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A 20-channel coil for improved magnetic resonance imaging of the optic nerve

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Introduction

Optic neuritis is characterized by inflammation of the optic nerve, which causes a loss of vision due to swelling and destruction of the myelin sheath covering the optic nerve (1). The most common etiology is Multiple Sclerosis (MS). Nearly 50% of patients with MS will develop optic neuritis and 20-30% of the time optic neuritis is the presenting sign of MS (2). Early diagnosis can decrease the risk of developing negative symptoms from optic neuritis (3, 4). In MRI the presence of lesions, which indicate demyelization of the white matter, is the strongest predictor for clinically definite MS. Other methods of determining optic neuritis include a fundoscopic exam, which in some cases does not give an adequate diagnosis.

MRI of the optic nerve, which is usually performed with a standard head coil such as the clinical 12-channel coil available on the Siemens TIM Trio 3T MRI scanner, is subject to several limitations such as artifacts from eye movement (5). Many limitations could be overcome using a coil built specifically to image the optic nerves (6-8). For example, increased SNR obtained using a dedicated optic nerve coil would enable reduced field-of-view (FOV) DTI images as well as faster scans through parallel imaging techniques to minimize patient movement artifacts.

Phased array receiver coil designs have been increasingly applied in MRI (9-13). Phased arrays are advantageous since multiple channels can provide an increase in signal sensitivity and imaging acceleration (14). A single surface coil typically has sufficient signal to noise ratio (rSNR) at the surface of a sample; however, at greater depths the SNR can suffer greatly compared to a traditional volume coil. Use of a phased array coil can compensate at the greater imaging depths and, if utilized correctly, can provide similar or greater SNR values compared to a volume coil (15). It has also been discovered that one of the factors that improves accelerated imaging is increasing the number of coils surrounding a volume of interest. This is due to the increased number of different coil sensitivity profiles. The increased parallel imaging performance can aid in the use of multiple parallel imaging techniques (16, 17). Furthermore, in addition to providing increased resolution and/or decreasing imaging time, the increase in SNR can enable diffusion tensor imaging (DTI) (18, 19), enhance temperature measurements (20), and fundamentally improve clinical diagnosis and patient treatment planning.

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In this study, an RF phased array coil was built to image the optic nerve from the orbits to the optic chiasm for the purposes of visualizing optic neuritis that typically manifests in Multiple Sclerosis patients. As opposed to previous designs, which are limited by the number of coils or generalized to imaging all of the head (9-11) we present a dedicated 20-channel coil for imaging of the optic nerve. This coil achieves improved SNR for higher resolution, faster imaging times and useful DTI performance.

Methods

Coil Construction

The optic nerve coil was constructed on a fiberglass former that was modeled after an average sized human head and fit over the head as a mask (see Fig. 1). This former was made to minimize the distance between the coil and the anatomy of interest. By doing this the higher signal sensitivity is located nearer to the target anatomy. Because the surface of interest for the former could be more simply described by a flat plane wrapped around the front of the head, a hexagonal pattern was used to layout the coils along the position of the optic nerve. One coil element was placed over each eye and a cluster of 7 loops was placed over the each side of the head. Four additional elements were placed between the eyes and over the forehead for a total of 20 coil elements. Critical overlap was maintained between all neighboring coils. The array layout of the coils was designed to achieve the best SNR at the optic nerve. The placement of the coil on the head that is closest to the optic nerve was found to be at the junction that is inferior to the temple and lateral to each eye. For several human models the distance from that junction to the optic nerve was found to be 3-5 cm. To get the best SNR at the location of the optic nerve posterior to the eye, three of the loops were overlapped at the inferior-temple junction on each side of the head, with two of the coils' overlap region lying parallel along the optic nerve. This was motivated by an increase in SNR at the transverse overlap of two coils (21). Based around these three primary coils on each side of the head, the rest of the array was designed and constructed using critical overlap positioning of coil elements to minimize mutual inductance. Also an additional four coils were placed above the eye coils on the forehead to achieve more SNR in the front portion of the optic nerve. The size of the coils was determined to be large enough for sufficient sample noise dominance compared to corresponding coil noise (22), and small enough to be overlapped along the optic nerve. This phased array construction technique has been shown to achieve the benefits of high sensitivity from the small individual coils and benefits of larger loops from the synthesis of larger loops from the smaller loops in the phased array reconstruction algorithms (14). The majority of the coils were circular with dimensions of 5.75 cm. $\times$ 5.75 cm. The eye elements were ovals with a dimension of 6.35 cm. $\times$ 7 cm. These two elements were mounted in such a way that they extended over the subject's eyes while allowing the subject to comfortably see out through the coils. A forehead coil with dimensions of 6.35 cm. $\times$ 8.25 cm. was placed between the two eye coils and three coils of dimensions 7 cm. $\times$ 7 cm. were placed over the forehead coil. These additional coils were slightly larger to fill the surface area of the fiberglass mask and to provide increased depth of penetration from that surface of the head. The surface coils (see Fig. 2) were etched from copper-clad kapton material (DuPont Pyralux, Durham, NC, USA). The conductor width was 5 mm. Gaps were etched through the copper coating of each coil to allow the insertion of four capacitors for each loop and the copper traces were narrowed to 3 mm at the point of overlap with adjacent coils to reduce capacitive coupling between loops (22). The coils were attached to the input of the preamp board with 24 cm of RG-316 cable (Thermal-wire, Naples, FL, USA). This small length of cable was used to reduce the electrical length of the outer shield in order to minimize cable-coupling problems that might otherwise arise from interactions with the body coil and with cables from adjacent channels.

The 20 low-input impedance preamps (Siemens Medical Solutions, Erlangen, Germany) were incorporated into a preamp board. These preamps are used to decouple the non-nearest neighbors due to the low input impedance. Transferring the short at the preamp to the matching circuit at the coil terminals causes a parallel resonance to occur at the coil, which limits current flow in the loop during the receive portion of the imaging sequence (23). The preamp board also contained a cable trap consisting of a 27-pF high voltage capacitor that bridged the outer conductor of a hand-wound four-turn inductor made of semi rigid coaxial cable (UT-070, Microstock, USA). A pi-network phase shifter was used to adjust the electrical length of the transmission line needed for the preamp decoupling. A DC bias line for controlling the PIN diode of the active detuning circuit is also incorporated into the receiver circuit board. The preamp outputs were connected by coaxial cables to plugs that fit into the sockets provided in the scanner bed near the patient's head.

Each coil element was actively decoupled during the transmit portion of the imaging sequence using a pin diode that was actively turned on to create short circuit across the series inductor and the shunt capacitor of the match circuit which, when tuned properly, create a parallel resonant circuit and an approximate open circuit at the coil input terminals. Each coil was also passively decoupled during Tx with the use of reverse diode pairs and a series inductor in parallel with one of the tuning capacitors of the loop. In the case of failure for the active diode to turn on, the large currents included in the loop from the Tx pulses of the body coil would turn on the diodes and again create a parallel resonant open circuit across the given capacitor. (24). The circular coil elements allowed for optimal coil overlap and for reduction of mutual inductance coupling between nearest neighbor elements.

Homogenous Phantom

The homogeneous head phantom (see Fig. 3) was constructed from fiberglass using the same human head mold that was used to construct the coil mask. It was filled with 5 liters of deionized water, and 1.955 g/L of Copper Sulfate ($\text{CuSO}_4 \cdot 5\text{H}_2\text{O}$ Fisher Scientific Company, Fair Lawn, NJ, USA) (25). Several coils were then tuned and matched on the mask former to a human head that had average loading. Then NaCl was added to the phantom until similar loading to that of the head was achieved.

Bench Tests—The coil elements were bench-tested for: tune, match, nearest-neighbor isolation, active and passive detuning, and optimization of the preamp detuning, the phase shifter, and cable trap resonance. The coils were positioned on the mask and S21 coupling measurement between neighboring coils, distortion of the single coil S11 curve, and a switch test, were performed before the coil was secured to the mask with adhesive transfer tape. The S21 coupling between nearest-neighbors was measured by hooking up two adjacent coils to the network analyzer, with all other coils detuned, and overlapping until the lowest achievable power in decibels (dB) transferred between the two coils was measured. The S11 test involved looking into one coil via the network analyzer with an adjacent coil resonating and all the other coils detuned and overlapping so that a resonant curve similar to a single coil curve was achieved. The switch test involved using a double loop probe to look at the resonance of a single coil while the surrounding coils were detuned. Then one at a time each adjacent coil is turned on. If the resonance of the primary coil is not distorted the coil passes the switch test. Final coil coupling was also verified through noise correlation measurements by insuring that no two coils have a noise correlation above 0.5.

An active detuning circuit was constructed using the coil match circuit in conjunction with a shunt PIN diode (MA4P4002B-402; Macom, Lowell, MA, USA). Using a small double loop probe we measured the effectiveness of the active decoupling circuit by comparing S21 of the double loop probe while lightly coupled to the RF coil of interest in the resonant

(without DC bias) and non-resonant states (with DC bias). The matching inductor was adjusted for optimal resonance with the match capacitor and optimal decoupling of the coil element. A similar method was used to optimize and assess the effectiveness of the passive decoupling circuit that was placed on one of the coil tuning capacitors. This passive decoupling was added as a safeguard in case the active decoupling failed and assumed that, in this situation, induced currents in the receiver coil from the transmit coil would be large enough to activate the passive decoupling diodes and trigger the passive decoupling resonance circuit.

Preamp decoupling was achieved using a short 24 cm cable, utilization of a phase shifter, and tuning of the preamplifier input matching circuit to transform the low impedance input impedance of the preamps to the coil detuning/matching circuit. Measurement of the preamp decoupling was performed using a double loop probe to measure the insertion loss difference, in dB, between a given single resonant loop with and without being attached to its associated preamplifier. Adjustment of the preamplifier decoupling was then achieved by adjusting a variable capacitor on the input tank circuit.

Imaging Experiments—Imaging experiments were performed on the homogenous head phantom and on human subjects to evaluate and quantify the effectiveness of the optic nerve coil array. For SNR comparisons, a proton density gradient-echo sequence was used with $(0.2 \text{ mm})^3$ isotropic spatial resolution, TR 25 ms, TE 4.12 ms, and flip angle 25° . Phantom images were used to assess the relative SNR (rSNR) between the 20-channel optic nerve coil array and a commercial 12-channel head coil array (Siemens Medical Solutions, Erlangen Germany). Three orthogonal (axial, coronal, and sagittal) rSNR profiles through the phantom at the approximate location of the optic chiasm were obtained. Both magnitude images and raw data were saved, and a noise reference was obtained by recording an image without RF excitation. The rSNR plots for each channel were calculated from the raw data by dividing the signal sensitivity by the standard deviation of the background (region of no signal or artifact) noise for each channel. Combining the rSNR images from all the channels using the square root of the sum-of-squares method generated composite image SNR maps (26-29). The noise correlation matrix was calculated from the noise image data obtained without RF excitation.

Inverse g-factor maps were computed from phantom data to assess parallel imaging performance. For a g-factor comparison between the commercial coil and the optic nerve coil, raw k-space data was acquired in an axial slice of the homogeneous head phantom using the same scan parameters as the SNR scans. The g-factor maps were plotted as the inverse g-factor, $1/g$, to allow all of the maps to be shown with the same color scale. The maximum g-factor in each slice was also recorded.

Human imaging studies—The optic nerve coil array was developed for and tested on a 32-channel 3T clinical scanner (MAGNETOM Trio, Tim System; Siemens Medical Solutions, Erlangen, Germany). The University of Utah's Institutional Review Board for human research approved the studies and informed consent was obtained from each volunteer.

Twelve human volunteers, with informed consent, were scanned using the optic nerve coil and the 12-channel head coil. The human imaging studies used a 2D turbo spin echo (TSE) acquisition with a $0.4 \times 0.7 \times 2.5 \text{ mm}^3$ voxel size and TR/TE of 4.5s / 115 ms. Slices were taken parallel through the optic nerve. Human images were obtained as a measure of coil sensitivity and image quality.

The optic nerve diffusion weighted image (DWI) from six healthy volunteers were acquired using a 2D singleshot interleaved-multislice-inner-volume diffusion-weighted EPI (2D ss-IMIV-DWEPI) (31) with the following parameters: $1 \times 1 \text{ mm}^2$ inplane-resolution with 2 mm slice thickness, TR 3.0 sec, TE 76 ms, 12 magnitude averages, and interleaved acquisition of 8 contiguous slices. Diffusion gradients were applied along 12 non-collinear directions using b-values of 0 and 500 s/mm^2 (31). DW images were post-processed using DTI analysis software written in IDL (ITT Visual Information Solutions., Boulder, CO) to calculate apparent diffusion coefficient (ADC) maps, the six independent diffusion tensor components, the eigen-values and eigenvectors, and fractional anisotropy (FA), and to visualize the results using a RGB representation of the principal eigenvector. The two regions-of-interest (ROIs) including only the center of the left and right optic nerve were manually selected on b_0 ($b=0$) images and used to measure the fractional anisotropy and the mean diffusivity value.

Results

A coil tune of 123.0 to 123.3 MHz and a match (S11) of -30 to -40 dB to the homogenous head phantom were achieved for all coils. The match for the human volunteers was not as precise due to the variability in head size. An active decoupling measurement of at least 30 dB isolation and a passive decoupling measurement of better than 35 dB isolation was achieved for all coils. Preamp decoupling provided approx 20 dB decoupling for all channels. The S21 values recorded for nearest neighbors varied between -11 and -16 dB and the noise correlation matrix values of nearest neighbors varied between 0.2009 and 0.4497 with a mean value of 0.2689.

The rSNR maps (Fig. 4) showed the extent to which the optic nerve coil provided increased SNR over the entire region of the optic nerve. The rSNR line profiles (Fig. 5) showed that the optic nerve Coil provided 35% greater rSNR at the optic chiasm and approximately 300% near the orbit compared to the 12-channel coil. The rSNR line profiles also show a 300 to 400% rSNR improvement near the coil elements when compared to the 12-channel coil. Inverse g-factor maps are shown in figure 6. These maps show that the coil performs well for parallel imaging over the region of the optic nerve.

Human volunteer images demonstrate the increased image quality that resulted from using the optic nerve coil for imaging the region of the optic nerve between the orbit and chiasm. (Fig. 7) Optic nerve lesions are indicated with arrows in figures 8 and 9. Figure 8 displays an image of the optic nerves of a patient who has chronic bilateral optic neuritis. The images show a plaque on the left and a signal abnormality on the right that corresponds well with the disease history. Figure 9 displays an optic nerve image where a plaque in the left immediate retrobulbar region is found within the center of the optic nerve head. A fundusoscopic exam confirmed a pronounced pallor on the left optic nerve.

A total 8 slice locations of diffusion tensor images were acquired from 6 healthy volunteers. Typical diffusion tensor images obtained using 20-channel optic nerve coil and 2D ss-IMIV-DWEPI of healthy subject are shown in figure 10. An FA map, RGB representation of principal eigenvector map and mean diffusivity map (ADC) were created from diffusion weighted images with 12 diffusion-encoding directions. The optic nerves are clearly seen surrounded on both sides of cerebral spinal fluid on the FA, RGB and ADC maps. The mean diffusivity of the normal optic nerve was measured as $0.470 \pm 0.05 \times 10^{-3} \text{ mm}^2/\text{s}$. Figure 11 displays (a) the FA map, (b) RGB fibermap, and (c) ADC map of one slice from a subject with optic neuritis and the corresponding T2w image. The FA value of the neuritis legion, which is indicated with the red arrows, is measured as $0.61 \times 10^{-3} \text{ mm}^2/\text{sec}$, which are larger than the contralateral normal nerves.

Discussion

The semi-flexible mask design of the optic nerve coil allowed a close fit to the head for all head sizes; large and small. The coils on the front of the head remained well matched while the elements toward the back of the head had the most variability in the match for different head sizes. The isolation between overlapping coils was sufficient but could be increased by reducing the capacitive coupling and flux shielding between the loops through the use of copper wire loops rather than the copper foil used in this study. The small element size provided significantly higher SNR at the skin surface and achieved a relatively high SNR in the center of the head compared to the 12-channel head coil.

The goals of this project were to create a coil that would provide improved visualization of the optic nerve and increase the diagnostic capability of optic nerve imaging. These goals have been achieved. The signal inhomogeneity of the close fitting phased array coil was of little concern because homogeneity of the coil sensitivity was not needed for the diagnostic techniques used. The increases in SNR provided by the optic nerve coil can be used to obtain images with higher spatial and/or temporal resolution. The improvement in spatial resolution helps to improve the visualization of fine detail in the optic nerves while improvements in temporal resolution help to reduce head and eye motion artifacts and reduce overall patient imaging time. The improvement in parallel imaging obtained by using the optic nerve coil are the greatest at the front of the head and diminishes towards the back of the head where the coil density is reduced.

Very early neoplastic processes that affect the optic nerve and appear normal on conventional MRI may reveal abnormalities in the DTI measurement. However, DTI of the optic nerve remains a difficult task due to the small size (≤ 3 mm diameter) of the optic nerve and the surrounding cerebral-spinal-fluid and fat, which often induce the geometric distortion using a conventional singleshot DWEPI technique that is widely used in human DTI study. As the accuracy of the DTI measurement increases with the increased SNR, our improved optic nerve coil may benefit to the DTI measurements of the optic nerve. To improve the accuracy of DTI measurement the RF coil should provide the decent SNR and coverage particularly around optic nerve area. DTI measurements in normal optic nerve using our 20-channel coil are relatively same as the results previously reported (30). Our results indicate that 20-channel coil allows for the reliable DTI measurement of the optic nerve.

Quantitative analysis of diffusivity and anisotropy in the human optic nerve using the 20-channel coil appears to be feasible and may enable more sensitive and specific detection and monitoring of structural changes caused by pathology including multiple sclerosis, optic neuritis, and neurofibromatosis.

Future modifications might include adding coil elements at the back of the head in order to achieve increased SNR throughout the entire visual pathway from the optic nerve to the visual cortex or to improve other brain imaging capabilities in addition to imaging of the optic nerve. This would also give flexibility in the use of the coil for many other head imaging uses. Implementation of wire loop coils could reduce the copper shielding and inter-coil capacitance between loops in the array and provide better isolation, which would increase the parallel imaging capabilities, and possibly boost the SNR.

Conclusions

A 20-channel phased array coil was successfully constructed specifically for the purpose of imaging the optic nerve in humans. It was shown that the optic nerve coil provided greater SNR and parallel imaging capabilities for imaging optic nerve from the eyeball to the optic

chiasm. Clinical imaging results showed that the findings in the images were consistent with the patient's optic nerve disease histories. The coil allowed for the successful DTI of the optic nerve.

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Figure 1.
Optic nerve coil consisting of 20 overlapped circular coil elements mounted on a semi-rigid flexible fiberglass mask former.

Figure 2. Single coil channel including the coil and preamp board that includes the cable trap, pi network phase shifter, and preamp.

Figure 3.
Head phantom. (A) Side view. (B) Top view.

Figure 4. Relative signal-to-noise ratio images (rSNR) of the 12- and 20-channel coils through the region of the optic chiasm from a homogeneous head-shaped phantom.

Figure 5.

The rSNR profiles of the 12- and 20-channel coils through the region of the optic chiasm from a homogeneous head phantom. Results from a single channel transverse electromagnetic coil are also included.

Figure 6.

Inverse g-factor maps from phantom data for a slice through the region of the optic nerve using the 20-channel ON coil. Images show results for reduction factors of 2, 3, and 4. $G_{\max}$ for each case is also presented. The g-factor is significantly reduced in the region of the optic nerve.

Figure 7. Example human imaging results of the region of the optic nerve from the 12-channel head coil (left) and 20-channel ON coil (right).

Figure 8. Patient studies: This patient had a history of chronic bilateral optic neuritis. These images show a plaque on the left and a signal abnormality on the right that corresponds well with the disease history (arrows).

Figure 9. Patient studies: Display of a plaque in the left immediate retrobulbar region within the center of the optic nerve where the nerve exits the eye. A fundoscopic exam showed a pronounced pallor on the left optic nerve (arrows).

Figure 10.

DTI measurements of a healthy human optic nerve using 20-channel optic nerve coil and 2D ss-IMIV DWEPI sequence: fractional anisotropy, principal eigenvector maps, ADC map. The color coding: red, Left/Right; blue, Superior/Inferior; green, Anterior/Posterior.

Figure 11. FA (a), RGB (b), and ADC (c) map and corresponding T2w image of a subject with optic neuritis. Red arrows indicate the area in which FA values are measured.