive and task negative networks, statistical analysis was performed using a cluster-defining voxel-wise height threshold of $p < 0.01$ (uncorrected). Since subcortical regions are noisier compared to cortical regions, a whole-brain $p_{FDR-corr} < 0.05$ was used for between-group comparisons. For all the networks, significant clusters were identified with an extent threshold of whole-brain Family Wise Error (FWE)-corrected $p < 0.05$ ($p_{FWE-corr} < 0.05$).

For the graph-theory based analysis [see (Bullmore and Sporns, 2009) for review], we chose **Global Efficiency**, **Local Efficiency** and **Cost**. These metrics are particularly relevant for probing brain networks because of their computational validity for unconnected and weighted graphs (Achard and Bullmore, 2007). *Global efficiency* of a node is the average inverse shortest-path distance between a given node and all other nodes in the network (targets). *Local efficiency* of a node is the average inverse shortest-path distance among the target nodes connected to a given node. *Cost* or *Degree* of a node is the proportion of nodes connected to a given node. Equivalent network-level measure of these metrics is the average (across all nodes in the network) of their corresponding node-specific measures. The computational formulas are given below (in Equations 3,4 and 5) where $|G|$, E and C denote the number of nodes (n) in graph G , efficiency and cost respectively.

Global Efficiency:

$$E^{global}(G) = \frac{1}{|G|} \cdot \overset{\circ}{\underset{n \in G}{\sum}} E_n^{global}(G) \quad (1)$$

Local Efficiency:

$$E^{local}(G) = \frac{1}{|G|} \cdot \overset{\circ}{\underset{n \in G}{\sum}} E_n^{local}(G) \quad (2)$$

Cost:

$$C(G) = \frac{1}{|G|} \cdot \sum_{n \in G} C_n(G) \quad (3)$$

Global Efficiency of a node is the ‘centrality’ of the nodes’ connectivity, i.e., the extent of connectivity of the node with the rest of the network, whereas on the network level, it serves as a measure of the extent of centrality as well as the “efficiency” of this connectivity (nodes with higher global efficiency are “better connected”). In contrary, *Local Efficiency* of a node represents the ‘locality’ of the nodes’ connectivity, i.e., the extent of connectivity of the node with its neighbors (as well as the “redundancy” or fault tolerance of the node), whereas on the network level, it provides a measure of the extent of locality i.e., nodes with high local efficiency are connected to neighbors that form a strong or well-connected local network, while nodes with low local efficiency are connected to neighbors that are sparsely connected or distant to each other. Global efficiency typically reflects the relevance of long-range connections (meaning higher global efficiency = better long-range connectivity); where as local efficiency is reflective of the relative relevance of short-range connections in the overall network connectivity (meaning higher local efficiency = better short-range connectivity). Finally, *Cost* of a node can be interpreted as the strength of connectivity of a node, whereas on the network level, cost indicates hypo/hyper connectivity in the overall network (e.g., higher cost=overall hyper-connectivity). The approach used is part of the CONN toolbox and has been described in great detail in a recently published manuscript (Whitfield-Gabrieli and Nieto Castanon, 2012).

To contrast network-level estimations of global efficiency, local efficiency and cost, a fixed percentile cost threshold (top 15% of ROI-to-ROI connectivity) was used to calculate

connectivity (adjacency) matrix (within the 84 Brodmann area ROIs), followed by a threshold of $p_{FDR-corr} < 0.05$, for both within- and between-group comparisons.

RESULTS

Figure 1 shows the coils' sensitivity on a human subject in terms of pixel-wise image SNR maps (top row) as well as the noise correlation matrices across individual elements from the 12Ch and 32Ch array coils (bottom row). Data was acquired from the same subject in both coils at the same scanning session. The comparison reveals that the 32Ch array outperforms the 12Ch coil by a factor of 2.3x averaged over the given signal area. The peripheral cortex, corpus callosum, and midbrain (thalamic) regions show SNR improvements by a factor of 2.7x, 1.4x, and 1.25x respectively, exhibiting the increased sensitivity offered by the 32Ch coil not only at the cortex, but also at sub-cortical regions and deeper structures. The 12Ch and 32Ch coil show average noise correlation of 12.3% and 9.7%, respectively. Some of the correlation is likely due to remaining inductive coupling, but some is due to shared resistance through the sample.

Figure 1 around here

Seed-based Analysis

Figure 2 shows group-level results for the task negative default networks (DMN and HCMN) from 32Ch and 12Ch array coils. Connections in all the seeds in the DMN (Figure 2A) and HCMN (Figure 2B) are significantly stronger in the 32Ch data set. Connections in left and right Inferior Temporal Gyrus (ITG) extending to left and right Parahippocampal Gyrus (PHG), Superior Parietal Cortex (SPC) and Middle Temporal Gyrus (MTG) are more significant for DMN in the $32Ch_{full} > 12Ch_{full}$ comparison (Table II). Even with half the data set (Table III),

connections within the Superior Frontal Gyrus (SFG), Superior Parietal Lobule (SPL) and Superior Temporal Gyrus (STG) were revealed with the 32Ch coil. For HCMN, connections in the left and right Secondary Visual Cortex (SVC), Orbitofrontal Cortex (OFC) and left and right MTG were significantly more pronounced in the $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ comparison (Figure 4A). Medial Pre-frontal Cortex (PFC) was significant even with $32\text{Ch}_{\text{half}} > 12\text{Ch}_{\text{full}}$ contrast (Table III).

Figure 2 around here

Table II around here

Table III around here

Group-level results for the task positive networks (DAN, ECN and SN) from 32Ch and 12Ch coils are shown in Figure 3. The 12Ch coil only revealed a small subset of the functional connectivity in DAN (Figure 3A). Connections in dorso-lateral PFC (DLPFC), left and right fusiform gyrus, anterior PFC (APFC), ITG and SPL were significantly stronger in the 32Ch data set (Table II). $32\text{Ch}_{\text{half}} > 12\text{Ch}_{\text{full}}$ comparison (Table III) revealed DLPFC, APFC, premotor cortex and SPL.

Figure 3 around here

Figure 3B shows the functional connectivity correlation maps generated at the second level for ECN. The $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ comparison (Table II) revealed significant differences in SFG, left and right MTG, DLPFC, SPC, left and right APFC and premotor cortex. The contrast $32\text{Ch}_{\text{half}} > 12\text{Ch}_{\text{full}}$ (Table III) also revealed significant differences, primarily in APFC. Entire network (dorsal medial PFC, left and right APFC and left and right SPC) was significantly stronger with half the data set from 32Ch coil (Figure 4B).

Figure 4 around here

Similarly to all the above-mentioned networks, the second level analysis for SN (Figure 3C) revealed only a smaller subset of the network for the 12Ch coil. Connections in left and right insular cortex were remarkably stronger with the 32Ch (both full and half data sets) in comparison to 12Ch coil (Table II). In addition, $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ comparison revealed dorsal Anterior Cingulate Cortex (ACC), APFC, left and right DLPFC, and SPC (Table III).

For all the resting state networks studied above, both $12\text{Ch}_{\text{full}} > 32\text{Ch}_{\text{full}}$ and $12\text{Ch}_{\text{full}} > 32\text{Ch}_{\text{half}}$ contrasts were not significant.

Figure 5 around here

To explore the sensitivity of the coils in deeper brain structures, the subcortical network of basal ganglia. Group level results based on seed-to-voxel analysis demonstrated stronger functional connectivity in all the sub-cortical ROIs with 32Ch coil (Figure 5). $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ comparison (whole-brain $p_{\text{FDR-corr}} < 0.05$, cluster-level $p_{\text{FWE-corr}} < 0.05$) revealed significantly stronger connections in bilateral pallidum, bilateral putamen, left precentral gyrus (BA 6) and STG (BA 22). ROI-to-ROI connectivity analysis for within-group comparisons ($p_{\text{FDR-corr}} < 0.05$) revealed several interesting findings: 1) With thalamus as seed ROI, 12Ch group failed to identify functional connectivity from relatively smaller structures such as substantia nigra and subthalamic nucleus; this was also the case when medial globus pallidus was chosen as the ROI; 2) With caudate (part of striatum) as seed ROI, correlations with medial globus pallidus was detected only by the 32Ch group, and 3) With lateral globus pallidus as seed ROI, the T-scores for positive correlations from substantia nigra and subthalamic nucleus were: 6.27 and 5.67 with 32Ch coil; and 2.26 and 2.17 with 12Ch coil.

Figure 6 around here

Graph-theory Based Analysis

Consistent with the small-world behavior of brain networks reported before (Achard and Bullmore, 2007), graph-theory analyses revealed monotonic increases in global and local efficiency as a function of cost in all brain networks (Figure 6). As shown in Figure 6 top row, the random graph had higher global efficiency than the lattice and vice versa, for costs (K) in the range $0 \leq K \leq 0.5$. Brain networks (solid black line pertaining to our data represents data from all subjects for both 12Ch and 32Ch coils), however in the cost range of $0.05 \leq K \leq 0.34$, had global efficiency greater than the lattice but less than the random graph, and local efficiency greater than random but less than lattice (Fig.6, bottom row). We therefore chose a cost threshold of 0.15 for our analyses. In the $32Ch_{full} > 12Ch_{full}$ contrast, analysis of global efficiency ($p_{FDR-corr} < 0.05$) only left and right ACC (BA 33) surpassed the top 15% ROI-to-ROI connectivity (cost threshold of 0.15) from the network of all sources (84 Brodmann areas). Results from network level analysis of cost are shown in Figure 7. Left and right ACC, left and right anterior entorhinal cortex (BA 34) and right perirhinal cortex (BA 35) surpassed the threshold ($p_{FDR-corr} < 0.05$) for $32Ch_{full} > 12Ch_{full}$ contrast (Figure 7C). Even with half the data set from 32Ch coil, analysis of cost ($p_{FDR-corr} < 0.05$) in the $32Ch_{half} > 12Ch_{full}$ contrast revealed bilateral ACC. Analysis of local efficiency ($p_{FDR-corr} < 0.05$) revealed right ACC and left PHG (BA 36). The opposite contrasts ($12Ch_{full} > 32Ch_{full}$ and $12Ch_{full} > 32Ch_{half}$) were not significant for all three measures. Global and local efficiency comparisons of full and half data sets from 32Ch, and full data sets from 12Ch coil at the cost threshold of 0.15 are summarized in Table IV. T-values from $32Ch_{full}$, $12Ch_{full}$, and $32Ch_{full} > 12Ch_{full}$ comparisons from the network level analysis of cost (depicted in Figure 7) for the top 15% ROI-to-ROI connectivity ($p_{FDR-corr} < 0.05$) from all Brodmann areas (number of nodes = 84, 16 subjects per group) is provided in Table V. Brain regions provided in column 1, correspond to the ROIs represented as circles in Figure 7C ($32Ch_{full} > 12Ch_{full}$

comparison). Precisely, these are the brain regions that surpassed the threshold ($p_{FDR-corr} < 0.05$) for $32Ch_{full} > 12Ch_{full}$ contrast during the network-level analysis of cost.

Figure 7 around here

Table IV around here

Table V around here

DISCUSSION

Multi-channel arrays offer close head-fittings, impressive increases in the image SNR, especially in cortical areas, and remarkable accelerated imaging capabilities. Because of these advantages, multi-channel array coils have recently become widely available as experimental devices (Wiggins et al. 2006, Keil et al., 2011; Keil et al., 2012) and also as clinical research tools (Knake et al., 2005, Parikh et al., 2011, Raoult et al., 2011). In this study, task positive (dorsal attention, executive control and salience), task negative (default mode and hippocampal cortical memory), and subcortical (basal ganglia) resting state networks were assessed to examine whether increases in tSNR with the additive coil sensitivity of a 32Ch brain array can translate to higher functional connectivity detectability when compared to a 12Ch coil. Our findings, from both seed-based and graph-theory-based functional connectivity analyses methods, demonstrated that the 32Ch brain array revealed stronger connections ($32Ch_{full} > 12Ch_{full}$ contrast) in all the resting state networks studied. Furthermore, precise localization of functional connectivity mapping was also observed using the 32Ch coil, when fMRI time-series acquisition time was reduced to half of its original duration (~3min vs. 6min scan).

The increased SNR capabilities of 32Ch coil in combination with the high resolution acquisition scheme enabled us to identify the resting state networks at the group level (16

subjects) in greater detail compared to the 12Ch coil. The core of the default mode network, according to the literature, is formed by PCC, MPFC, left and right LPC, and left and right inferior temporal cortices (Fox et al., 2005). The fact that the Inferior Temporal Gyrus was detected to a significantly stronger extent by the 32Ch coil, reflects one of the major limitations of 12Ch coil in terms of SNR at higher resolution acquisitions. Furthermore, connections from relatively smaller brain structures, like PHG, which are not typically identified even at group level by 12Ch coil, was detected by the 32Ch coil in our group data. This was particularly accurate in the HCMN comparison between the two coils. Significantly better detection of connections between hippocampus and orbitofrontal gyrus, offer a clear advantage for using 32Ch coil in studies involving hippocampal-orbitofrontal connectivity, particularly in the context of epilepsy (Catenoix et al., 2005). For the Dorsal Attention Network (DAN), at which FEF, IPS and MTG comprise the core signal components (Fox et al., 2005), functional connectivity was extremely weak, particularly in the frontal and parietal cortices when the data from the 12Ch coil was used. Core signal components of the Executive Control Network (ECN), IFG, MFG and SFG (Seeley et al., 2007), as well as insula and cingulate cortex, that comprise the Salience Network (SN) (Taylor et al., 2009), were significantly stronger for $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ comparison. Impairments in executive network have been suggested in social anxiety disorders (Qiu et al., 2011), where the 32Ch coil could offer a clear benefit. Right insular cortex even surpassed a much stronger threshold (whole-brain $p_{FDR\text{-corr}} < 0.05$) for $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ contrast. This is of particular significance in light of previous research (Sridharan et al., 2008) and more recently in addiction studies (Sutherland et al., 2012), which showed that right-fronto insular cortex is a network hub that plays a critical role in initiating the spontaneous switching between the task-positive (ECN) and task-negative (DMN) networks (Honey et al., 2007).

To explore further the CNR advantages of the 32Ch we have investigated functional connectivity in the deeper structures of the thalamus and basal ganglia (i.e. sub-cortical network). Between group comparisons ($32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$) revealed significantly stronger connections in bilateral pallidum, bilateral putamen, left pre-central gyrus, superior temporal gyrus, as well as within the basal ganglia structures, consistent with recent reports using 16Ch coil at 7T (Lenglet et al., 2012). Furthermore, the 12Ch coil failed to identify connections with substantia nigra and subthalamic nucleus, which are relatively smaller structures and typically excluded from analysis when low-resolution acquisition ($3 \times 3 \times 3 \text{ mm}^3$) are employed (Di Martino et al., 2008). This result is in agreement with the SNR performance showing in Figure 1, where the 32Ch array outperforms the 12Ch coil in SNR by a factor of 1.25x and 2.7x at the central and peripheral cortex respectively. By increasing the channel count of a head array coil from 12 to 32, an overall 1.8-fold SNR improvement can be expected (Wiesinger et al., 2004). The main SNR gain contribution is expected to occur at the peripheral regions (e.g. brain cortex), while the central SNR will remain relatively the same. However, product available 32Ch head array coils provide a tighter fit compared to dimensionally larger designed 12Ch coils. For the arrays under study in the present work (32Ch and 12Ch coils) since the 32Ch array is also constructed on a tighter fitting helmet than the 12Ch (or most other arrays), it also enjoys a sensitivity benefit from the closer proximity between receive-element and brain. This benefit extends to deep structures as well as superficial cortex. Increases in SNR obtained from 32Ch coils can then be traded off for acquisitions with higher spatial resolution, which becomes particularly important for fMRI.

Graph theory analyses revealed significantly higher overall global efficiency of nodes (i.e. stronger connections) with 32Ch coil compared to 12Ch coil, particularly in ACC. There is converging evidence from recent publications that functional connectivity in ACC changes

through brain maturation in healthy subjects (Kelly et al., 2009) and also in ADHD patients (Qiu et al., 2010). Significantly higher local efficiency of PHG revealed by 32Ch coil compared to 12Ch coil showing that this region formed a strong/well-connected network with its neighbors. In the $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ contrast, there is a trend ($p=0.056$) towards higher global efficiency. This could be indicative of the sensitivity of the 32Ch coil to detect long-range connections. In particular, differences in global efficiency between networks are typically related to differences in the amount of long-range connectivity within the network for small-world networks. Similarly, significantly different local efficiency ($p=0.034$) in the $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ contrast indicates the sensitivity of the 32Ch coil to detect short-range connections (because short-range connections are associated to higher local efficiency, i.e., how well are still its neighbors connected if we eliminate this node). Since these two metrics are vital to understanding brain's ability to integrate information at the global level (i.e., functional integration) and cluster level (i.e., functional segregation) (Rubinov and Sporns 2010), 32Ch coil would prove more beneficial to elucidate the intricacies of brain networks. Cost advantages of 32Ch coil are depicted in Figure 7, which provides a graph visualization of the network-level analysis of cost for the top 15% ROI-to-ROI connectivity ($p_{\text{FDR-corr}} < 0.05$) from all Brodmann areas (number of nodes = 84; 16 subjects per group). Network level analysis of cost revealed that connections in ACC are significantly stronger in both $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ and $32\text{Ch}_{\text{half}} > 12\text{Ch}_{\text{full}}$ comparisons. Perirhinal, entorhinal, and parahippocampal cortices are part of the medial temporal lobe (MTL) and based on our results, especially from graph-theory analysis, we suggest that 32Ch coil would be better suited for studies involving MTL pathologies such as Alzheimer's disease and epilepsy. Significantly higher functional connectivity, observed in inferior/medial temporal regions with 32Ch coil, consistently in DMN, HCMN, DAN and ECN reiterates this benefit.

In this study time-series data were acquired in a relatively higher spatial resolution ($2 \times 2 \times 2 \text{ mm}^3$) compared to typically employed low resolution ($3 \times 3 \times 4 \text{ mm}^3$) in fMRI, based on findings from a recent work (Triantafyllou et al., 2011) which demonstrated that array coils provide biggest increases in tSNR at high spatial resolutions (small voxel size). In our resting state protocol we chose to acquire data at a 2mm isotropic voxel size to utilize the benefits of the multichannel array as well as to increase the spatial specificity and localization of the networks and minimize partial volume effects and physiological noise contamination.

In functional MRI (fMRI) studies, events/blocks are repeated several times so that task related activations are detected more reliably. This often leads to long experiments inducing subject fatigue and/or head motion, the levels of which may confound the results. Moreover, such long experiments might not be feasible on specific subject populations, such as pediatrics or patients. Typically, reduction in scan time is possible only at the expense of SNR, but not necessarily if one could capitalize on the increased sensitivity afforded by multi-channel arrays or high magnetic field strength. In this study, we demonstrate that increases in tSNR offered by 32Ch coil can also translate to reductions in scan time, i.e. less number of time points per functional run or ultimately less runs of the same experiment in fMRI. In our resting state experiments, particularly, connections within SFG were significantly stronger in the $32\text{Ch}_{\text{half}} > 12\text{Ch}_{\text{full}}$ comparison for both DMN and HCMN, as were DLPFC, APFC, premotor cortex and SPL in DAN (task-positive network). Similarly, the ECN and the salience network revealed, within network, significantly strong connections with half the duration of the 32Ch fMRI acquisition. Similarly, this can also be translated to power calculations for group fMRI studies, (Mumford and Nichols, 2008), by having a lesser sample size or preventing the collection of additional data that will have little impact on power.

The signal dynamic range of the 32Ch receive coil in the brain is approximately twice as that of the 12Ch head coil. This steeply varying spatial sensitivity profile of the small receiver coil elements of the 32Ch array has two important consequences. Firstly, it causes a non-uniform detection sensitivity that spatially modulates the ability to detect BOLD fluctuations. This adds to other sources of BOLD detection variation such as biological effects (differences in CBF and CBV responses and differing hemodynamic response functions) as well as other instrumental effects such as imperfect B_0 shimming (which creates $T2^*$ variation and subsequently degrades the optimality of the TE setting). The effect of the coil sensitivity can be easily visualized by creating a tSNR map of the resting brain. Alternatively, the BOLD sensitivity map (Deichmann et al., 2002; Gorno-Tempini et al., 2002) includes this information through the explicit incorporation of the image signal intensity profile.

The most problematic issue concerning the spatially varying reception is the increased sensitivity to motion. Motion effects in resting state have well known detrimental consequences (Power et al., 2012; Satterthwaite et al., 2012; Van Dijk et al., 2012), which are exacerbated when parallel imaging acceleration is used incorporating reference data or coil sensitivity maps taken at the beginning of the scan. Movement then leads to changing levels of residual aliasing in the time-series. Even for non-accelerated imaging, problems derive from the spatially varying signal levels present in an array coil image. Even after perfect rigid-body alignment (motion correction), the signal time-course in a given brain structure remain modulated by the motion of that structure through the steep sensitivity gradient. Motion correction (prospective or retrospective) brings brain structures into alignment across the time-series but does not alter their intensity changes incurred from movement through the coil profiles of the fixed-position coils. This effect can be partially removed by regression of the residuals of the motion parameters; a

step that has been shown to be very successful in removing nuisance variance in ultra-high field array coil data (Hutton et al., 2011). An improved strategy might be to model and remove the expected nuisance intensity changes using the motion parameters and the coil sensitivity map.

As it has been already demonstrated, the achievable SNR improvements of the 32Ch over the 12Ch head coil at the cortex but also at deeper brain areas are due to the increased number of elements and the tight fitting helmet design of the 32Ch array, respectively. However, in practice there are potentially two limitations associated to the tight fitting design; a) not all head sizes fit in the helmet and b) there is no room for the commonly used MRI compatible headphones with big earmuffs. For the later, alternative solutions should be consider, for example inner-ear headphones or ultra-slim earmuffs available in the market.

CONCLUSIONS

In this work we demonstrated that the improved signal detection capability of the 32Ch coil and its higher sensitivity result in increased functional connections and stronger correlation strengths, which potentially offer opportunity for smaller sample size in group level statistics and therefore preventing additional data collection. Our characterization of multi-channel arrays was performed on the particular design and coil manufacturer. Other multi-channel arrays might offer different degree of sensitivity in the cortical brain areas compared to deeper structures due to variability in the design configuration.

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Disclosure Statement

No competing financial interests exist.

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FIGURE CAPTIONS

Brain Connectivity
Exploring Functional Connectivity Networks with Multi-Channel Brain Array Coils (doi: 10.1089/brain.2012.0113)
This article has been peer-reviewed and accepted for publication, but has yet to undergo copyediting and proof correction. The final published version may differ from this proof.

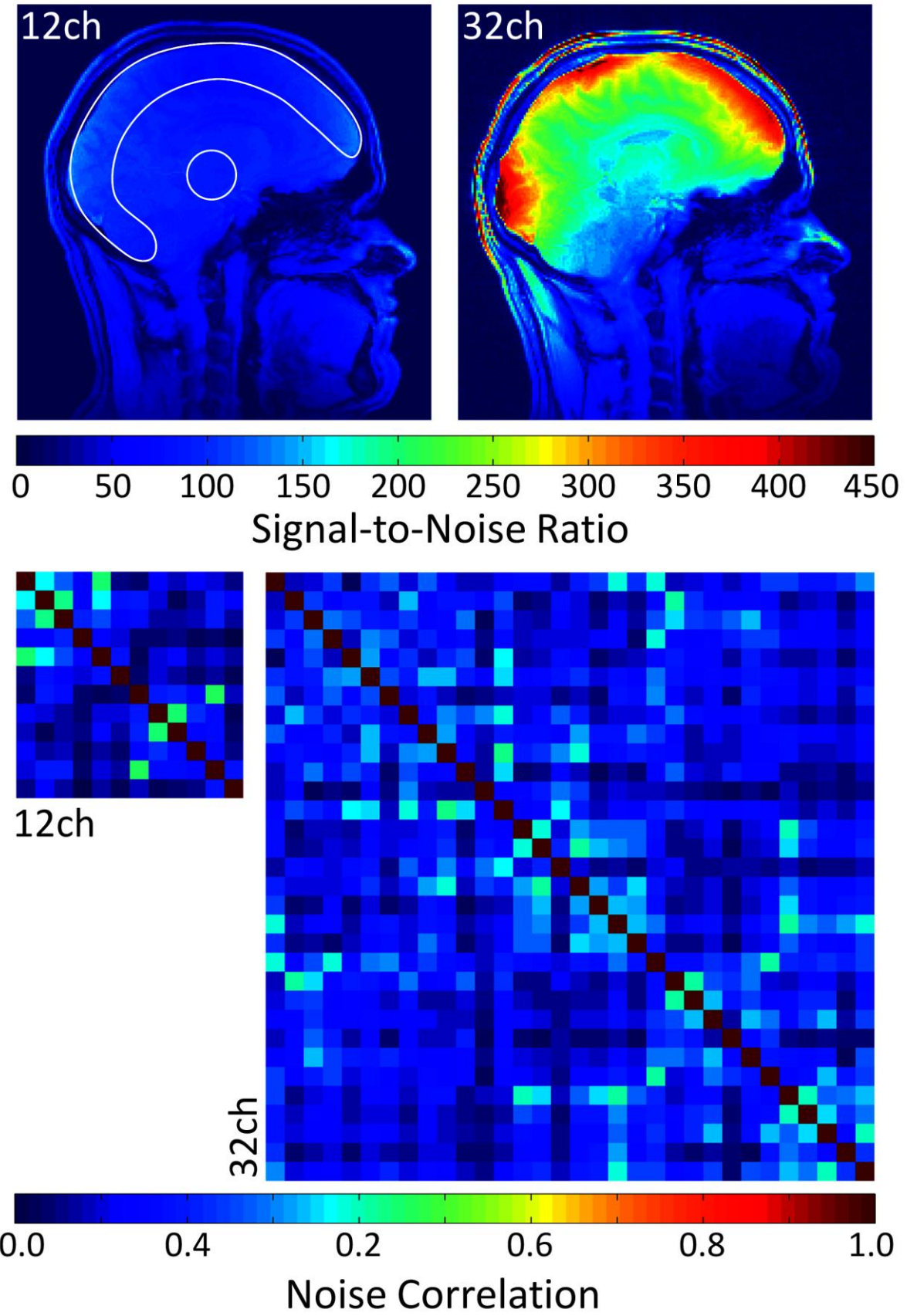


Figure 1: Pixel-wise SNR maps and noise correlation matrices from the 12Ch and 32Ch array coils. The 32Ch coil outperforms the 12Ch coil by a factor of 2.3x averaged over the given signal area. SNR from the peripheral cortex and the central brain region were obtained from the labeled Regions-of-Interest. The 32ch coil shows a 1.25-fold and 2.7-fold SNR improvement in the brain center and cortex, respectively. SNR increase in the corpus callosum region is 1.4-fold. SNR gain of the 32Ch coil can be attributed to both higher channel count and smaller helmet size. The 12Ch and 32Ch coil show average noise correlation (bottom row) of 12.3% and 9.7%, respectively.

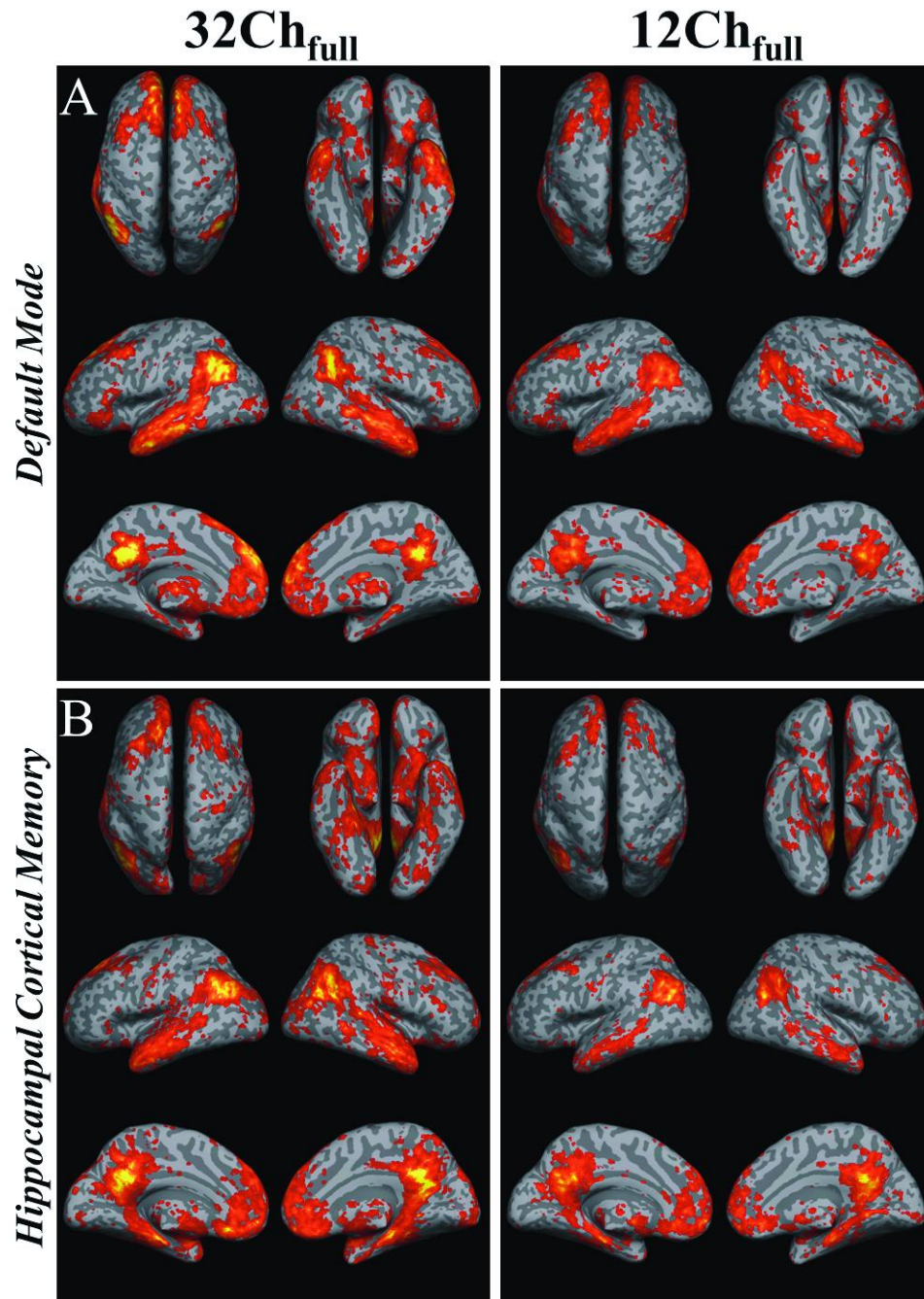


Figure 2: Statistical functional connectivity maps for the task negative networks from 32Ch and 12Ch coils (second-level analysis, $n=16$ per group; whole-brain $p_{FDR-corr} < 0.05$).

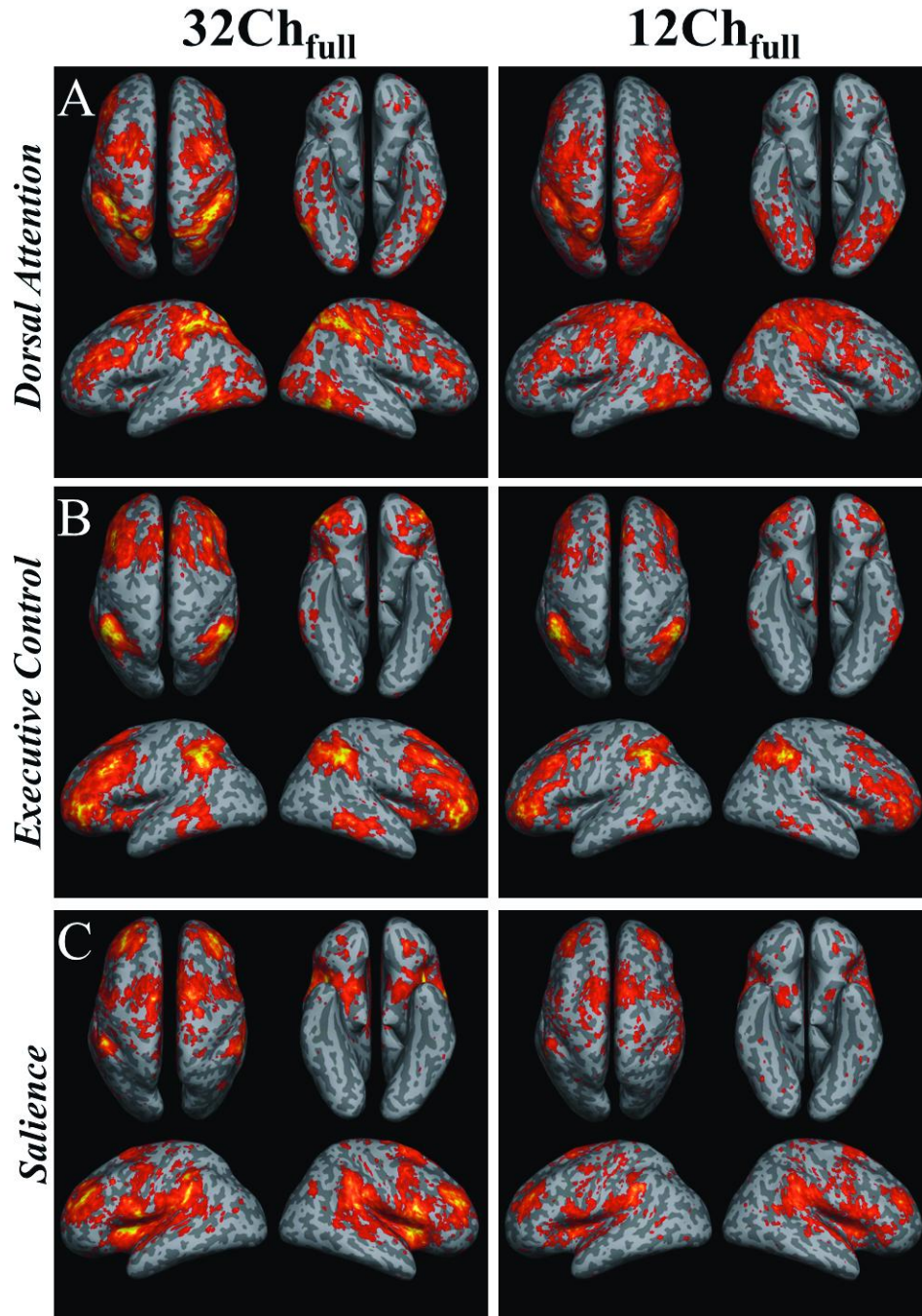


Figure 3: Statistical functional connectivity maps for the task positive networks from 32Ch and 12Ch coils (second-level analysis, $n=16$ per group; whole-brain $p_{FDR-corr}<0.05$).

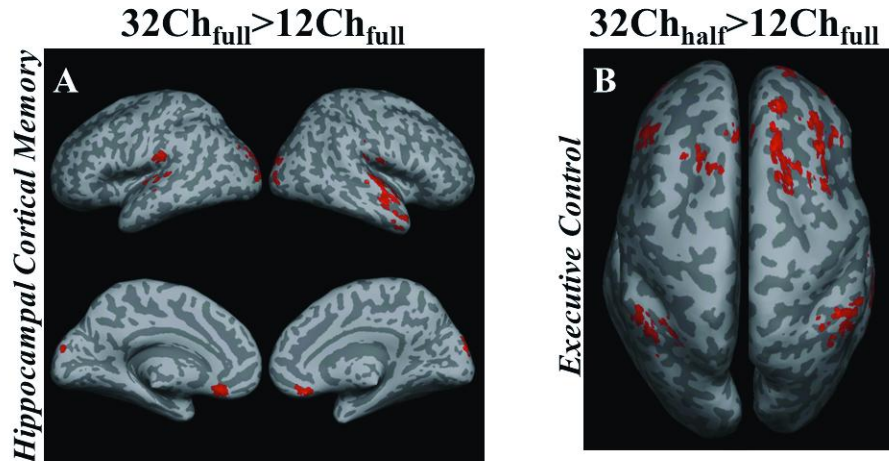


Figure 4: Representative examples from a task negative and task positive resting state network for between group comparisons from (A) $32Ch_{full} > 12Ch_{full}$ and (B) $32Ch_{half} > 12Ch_{full}$ contrasts (second-level analysis, $n=16$ per group; cluster-level $p_{FWE-corr} < 0.05$; height threshold: $T = 2.46$). (A) Connections in left and right temporal gyri and medial prefrontal cortices of the default network are revealed significantly more with 32Ch coil. (B) Entire ECN (dorsal mPFC, left and right anterior pre-frontal cortices and left and right superior parietal cortices) is revealed significantly more with half the data set from 32Ch coil.

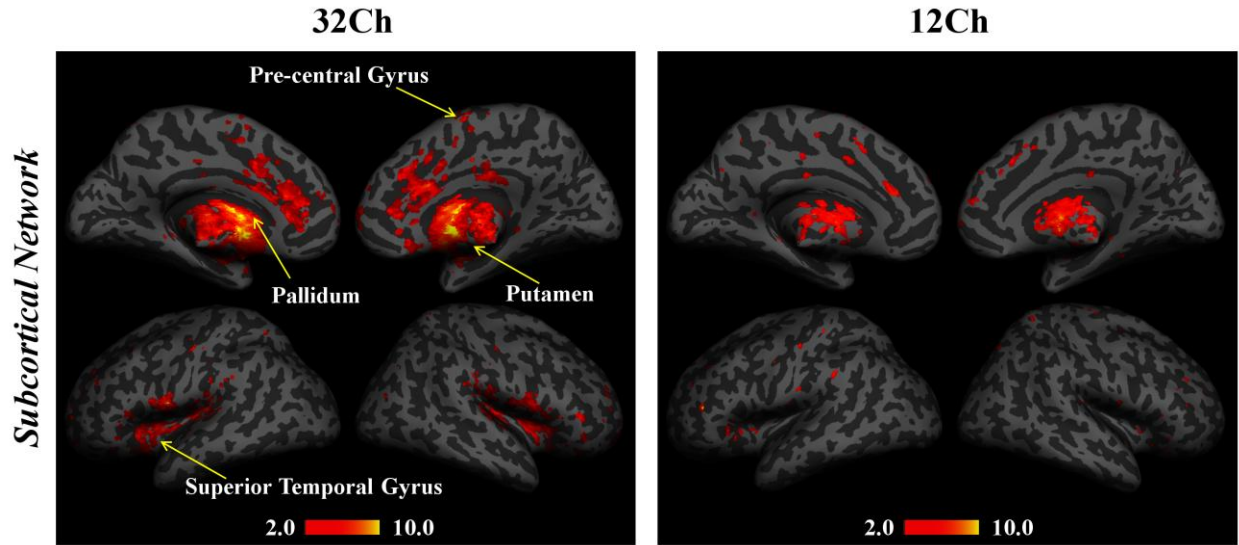


Figure 5: Statistical functional connectivity maps for the subcortical network from 32Ch and 12Ch coils (full data sets, second-level analysis, $n=16$ per group; whole-brain $p_{FDR-corr}<0.05$). Yellow arrows indicate regions that are significantly different in $32Ch_{full} > 12Ch_{full}$ contrast (whole-brain $p_{FDR-corr}<0.05$, cluster-level $p_{FWE-corr}<0.05$).

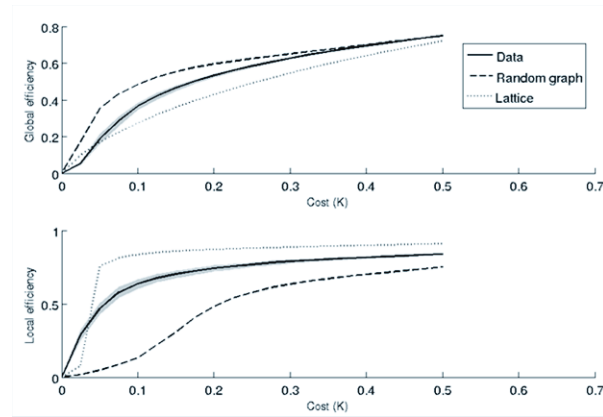


Figure 6: Global and local efficiency (y-axis) as a function of cost (x-axis) for a random graph, a regular lattice and brain networks. On average, over all subjects in both 12Ch and 32Ch groups, brain networks have efficiency curves located between the limiting cases of random and lattice topology. Solid black line represents data from all subjects for both 12Ch and 32Ch coils.

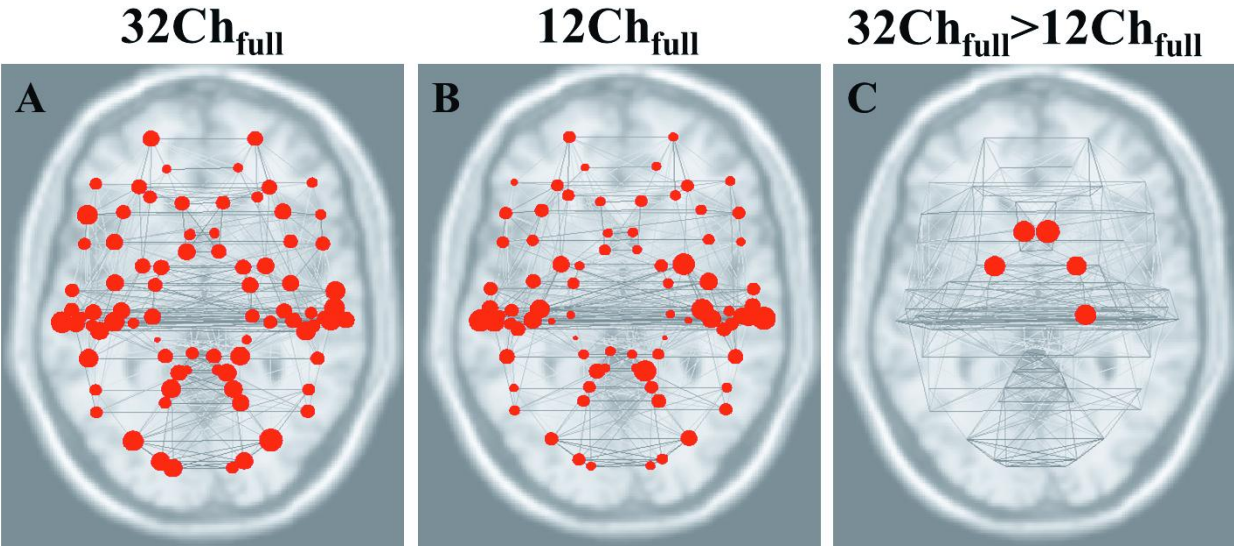


Figure 7: Graph visualization of the network-level analysis of cost for the top 15% ROI-to-ROI connectivity ($p_{FDR-corr} < 0.05$) from all Brodmann areas (number of nodes = 84,16 subjects per group). Circle-sizes represent T-values.

TABLE CAPTIONS

Table I: Peak foci of seed regions for all networks.

Table II: Positively correlated brain regions for $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ contrast (second-level group analysis, $n = 16$ per group; cluster-level $p_{\text{FWE-corr}} < 0.05$; height threshold: $T = 2.46$); opposite contrast was not significant.

Table III: Positively correlated brain regions for $32\text{Ch}_{\text{half}} > 12\text{Ch}_{\text{full}}$ contrast (second-level group analysis, $n = 16$ per group; cluster-level $p_{\text{FWE-corr}} < 0.05$; height threshold: $T = 2.46$); opposite contrast was not significant.

Table IV: Global and local efficiency comparisons of 32Ch and 12Ch coils (graph theory analysis), for the top 15% ROI-to-ROI connectivity ($p_{\text{FDR-corr}} < 0.05$), from from all Brodmann areas (number of nodes = 84, 16 subjects per group).

Table V: T-values from $32\text{Ch}_{\text{full}}$, $12\text{Ch}_{\text{full}}$, and $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ comparisons from the network level analysis of cost (depicted in Figure 7) for the top 15% ROI-to-ROI connectivity ($p_{\text{FDR-corr}} < 0.05$) from all Brodmann areas (number of nodes = 84, 16 subjects per group). Brain regions provided in column 1, correspond to the ROIs represented as circles in Figure 7C ($32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ comparison). Precisely, these are the brain regions that surpassed the threshold ($p_{\text{FDR-corr}} < 0.05$) for $32\text{Ch}_{\text{full}} > 12\text{Ch}_{\text{full}}$ contrast during the network-level analysis of cost.

Table I: Peak foci of seed regions for all networks.

Brodmann Area	x y z	Brain Region
	<i>DMN</i>	
30	0 -52 27	Posterior Cingulate Cortex (PCC)
8	-1 54 27	medial Pre-Frontal Cortex (mPFC)
39	-46 -66 30	left Lateral Parietal Cortex (LPC)
39	49 -63 33	right LPC
20	-61 -24 -9	left Inferior Temporal (IT)
20	58 -24 -9	right IT
-	0 -12 9	medial Dorsal Thalamus (mDT)
-	-25 -81 -33	left Posterior Cerebellum (PC)
-	25 -81 -33	right PC
	<i>HCMN</i>	
27	-21 -25 -14	left Hippocampal Formation (HF)
27	24 -19 -21	right HF
8	0 51 -7	ventro-medial Pre-Frontal Cortex (vmPFC)
30	1 -55 15	PCC
40	-47 -71 29	left posterior Inferior Parietal Lobule (IPL)
40	50 -64 27	right posterior IPL
	<i>DAN</i>	
6	-29 -9 54	left Frontal Eye Field (FEF)
6	29 -9 54	right FEF
7	-26 -66 48	left posterior Intraparietal sulcus (IPS)
7	26 -66 48	right posterior IPS
40	-44 -39 45	left anterior IPS
40	41 -39 45	right anterior IPS
21	-50 -66 -6	left Middle Temporal (MT)
21	53 -63 -6	right MT
	<i>ECN</i>	
8	0 24 46	dorsal mPFC
10	-44 45 0	left anterior PFC
10	44 45 0	right anterior PFC
40	-50 -51 45	left Superior Parietal Cortex (SPC)
40	50 -51 45	right SPC
	<i>SN</i>	
32	0 21 36	dorsal Anterior Cingulate Cortex (ACC)
10	-35 45, 30	left anterior PFC
10	32 45 30	right anterior PFC
13	-41 3 6	left Insula
13	41 3 6	right Insula
40	-62 -45 30	left LPC
40	62 -45 30	right LPC

Peak foci of seed regions for all networks.

Table II: Positively correlated brain regions for $32Ch_{full} > 12Ch_{full}$ contrast (second-level group analysis, $n = 16$ per group; cluster-level $p_{FWE-corr} < 0.05$; height threshold: $T = 2.46$); opposite contrast was not significant.

Brain Region	Brodmann Area	Peak cluster	Voxels per cluster	T_{max}
	DMN			
left Inferior Temporal Gyrus (ITG)	BA 20	-48 -6 -38	1016	6.35
right Superior Parietal Cortex (SPC)	BA 40	38 -50 28	161	5.32
right Secondary Visual Cortex (SVC)	BA 18	34 -76 -20	212	4.85
right ITG	BA 20	40 -2 -48	244	4.38
right Middle Temporal Gyrus (MTG)	BA 21	46 -4 -20	160	3.89
	HCMN			
left SVC	BA 18	-16 -100 18	255	6.51
right MTG	BA 21	56 -12 6	614	5.15
Orbitofrontal Cortex (OFC)	BA 11	4 24 -22	263	4.99
left MTG	BA 21	-56 -18 8	207	4.96
right SVC	BA 18	24 -98 12	223	4.34
	DAN			
right Dorso Lateral Pre-Frontal Cortex (DLPFC)	BA 9	44 26 38	212	5.26
right Fusiform Gyrus	BA 37	48 -50 -14	309	5.09
right anterior Pre-Frontal Cortex (PFC)	BA 10	24 66 -10	144	4.88
left Fusiform Gyrus	BA 37	-50 -56 -2	232	4.66
left ITG	BA 20	-58 -32 -20	151	4.45
left Superior Parietal Lobule (SPL)	BA 7	-16 -48 50	114	4.06
	ECN			
right Superior Frontal Gyrus	BA 8	40 26 38	439	8.73
left MTG	BA 21	-58 -38 -8	285	5.32
right MTG	BA 21	56 -44 0	262	4.88
left DLPFC	BA 9	-36 2 32	225	4.76
left SPC	BA 40	-52 -38 52	112	4.71
left anterior PFC	BA 10	-20 62 10	254	4.50
right anterior PFC	BA 10	36 40 2	138	4.33
Premotor Cortex	BA 6	20 14 56	128	3.72
	SN			
left Insular Cortex	BA 13	-34 6 -2	917	7.07
right Insular Cortex	BA 13	32 16 6	1243	6.15
dorsal Anterior Cingulate Cortex	BA 32	-4 26 24	920	5.51
right anterior PFC	BA 10	34 3 8	143	4.91
left DLPFC	BA 9	-26 38 20	507	4.86
right DLPFC	BA 9	32 46 36	483	4.81
left DLPFC	BA 9	-46 2 20	112	4.31
left SPC	BA 40	-58 -38 40	182	4.03

Positively correlated brain regions for 32Chfull > 12Chfull contrast (second-level group analysis, n = 16 per group; cluster-level pFWE-corr<0.05; height threshold: T = 2.46); opposite contrast was not significant.

Table III: Positively correlated brain regions for 32Ch_{half} > 12Ch_{full} contrast (second-level group analysis, n = 16 per group; cluster-level $p_{FWE-corr} < 0.05$; height threshold: T = 2.46); opposite contrast was not significant.

Brain Region	Brodmann Area	Peak cluster	Voxels per cluster	T _{max}
	DMN			
Superior Frontal Gyrus (SFG)	BA 8	14 40 44	98	4.72
Superior Parietal Lobule (SPL)	BA 7	34 -64 50	91	4.43
left Superior Temporal Gyrus	BA 22	-54 -34 -2	106	4.16
	HCMN			
Premotor Cortex	BA 6	32 -6 54	95	5.08
Orbitofrontal Cortex (OFC)	BA 11	-6 16 -24	337	4.93
SFG	BA 8	0 34 40	480	4.72
left Dorsolateral PFC (DLPFC)	BA 46	-44 32 10	102	4.56
Dorsal Anterior Cingulate Cortex	BA 32	6 34 8	153	4.48
right SPL	BA 7	42 -64 48	113	4.44
right DLPFC	BA 46	44 46 12	130	4.24
	DAN			
right DLPFC	BA 9	48 36 26	107	6.41
right anterior Pre-Frontal Cortex (PFC)	BA 10	38 56 14	210	5.12
Premotor Cortex	BA 6	60 -14 16	116	4.83
left SPL	BA 7	-24 -54 48	102	4.67
	ECN			
right Inferior Temporal Gyrus	BA 20	54 -22 -18	127	6.78
right anterior PFC	BA 10	40 26 36	789	6.76
right Insular Cortex	BA 13	52 -12 2	175	5.69
right anterior PFC	BA 10	34 52 -6	521	5.20
right Superior Parietal Cortex (SPC)	BA 40	52 -52 50	286	5.14
left SFG	BA 8	-8 28 46	245	5.10
left DLPFC	BA 9	-40 28 26	94	4.91
right Fusiform Gyrus	BA 37	54 -44 -6	93	4.90
left anterior PFC	BA 10	-4 46 8	130	4.22
left SPC	BA 40	-36 -62 50	147	4.12
	SN			
right Insular Cortex	BA 13	36 -4 -6	146	6.30
left Insular Cortex	BA 13	-32 -4 -12	215	5.82
SPL	BA 7	-2 -66 32	170	4.31

Positively correlated brain regions for 32Ch_{half} > 12Ch_{full} contrast (second-level group analysis, n = 16 per group; cluster-level $p_{FWE-corr} < 0.05$; height threshold: T = 2.46); opposite contrast was not significant.

nd-level group analysis, n = 16 per group; cluster-level $p_{FWE-corr} < 0.05$; height threshold: T = 2.46); opposite contrast was not significant.

Table IV: Global and local efficiency comparisons of 32Ch and 12Ch coils (graph theory analysis), for the top 15% ROI-to-ROI connectivity ($p_{FDR-corr} < 0.05$), from from all Brodmann areas (number of nodes = 84, 16 subjects per group).

Coil	Global Efficiency	Local Efficiency
32Ch _{full}	0.471	0.717
32Ch _{half}	0.455	0.697
12Ch _{full}	0.459	0.688
$p_{FDR}(32Ch_{full} > 12Ch_{full})$	0.056	0.034
$p_{FDR}(12Ch_{full} > 32Ch_{full})$	n.s	n.s

Global and local efficiency comparisons of 32Ch and 12Ch coils (graph theory analysis), for the top 15% ROI-to-ROI connectivity ($p_{FDR-corr} < 0.05$), from from all Brodmann areas (number of nodes = 84, 16 subjects per group).

Table V: T-values from $32Ch_{full}$, $12Ch_{full}$, and $32Ch_{full} > 12Ch_{full}$ comparisons from the network level analysis of cost (depicted in Figure 7) for the top 15% ROI-to-ROI connectivity ($p_{FDR-corr} < 0.05$) from all Brodmann areas (number of nodes = 84, 16 subjects per group). Brain regions provided in column 1, correspond to the ROIs represented as circles in Figure 7C ($32Ch_{full} > 12Ch_{full}$ comparison). Precisely, these are the brain regions that surpassed the threshold ($p_{FDR-corr} < 0.05$) for $32Ch_{full} > 12Ch_{full}$ contrast during the network-level analysis of cost.

T-values	Brain Region	T-scores			
		32Ch	12Ch	32Ch>12Ch	
	left Anterior Entorhinal Cortex	9.73	6.21	3.18	from $32Ch_{full}$, $12Ch_{full}$, and $32Ch_{full} >$ $12Ch_{full}$ comparisons
	right Anterior Entorhinal Cortex	9.68	8.82	2.99	
	right Perirhinal Cortex	8.45	5.83	3.17	
	right Anterior Cingulate Cortex (ACC)	6.01	3.31	3.58	
	left ACC	4.70	3.57	3.26	

from the network level analysis of cost (depicted in Figure 7) for the top 15% ROI-to-ROI connectivity ($p_{FDR-corr} < 0.05$) from all Brodmann areas (number of nodes = 84, 16 subjects per group). Brain regions provided in column 1, correspond to the ROIs represented as circles in Figure 7C ($32Ch_{full} > 12Ch_{full}$ comparison). Precisely, these are the brain regions that surpassed the threshold ($p_{FDR-corr} < 0.05$) for $32Ch_{full} > 12Ch_{full}$ contrast during the network-level analysis of cost.