Published in final edited form as:

Magn Reson Imaging. 2020 May ; 68: 167–172. doi:10.1016/j.mri.2020.02.008.

A 16-Channel Loop Array for *in vivo* Macaque Whole-brain Imaging at 3T

Zhiyan Quan^a, Yang Gao^{a,c}, Shuxian Qu^a, Xiaojie Wang^b, Robert M. Friedman^b, Mykyta M. Chernov^b, Christopher D. Kroenke^b, Anna Wang Roe^{a,b,c}, Xiaotong Zhang^{a,c,*}

^aInterdisciplinary Institute of Neuroscience and Technology, Key Laboratory for Biomedical Engineering of Ministry of Education, College of Biomedical Engineering & Instrument Science, Zhejiang University, Hangzhou, China

^bDivision of Neuroscience, National Primate Research Center, Oregon Health and Science University, Beaverton, OR, United States

^cSchool of Medicine, Zhejiang University, Hangzhou, China

Abstract

Non-human primates (NHPs) are vital models for neuroscience research. These animals have been widely used in behavioral, electrophysiological, molecular, and more recently, multimodal neuroimaging and neuro-engineering studies. Several RF coil arrays have been designed for functional, high-resolution brain magnetic resonance imaging (MRI), but few have been designed to accommodate multimodal devices. In the present study, a 16-channel array coil was constructed for brain imaging of macaques at 3 Tesla (3T). To construct this coil, a close-fitting helmet-shaped form was designed to host 16 coil loops for whole-brain coverage. This assembly is mountable onto stereotaxic head frame bars, and the coil functions while the monkey is in the sphinx position with a clear line of vision of stimuli presented from outside of the MRI system. In addition, 4 openings were allocated in the coil housing, allowing multimodal devices to directly access visual cortical regions such as V1-V4 and MT. Coil performance was evaluated in an anesthetized macaque by quantifying and comparing signal-to-noise ratios (SNRs), noise correlations, and g-factor maps to a vendor-supplied human pediatric coil frequently used for NHP MRI. The result from *in vivo* experiments showed that the NHP coil was well-decoupled, had higher SNRs in cortical regions, and improved data acquisition acceleration capability compared with a vendor-supplied human pediatric coil that has been frequently used in macaque MRI studies. Furthermore, whole-brain anatomic imaging, diffusion tensor imaging and functional brain imaging have also been conducted: the details of brain anatomical structure, such as cerebellum and brainstem, can be clearly visualized in T2-SPACE images; b0 SNR calculated from b0 maps was higher than the human pediatric coil in all regions of interest (ROIs); the time-course SNR (tSNR) map calculated for GRE-EPI images demonstrates that the presented coil can be used for high-resolution functional imaging at 3T.

*Corresponding author at: Interdisciplinary Institute of Neuroscience and Technology, Key Laboratory for Biomedical Engineering of Ministry of Education, College of Biomedical Engineering & Instrument Science, Zhejiang University, Hangzhou, China & School of Medicine, Zhejiang University, Hangzhou, China, zhangxiaotong@zju.edu.cn (X.T. Zhang).

Keywords

Non-human primates; 3T; RF coil; multimodal neuroimaging

1. Introduction

Non-human primates (NHPs) are useful models for neuroscience research due to similarities in brain organization with humans. Studies using NHPs have led to important advances in the medical sciences. Functional studies have revealed commonalities in cognition and their underlying circuits between NHPs and humans [1]. Non-invasive techniques, such as anatomical and functional magnetic resonance imaging (MRI) with high spatial and temporal resolutions and excellent soft-tissue contrast have proven indispensable for studying functional organization and connectivity in NHP brains [2–6]. The integration of multiple methods, such as optical and electrophysiological recording methods, and optogenetic, ultrasound, and infrared neural stimulation methods, with MRI expands the possible range of spatial/temporal scales and size scales [7–13].

There are significant challenges in achieving high spatial and temporal resolution imaging in NHP brains. For the small head size of NHPs, such as macaques, a higher resolution is required to detect the same structural and functional details as in the human brain. The goal of increasing signal-to-noise ratios (SNRs) can be achieved with use of high (> 1.5T) magnetic field strengths. In addition, in lieu of single-channel coils which suffer from distortions and poor signal sensitivity, incorporation receiver coil arrays can provide higher quality, distortion-free results. Another design consideration for studies of the visual system is that having multi-element coils may lead to geometries that obstruct the field of view of the subject and prevent the presentation of visual stimuli. This can be overcome by careful coil array designs that allows for positioning and animal's head in a manner compatible with presentation of visual stimuli. Recently, a number of radiofrequency (RF) coil arrays have been designed for NHP imaging [14–23], for example, an 8-channel receive-only array coil implanted on the skull of a macaque was constructed to achieve 5.4 fold increasing in SNR when compared with a single-channel coil placed externally around the animal's head [17]; however, few have been compatible with multimodal imaging/stimulation approaches.

In the present study, a 16-channel coil array was constructed for macaque monkeys in the MRI at 3 Tesla (3T). Its performance was evaluated over an anesthetized macaque and compared with a vendor-supplied 16-channel human pediatric coil that has been frequently used in macaque MRI studies [5].

2. Methods

2.1. 16-channel Receive-only Coil Design

The 3D drawing of the coil mounted over a stereotactic apparatus and the circuit schematic of the coil element are illustrated in Fig. 1. The macaque is placed in a sphinx position for scanning (Fig. 1a). The coil was constructed with sixteen 2–5 cm diameter receive-only loops tuned to 123.23 MHz in an overlapping array configuration (Fig. 1c). Decoupling

between coil elements was achieved by critical overlapping and preamplifier decoupling [24], and each coil element contained active detuning circuitry, formed by PIN diode D , variable inductor L_D and a fixed-value capacitor C_1 , to detune the coil during RF transmission. Sixteen low-impedance preamplifiers (WanTcom Inc., Chanhassen, MN, USA) were placed roughly a quarter-wavelength distance away from each corresponding coil element, and lattice balance-unbalance converters were placed between each connected coil element and preamplifier to minimize common-mode currents on the cable (Fig. 1d).

The coil was constructed on a 3D-printed form made from polycarbonate using Fused Deposition Modelling (FDM) (Virtumed LLC, Minneapolis, MN, USA). A fairly large macaque (large, 10.3 kg) that was available during the time period of coil design received an MPRAGE scan, and the proposed coil housing was designed in a helmet-like shape to conform to the selected macaque's head (approximating a sphere of 11 cm diameter in shape). Four 0.75-inch Loc-Line modular hoses were installed on top of the helmet for coil mounting to a stereotaxic head frame, flexible positioning, and stabilization (Fig. 1a&b). Two removable receive loops (~5 cm in diameter) near the animal's ears were designed with two QMA connector pairs soldered on each loop, so half of each ear loop can be plugged off/on for convenient installation of ear-bars and animal handling (Fig. 1b&d). In addition, since our research interest mainly focuses on functional organizations and their connectivity in the visual cortex through multimodal technology, four openings (3.5 cm in diameter, see Fig. 1e) were placed in the housing to accommodate multimodal apparatus, allowing direct access to visual cortical regions such as V1-V4 and MT. Two 3D model files have also been included as supplementary materials for a better demonstration of the general arrangement of all 16 loops and the coil former.

2.2. Animal Preparation

One anesthetized macaque (male, 9.6 kg) was used for the imaging studies. All procedures were in accordance with NIH standards and approved by the local Institutional Animal Care Committee. Anesthesia was initiated with sufentanil (3 $\mu\text{g/kg}$), and gradually transitioned with isoflurane (1.4% – 2%) through an endotracheal tube. Anesthesia was maintained on sufentanil (2 $\mu\text{g/kg}\cdot\text{h}$) intravenously (IV), and an end-tidal volume of 0.3% isoflurane was given via inhalation. The body temperature was maintained at a constant temperature using a recirculating water heater, and heart rate, SpO_2 , and CO_2 were monitored continuously.

2.3. MR Data Acquisition

All measurements were performed on a 3T clinical scanner (Prisma, Siemens, Erlangen, Germany) equipped with a whole-body gradient set (80 mT/m and 200 T/m/s). The SNR, noise correlation, and g-factor were obtained from proton-density-weighted FLASH images: $\text{TR/TE}/\alpha = 10 \text{ ms}/4.8 \text{ ms}/15^\circ$, slice thickness = 5 mm, matrix size = 128×128 , field-of-view (FOV) = $128 \times 128 \text{ mm}^2$, bandwidth (BW) = 390 Hz/pixel. Noise covariance information was acquired using the same pulse sequence, but with no RF excitation. The image noise was magnified by a factor of $g\sqrt{R}$ (g is the g-factor, and R is the acceleration factor) under parallel imaging acceleration.

In addition, T2-SPACE (SE-based) images (TR/TE = 3200 ms/385 ms, matrix size 320×320 , FOV = $160 \times 160 \text{ mm}^2$, slice thickness = 0.5 mm, turbo factor = 218, acceleration rate = 2, BW = 473 Hz/pixel) were acquired in 9'54" to further show the performance of the 16-channel receive-only coil. High-resolution diffusion tensor images (six $b = 0$ images and thirty $b = 1000 \text{ s/mm}^2$) were acquired in 28'30" (TR/TE = 6700 ms/73 ms, matrix size = 118×118 , FOV = $128 \times 128 \text{ mm}^2$, acceleration rate = 2, echo spacing = 1 ms, 56 slices, BW = 1570 Hz/pixel, EPI factor = 118) to calculate the fractional anisotropy (FA). Finally, single-shot GRE-EPI was acquired to evaluate tSNR of the presented coil in NHP fMRI imaging: TR/TE/ α = 2000 ms/30 ms/90°, matrix size = 64×64 , FOV = $128 \times 128 \text{ mm}^2$, slice thickness = 2 mm, acceleration rate = 2, echo spacing = 0.68 ms, 24 slices, BW = 1906 Hz/pixel, 100 measurements.

The same imaging parameters were employed when using the vendor-supplied 16-channel human pediatric coil designed for imaging similarly-sized objects (<https://www.siemens-healthineers.com/en-us/magnetic-resonance-imaging/options-and-upgrades/coils/pediatric-16>). To utilize this coil, it is necessary to place the macaque in a supine position.

2.4. Data Analysis

The calculation of noise correlation matrix, SNR and g-factor was completed online using IceMGHCoilArrayUntil software (Athinoula A. Martinos Center for Biomedical Imaging, MGH Department of Radiology, Harvard Medical School Charlestown, MA, USA). Noise correlation matrix was calculated by following equation:

$$r_{i,j} = \frac{(n_i - a_i)(n_j - a_j)^H}{\sqrt{(n_i - a_i)(n_i - a_i)^H} \sqrt{(n_j - a_j)(n_j - a_j)^H}} = \frac{n_i n_j^H - a_i a_j^* N_{samp}}{\sigma_i \sigma_j} \# (1.)$$

where $r_{i,j}$ denotes the element of the i th row and j th column of covariance coefficient matrix $\mathbf{R}$, and σ_i and σ_j terms are the square root of the noise variances. $1 \times N_{samp}$ (N_{samp} , noise samples) row vector n_i represents the noise signal in coil i , similarly for n_j , and H denotes the Hermitian transpose. a_i is the mean of noise signal n_i of coil i given by

$$a_i = \frac{1}{N_{samp} - 1} \sum_{k=1}^{N_{samp}} n_i[k] \# (2.)$$

and a_j is the mean of noise signal n_j of coil j given by

$$a_j = \frac{1}{N_{samp} - 1} \sum_{k=1}^{N_{samp}} n_j[k] \# (3.)$$

where k is k th element of coil i or j . Finally, a_j^* denotes the complex conjugate of a_j .

The optimal SNR was used to calculate SNR map by given formula:

$$SNR = \sqrt{s^H \Psi^{-1} s} \# (4.)$$

where s denotes the spatial distributions of coil sensitivity and Ψ is the noise covariance matrix, given by

$$\varphi_{i,j} = \frac{1}{N_{smp} - 1} (n_i n_j^H - a_i a_j^* N_{smp}) \# (5)$$

where $\varphi_{i,j}$ denotes the element of the i th row and j th column of noise covariance matrix Ψ .

The g-factor maps were computed based on the differences between the noise with and without acceleration to assess noise amplification in sensitivity encoding (SENSE) reconstructions of the accelerated images [25]. For a fair assessment of the g-factor, the FOV along the phase-encoded direction was tightly fit to the head size to maximize the aliasing in the accelerated image acquisition [26]. The phase-encoding direction was selected as anterior-to-posterior in the sagittal orientation, foot-to-head in the coronal orientation, and right-to-left in the coronal orientation, relative to the MRI bore when the macaque is in the sphinx position. The corresponding g-factor map was found using following formula:

$$g_p = \sqrt{\left[(s^H \Psi^{-1} s)^{-1} \right]_{p,p} \left[(s^H \Psi^{-1} s) \right]_{p,p}} \# (6)$$

where g_p denotes the g-factor corresponding the pixel p .

The quality of DTI images was evaluated by calculating SNR in averaged DTI datasets with $b = 0$ through using MATLAB (The Mathworks, Inc. Natick, MA, USA). The SNR was defined as the ratio of the mean MR image signal intensity of the frontal lobe, thalamus, and occipital lobe to the standard deviation of the signal intensity in background ROI outside the brain. All the ROIs were drawn manually on FA maps and copied among all DTI images of the macaque, similarly to [27]. Also, to quantify the temporal stability of the GRE-EPI images, the time-course SNR (tSNR) was calculated for each voxel as the mean value across 100 time points divided by the temporal standard deviation through using MATLAB.

3. Results

Fig. 2 shows SNR map comparisons between the pediatric coil and custom-built 16-channel macaque coil. The custom-built macaque coil had an approximately 2-fold SNR improvement in cortical regions and a slightly lower SNR in the deep brain areas.

The element decoupling in each of the two coils was verified *in vivo* by measuring the noise correlation coefficient matrices (see Fig. 3). The macaque coil showed an average value of 13.2% (ranging from 1.4% to 39.7%) comparable to the human pediatric coil that had an average value of 11.2% (ranging from 0.8% to 33.9%).

An important property of the macaque coil is its improved ability to provide parallel imaging due to a higher loop density and shorter loop-to-subject distance [28]. Fig. 4 shows a comparison of the inverse g-factor maps in three orthogonal phases encoding different directions (i.e., anterior-posterior, foot-head, and right-left) under accelerations of 2, 3, and 4, respectively. Compared with the human pediatric coil, the $1/g$ of the macaque coil was much higher in all cases, and at high acceleration rates, a $1/g$ of 0.8 (GRAPPA 3) and 0.7 (GRAPPA 4) could be maintained in most regions of the brain.

Fig. 5 shows whole-brain T2-SPACE images at an isotropic resolution of 0.5 mm, acquired with the macaque coil. These images reveal that the details of the cerebellum and brainstem can be clearly visualized.

Fig. 6 shows fractional anisotropy (FA) maps of coronal slices, calculated from 1mm-isotropic-resolution diffusion tensor images, acquired with the macaque coil. Three different regions on three single slices (enclosed in white rectangles), including the frontal lobe, thalamus, and occipital lobe, were selected for the b_0 SNR calculations. Due to the combined higher sensitivity of the macaque coil and improved parallel imaging capabilities, and the macaque coil showed higher SNR in all region of interest positions. As an example, in the frontal lobe, the SNR was 23.8 for the macaque coil and 12.6 for the human pediatric coil demonstrating an almost 2-fold increase in the SNR of the macaque coil.

The tSNR maps of single-shot GRE-EPI for two coils are shown in Fig. 7. With an acceleration rate of 2 and at 2mm isotropic resolution, the macaque coil provides an overall elevated tSNR within the entire brain and distinct tSNR improvement in cortical regions, suggesting that the macaque coil allows for high-resolution EPI acquisition at 3T.

4. Discussion

In this study, a close-fitting helmet-shaped 16-channel macaque head array coil was constructed specifically for macaque brain neuroimaging at 3T. It produced improved receive sensitivity and acceleration performance, compared with a vendor-supplied 16-channel human pediatric coil commonly used for macaque imaging. Using the macaque coil, as illustrated in Fig. 2, apparent SNR gains in cortical is attributable to the small size of the surface coil elements and a shorter coil-to-subject distance provided by close-fitting helmet-shape. However, compared with the pediatric coil, the SNR was not improved in deep brain images due to less head coverage. Nevertheless, the helmet design of the macaque coil exhibited improved acceleration performance and less SNR distortions in parallel imaging, and, as shown in Fig. 3, offered a stronger spatial modulation of the signal intensity and thus, an “upgraded” ability to unalias folded images (SENSE methods) [25] or synthesize spatial harmonics (SMASH or GRAPPA methods) [29].

Compared with the human pediatric coil, the macaque array shows improved parallel imaging performance, and thus, could effectively facilitate microscopic structural and functional MRI *in vivo* within a safe anesthetic period. In addition, four 0.75-inch Loc-Line modular hoses offered great flexibility and rigid stabilization in coil mounting to a stereotaxic device, while two removable ear loops facilitated the installation of ear-bars for

animal handling, which are all important considerations in macaque MRI studies. In contrast to the human pediatric coil via which the macaque must be placed in a supine position, the proposed macaque coil enables images to be acquired while the animal is in a sphinx or prone position, with a clear line of vision for presentation of visual stimuli in front of the animal.

The openings placed in the housing were designed to accommodate multimodal devices. This provides an opportunity to combine high-field functional MRI with optical and electrophysiological recordings of neural activity and molecular neuroimaging [10], as well as introduction of devices for optogenetic [30,31] and focal infrared neural stimulation [13]. This makes possible the incorporation of multiple modalities for the study of brain networks *in vivo* within individual subjects, an approach which may lead to qualitatively new understanding of brain function.

5. Conclusions

In this study, we constructed a close-fitting helmet-shaped, 16-channel array for macaque brain 3T MRI. Through *in vivo* experimentation using an anesthetized macaque, we evaluated the performance of this macaque 16-array coil compared with a vendor-supplied 16-channel human pediatric coil that has been routinely used for macaque MRI. The proposed coil showed improved SNR parallel imaging performance compared with the pediatric coil. Moreover, the large flexibility and rigid stabilization of the coil mounting procedure, as well as the design of the removable ear loops, make animal handling easier and provide greater convenience and efficiency when using human MR scanners and with visual stimuli presented in front of the animal. Furthermore, four openings placed in the housing were designed to incorporate other brain recording and stimulation devices for multimodal neuroimaging, a capability important for advancing our understanding of networks in the primate brain.

Acknowledgments

The authors would like to thank the following people for assisting with this work: Michael Reusz, Zheng Liu, and Jialu Zhang for helpful discussions and technical support. Funding support was provided by the National Key R&D Program of China (2018YFA0701400), National Natural Science Foundation of China (81701774, 61771423, 31627802 and 81430010), Zhejiang Lab (2018EB0ZX01), Science and Technology Program of Guangdong (2018B030333001) and Oregon Health & Science University Institutional funds.

References

- [1]. Chudasama Y, Robbins TW. Functions of frontostriatal systems in cognition: Comparative neuropsychopharmacological studies in rats, monkeys and humans. *Biological Psychology* 2006;73:19–38 [PubMed: 16546312]
- [2]. Stefanacci L, Reber P, Costanza J, Wong E, Buxton R, Zola S, et al. fMRI of Monkey Visual Cortex. *Neuron* 1998;20:1051–7 [PubMed: 9655492]
- [3]. Logothetis NK, Guggenberger H, Peled S, Pauls J. Functional imaging of the monkey brain. *Nat. Neurosci.* 1999;2:555–62 [PubMed: 10448221]
- [4]. Bell AH, Hadj-Bouziane F, Frihauf JB, Tootell RBH, Ungerleider LG. Object Representations in the Temporal Cortex of Monkeys and Humans as Revealed by Functional Magnetic Resonance Imaging. *Journal of Neurophysiology* 2009;101:688–700 [PubMed: 19052111]
- [5]. Kroenke CD. *MRI of the Nonhuman Primate Brain*. Elsevier 2010;50(3):123–4

- [6]. Arcaro MJ, Pinsk MA, Li X, Kastner S. Visuotopic Organization of Macaque Posterior Parietal Cortex: A Functional Magnetic Resonance Imaging Study. *Journal of Neuroscience* 2011;31:2064–78 [PubMed: 21307244]
- [7]. Logothetis NK, Pauls J, Augath M, Trinath T, Oeltermann A. Neurophysiological investigation of the basis of the fMRI signal. *Nature* 2001;412:150–7 [PubMed: 11449264]
- [8]. Chernov M, Roe AW. Infrared neural stimulation: a new stimulation tool for central nervous system applications. *Neurophoton* 2014;1:011011
- [9]. Roe AW, Chernov MM, Friedman RM, Chen G. In Vivo Mapping of Cortical Columnar Networks in the Monkey with Focal Electrical and Optical Stimulation. *Front. Neuroanat.* 2015;9:135 [PubMed: 26635539]
- [10]. Albers F, Wachsmuth L, van Alst TM and Faber C. Multimodal Functional Neuroimaging by Simultaneous BOLD fMRI and Fiber-Optic Calcium Recordings and Optogenetic Control. *Mol. Imaging Biol.* 2018;20:171–82 [PubMed: 29027094]
- [11]. Deffieux T, Demene C, Pernot M, Tanter M. Functional ultrasound neuroimaging: a review of the preclinical and clinical state of the art. *Current Opinion in Neurobiology* 2018;50:128–35 [PubMed: 29477979]
- [12]. Leong ATL, Dong CM, Gao PP, Chan RW, To A, Sanes DH, et al. Optogenetic auditory fMRI reveals the effects of visual cortical inputs on auditory midbrain response. *Sci. Rep.* 2018;8:8736 [PubMed: 29880842]
- [13]. Xu AG, Qian M, Tian F, Xu B, Friedman RM, Wang J, et al. Focal infrared neural stimulation with high-field functional MRI: A rapid way to map mesoscale brain connectomes. *Sci. Adv.* 2019;5:eaau7046
- [14]. Khachaturian MH. A 4-channel 3 Tesla phased array receive coil for awake rhesus monkey fMRI and diffusion MRI experiments. *JBSE* 2010;03:1085–92
- [15]. Papoti D, Yen CCC, Mackel JB, Merkle H, and Silva AC. An embedded four-channel receive-only RF coil array for fMRI experiments of the somatosensory pathway in conscious awake marmosets. *NMR in Biomedicine* 2013;26(11):1395–1402. [PubMed: 23696219]
- [16]. Goense J, Logothetis NK, Merkle H. Flexible, phase-matched, linear receive arrays for high-field MRI in monkeys. *Magnetic resonance imaging* 2010, 28(8):1183–1191. [PubMed: 20456890]
- [17]. Janssens T, Keil B, Farivar R, McNab JA, Polimeni JR, Gerits A, et al. An implanted 8-channel array coil for high-resolution macaque MRI at 3T. *NeuroImage* 2012;62:1529–36 [PubMed: 22609793]
- [18]. Janssens T, Keil B, Serano P, Mareyam A, McNab JA, Wald LL, et al. A 22-channel receive array with Helmholtz transmit coil for anesthetized macaque MRI at 3 T: A 22-channel monkey array coil for 3T MRI. *NMR Biomed.* 2013;26:1431–40 [PubMed: 23703859]
- [19]. Roopnariane CA, Ryu Y, Tofighi M, Miller PA, Oh S, Wang J, et al. Quadrature RF coil for in Vivo brain MRI of a macaque monkey in a stereotaxic head frame. *Concepts Magn. Reson.* 2012;41B:22–7
- [20]. Gilbert KM, Gati JS, Barker K, Everling S, Menon RS. Optimized parallel transmit and receive radiofrequency coil for ultrahigh-field MRI of monkeys. *NeuroImage* 2016;125:153–161. [PubMed: 26497267]
- [21]. Gilbert KM, DiPrimio MR, Hughes S, Manning KY, Menon RS. Whole-brain and local receive arrays for imaging non-human primates. 21th Annu. Meeting (ISMRM) 2013:2776.
- [22]. Adrian G, Harel N, Yacoub E, Moeller S, Ghose G, Ugurbil K. A 21 channel Transceiver Array for Non-human Primate Applications at 7 tesla Proceedings of the 18th Annual Meeting of ISMRM. 2010:1490.
- [23]. Logothetis NK, Merkle H, Augath M, Trinath T, Ugurbil K. Ultra High-Resolution fMRI in Monkeys with Implanted RF Coils. *Neuron* 2002;35(2):227–242. [PubMed: 12160742]
- [24]. Roemer PB, Edelstein WA, Hayes CE, Souza SP, Mueller OM. The NMR phased array. *Magn. Reson. Med.* 1990;16:192–225 [PubMed: 2266841]
- [25]. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: Sensitivity encoding for fast MRI. *Magn. Reson. Med.* 1999;42:952–62 [PubMed: 10542355]

- [26]. Adriany G, Van de Moortele PF, Wiesinger F, Moeller S, Strupp JP, Andersen P, et al. Transmit and receive transmission line arrays for 7 Tesla parallel imaging. *Magn. Reson. Med.* 2005;53:434–45 [PubMed: 15678527]
- [27]. Wang P, Wang D, Zhang J, Bai R, Qian M, Sun Y, et al. Evaluation of Submillimeter Diffusion Imaging of the Macaque Brain Using Readout-Segmented EPI at 7T. *IEEE Transactions on Biomedical Engineering* 2019;66(10):2945–51 [PubMed: 30762524]
- [28]. Gao Y, Chen W, Zhang X. Investigating the Influence of Spatial Constraints on Ultimate Receive Coil Performance for Monkey Brain MRI at 7 T. *IEEE Trans. Med. Imaging* 2018;37:1723–32 [PubMed: 29969422]
- [29]. Griswold MA, Jakob PM, Heidemann RM, Nittka M, Jellus V, Wang J, et al. Generalized autocalibrating partially parallel acquisitions (GRAPPA). *Magn. Reson. Med.* 2002;47:1202–10 [PubMed: 12111967]
- [30]. Gerits A, Farivar R, Rosen BR, Wald LL, Boyden ES, Vanduffel W. Optogenetically Induced Behavioral and Functional Network Changes in Primates. *Current Biology* 2012;22:1722–6 [PubMed: 22840516]
- [31]. Chernov MM, Friedman RM, Chen G, Stoner GR, Roe AW. Functionally specific optogenetic modulation in primate visual cortex. *PNAS USA* 2018;115:10505–10 [PubMed: 30257948]

Fig. 1.

The integrated design of the custom-built macaque coil. (a) Diagram of macaque placement in a sphinx position. (b) Schematic diagram of the coil design: four 0.75-inch Loc-Line modular hoses were on the top of the coil former, two removable ear loops were near the animal's ears and four openings were placed in the housing, and four clamps were used for fixing the coil to the stereotaxic frame. (c) The layout of the loop elements within the housing. (d) Circuit schematic of the coil element: each element was tuned to 123.23 MHz and matched to $50\ \Omega$ by adjusting tuning capacitor C_T and matching capacitor C_M ; PIN Diode D , variable inductor L_D and a fixed-value capacitor C_1 act as the active detuning trap; preamplifier decoupling was fine-tuned by choosing a proper capacitor value of C_{pd} ; the two removable receive loops near the animal's ears were designed with two QMA connector pairs soldered on each loop, so half of each ear loop can be conveniently plugged off/on for easy installation of ear-bars during animal handling. (e) Locations of the 4 openings in the coil housing.

16CH Pediatric Coil 16CH Macaque Coil

Fig. 2.

A comparison of the signal-to-noise ratios (SNRs) between a vendor-supplied human pediatric coil (left) and the custom-built macaque coil (right). a.u.: arbitrary unit.

16CH Pediatric Coil 16CH Macaque Coil

Fig. 3. Noise correlation matrices. The average correlation value for the 16 coils of the macaque coil was 13.2% (right, range 1.4% to 39.7%) and was 11.2% for the human pediatric coil (left, range 0.8% to 33.9%).

Fig. 4.
The inverse g-factor maps of the human pediatric and macaque coils in three orthogonal phase-encoding acceleration directions of 2, 3, and 4. AP: anterior-posterior. FT: foot-head. RL: right-left.

Fig. 5. Axial, sagittal, and coronal views of whole-brain T2-SPACE anatomical images at an isotropic resolution of 0.5 mm when the macaque coil was used.

Fig. 6. Fractional anisotropy (FA) maps showing a comparison of coronal slices for both the 16-channel human pediatric coil and the custom-built macaque coil, calculated from high-resolution diffusion tensor images. Three different regions (enclosed in white rectangles), including the frontal lobe, thalamus, and occipital lobe, were selected to calculate the b0 SNR.

Fig. 7. tSNR of single-shot GRE-EPI on three orthogonal planes (i.e. axial, coronal and sagittal views) with an acceleration rate of 2 and 2 mm isotropic resolution.