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Interchangeable Neck-Shape-Specific Coils for a Clinically Realizable Anterior Neck Phased Array System

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Abstract

Purpose—To demonstrate the interchangeable neck-shape-specific (NSS) coil concept that supplements standard commercial spine and head/neck coils to provide simultaneous high-resolution (hi-res) head/neck imaging with high signal-to-noise ratio (SNR).

Methods—Two NSS coils were constructed on formers designed to fit two different neck shapes. A 7-channel (7ch) ladder array was constructed on a medium neck former and a 9-channel (9ch) ladder array was constructed on large neck former. Both coils were interchangeable with the same preamp housing.

Results—The 7ch and 9ch coils demonstrate SNR gains of approximately 4× and 3× over the Siemens 20 channel head/neck coil in the carotid arteries of our volunteers, respectively. Coupling between the Siemens 32 channel spine coil, Siemens 20 channel head/neck coil, and the NSS coils was negligible allowing for simultaneous hi-res head/neck imaging with high SNR.

Conclusion—This study demonstrates that supplementing existing commercial spine and head/neck coils with an NSS coil allows uniform simultaneous hi-res imaging with high SNR in the anterior neck, while maintaining SNR of the commercial coil in the head and posterior neck.

Keywords

magnetic resonance imaging; RF coil; phased-array; signal-to-noise ratio; cervical; neck

Introduction

There are some pathologies that require high resolution (hi-res) simultaneous imaging of the head and neck. For example, MRI for stroke workup requires imaging the carotid arteries

from the aortic arch to the circle of Willis in order to rule out all possible sources of emboli (1–3).

Unlike the head and posterior neck, which have consistent shapes and are relatively easy to image, the anterior neck is difficult to image because of the many different shapes and sizes of the human neck anatomy (see for example, Figure 1). The shape and size of the head does not differ significantly in adults, therefore, manufacturers have been able to design head coils that accommodate all head shapes that are form fitting and provide high signal sensitivity or signal-to-noise ratio (SNR) images (4–7). Similarly, the cervical spine is consistent enough in shape variation that manufacturers have been able to develop single coil solutions that allow for general-purpose high SNR imaging of the cervical spine and posterior neck. The adult anterior neck, however, with its significant variation in shape and size does not have a commercially available high SNR general-purpose coil solution that is currently available for any of the major MRI manufacturers. For example, current anterior neck coil components of commercial head/neck coils have been designed with coil elements that are relatively far from the surface of the neck in order to accommodate all neck shapes. Since SNR is significantly improved by positioning the coil elements closer to the anatomy of interest, these large volume neck coils are used at the expense of substantially reduced SNR compared to coils that form well to the anterior neck anatomy.

A number of close fitting anterior neck coils have been designed (8–11), primarily for carotid bifurcation imaging, but for several reasons, they are hard to use clinically. For example, these coils do not fit all patients or provide full anterior neck coverage, are not easily positioned/fastened for quick patient setup times, and typically, cannot be used simultaneously with OEM coils. The first surface coil used for hi-res carotid imaging was the four-channel bilateral carotid coil designed by Hayes et al (8) and a carotid coil based on the same concept is available commercially for several different MRI vendors (Machnet, Roden, The Netherlands). These coils provide high SNR over a small volume, but often must be repositioned to place the coil's high signal sensitivity volume over the carotid bifurcation, which varies in location among patients (8). Coil repositioning during the patient exam is not clinically desirable because it decreases patient throughput. The coil's high sensitivity volume and SNR was improved by the eight-channel bilateral carotid coil design presented by Balu et al (9) and developed into a commercially available neck coil for Philips MRI systems (Shanghai Chenguang Medical Technologies, Shanghai, China). Also, a bilateral six-channel carotid coil is currently available for General Electric MRI systems (NeoCoil, Pewaukee, Wisconsin, USA), and a 4-channel, unilateral, (8 channel bilateral) general-purpose coil that can be used for carotid imaging is available for Siemens MRI systems (Siemens Healthcare, Erlangen, Germany). In some instances, these coils will not fit patients with short necks. All of the above bilateral coils can be positioned with uneven pressures on the neck, causing variations in position and flow in the neck blood vessels, and reducing consistency in longitudinal studies (12). Finally, Tate et al. designed a single rigid former sixteen channel coil that has a large imaging volume and high SNR throughout the anterior neck and eliminates many of the bilateral coil problems (10). However, the rigid single former design only fits closely against the neck for a subset of patients and there are some patients such as patient 3 in Figure 1 for whom it would not fit. Rigid single former designs can provide inconsistent SNR between patients of differing neck sizes. Finally, using

the above mentioned coils requires removal of any OEM head coils on the patient table. Comprehensive, simultaneous, high-resolution, high-SNR imaging of blood vessels in the head and neck is not achievable with these coils. Though these coils could be redesigned to work simultaneously with head, neck, or spine coils, they still have the previously discussed shortcomings.

This work demonstrates that comprehensive, simultaneous, high-resolution, high-SNR imaging of the head and neck can be accomplished by using interchangeable neck coils on multiple form-fitting neck formers in conjunction with existing clinical spine and head/neck coils. We present two interchangeable neck-shape-specific (NSS) anterior neck coils that are designed to integrate with existing commercially available spine and head/neck arrays. These NSS coil systems use an independent multi-channel pre-amp module that connects to any one of the interchangeable anterior NSS arrays. The SNR gains and corresponding image improvements that can be obtained using interchangeable NSS coil arrays is demonstrated in this work by comparing these two different sized NSS arrays to a typical commercial anterior neck coil.

Methods

The challenge of designing a close-fitting full-coverage neck coil is illustrated by the different neck shapes shown in Figure 1. To test and illustrate the concept of the NSS coils, a medium and a large coil former were shaped to fit two different neck sizes and on each former, a coil was designed, constructed and tested.

Phantom and human studies were performed to evaluate and demonstrate the value of the NSS coils. All imaging studies were performed on a Siemens MAGNETOM Prisma 3 Tesla MRI scanner, supplemented with the Siemens 20 channel head/neck and Siemens 32 channel spine coil. For the remainder of the text in this document the Siemens 20 channel head/neck coil will be referred to as either the OEM head/neck, OEM head, or OEM neck coil depending on whether the 20 channels of the head/neck, 16 channels of the head, or 4 channels of the neck are utilized. Additionally, there are 2 anterior (OEM anterior neck) and 2 posterior (OEM posterior neck) coil elements in the OEM neck coil and for the Siemens 32 channel spine (OEM spine) coil only 4 superior coil elements were used.

Coil Requirements

The goal of this work was to develop neck coils that would 1) provide high SNR over the entire region of the anterior neck, 2) integrate and operate with the existing OEM head/neck/spine coils, 3) be robust and economical in construction and 4) be easily used and provide consistent image quality in a busy clinical environment.

The most direct method to improve SNR in the anterior neck is to use RF coils that are designed to fit closer to the anterior neck anatomy. This can be accomplished by placing the coil elements on a former that fits closely over the entire neck. We refer to such a coil as a neck-shape-specific (NSS) coil. A coil former that fits well to the patient neck allows tiled coil elements to cover the entire anterior neck and thereby significantly increases surface

coverage of the NSS coil compared to previously published bilateral carotid coils. This complete coverage makes repositioning unnecessary.

Operation with the OEM head/neck coil required that the NSS coils be sufficiently thin to fit between the patient and existing anterior neck portion of the OEM head/neck coil.

Otherwise, for some subjects, removal of the anterior portion of the OEM head/neck coil would be required to scan with the NSS coil in place. Although small clearances only exist for very large necked patients, such as patient 3 of Figure 1; we chose to use the same pre-amp set regardless of interchangeable former used. The thin coil constraint required that the pre-amps not be mounted on the coil, but in close proximity. Cables were used to connect the coils to preamp housings that were placed on the left and right side of the OEM head/neck coil. These cables were attached to the inferior side of the NSS coils unlike the bilateral coils, which connect to the superior side of the coil. The location of the preamps for both the bilateral coils as well as the single rigid formers that were previously published (8–10), required removal of the head/neck coil to make room for the preamp housings and positioning hardware. The NSS coils do not require positioning hardware or large preamp housings allowing the coil to be used with the head/neck coil in place.

The desire that the NSS coils be constructed using robust and economical techniques added several additional requirements. First, to ensure proper preamp operation the preamps were oriented properly relative to the DC magnetic field. Improper orientation will affect the gain, input impedance, and noise figure (15,16). Second, to reduce cost the NSS coils were designed to connect interchangeably with a single pre-amp assembly. Third, to reduce construction time and cost, a ladder array was chosen because it provided full coverage of the neck former and, in our experience, it was faster and easier to use capacitor decoupling in a ladder structure than it was to adjust overlapping elements for minimum mutual inductance. Fourth, although a flexible coil former could be used, a semi-rigid coil former was chosen to ensure long-term reliability.

Finally, the use of a semi-rigid coil former, has the added benefit that the coil is easy to position on the patient neck for improved clinical utility and minimizes the likelihood of applying uneven pressure against the sides of the neck. To ensure a close fit and to minimize coil motion we designed the single semi-rigid former to fit on the neck similar to an anterior neck brace. Because of the variation in neck size and shape between individuals, coils designed on multiple different NSS formers are needed to fit the range of shapes and sizes encountered.

Former Selection

The formers were based on two different sized volunteers that were typical of patients with carotid artery disease seen at our institution. The smaller former was molded from a male volunteer whose neck diameter was 12 cm and had a length of 8 cm measured from the chin to the clavicle, typical of a “medium” neck. The large former was molded from a male volunteer with a neck diameter of 14.7 cm and a length of 7 cm, typical of a “large” neck. The large former was also designed to account for and accommodate patients with a buccula (“double-chin”). Aquaplast pelvis thermo-plastic (QFix, Avondale, PA, USA) was chosen as

the former material for both coils because of its semi-rigid properties and ease of use for attaching coil elements.

7ch and 9ch Design and Construction

Once the shapes and sizes of the formers were selected, an estimate of the number of channels to use was obtained by using Biot-Savart field simulations of a simple ladder coil layout on cylindrical phantoms that roughly matched the diameters of the NSS formers. In these simulations, the spacing of the ladder rungs and the number of channels were chosen based on maximizing the SNR over a small extended volume at the depth of typical carotid arteries (13,14). These results yielded a reasonable number of coil elements to be used for each former. It was felt that these estimates, although not guaranteed to be optimal, would be adequate for testing the concept of NSS coils that can be used simultaneously with the OEM head coil.

The NSS coils were built on two separate formers with the copper traces bonded to the formers in a ladder array layout as shown in Figure 2. Copper traces 6.35 mm wide were adhered to the former using adhesive transfer tape (3M, St. Paul, MN, USA). Conducting pads for the capacitors and for shorting copper traces were fabricated on 5 mm square FR4 material with vias connecting the pads on opposite sides of the FR4 material for easier soldering.

For both the 7ch and 9ch ladder arrays nearest neighboring coil elements were decoupled using a capacitor in each common rung instead of by overlapping (17). Also, for both arrays the non-central rungs were separated by 3.1 cm. The two central rungs, were separated by 3.6 cm and 4.1 cm for the 7ch and 9ch coils, respectively. The most posterior loops were 8.1 cm in length and decreased to 7.1 cm in length for the central loop of the 7ch coil and were 12.6 cm in length and gradually decreased to 6.6 cm in length for the central loop of the 9ch coil. Coil loop measurements were made from the center of the copper traces. A hole was cut out of the former in the center loop of the 7ch coil to accommodate for a laryngeal prominence, to improve patient comfort, and to reduce coil movement if the patient swallows. A hole could have been cut out of the former in the central loop of the 9ch coil, but this did not seem to be necessary because the laryngeal prominence has been less prominent for our large neck subjects.

All coil elements for each array were connected to preamps through non-magnetic RG-316 cables. The cable length for the central coil element of both NSS arrays was 65 cm, and all non-central coil element cables were 60 cm in length. Low pass pi phase shifter circuits completed the 180° phase shift between coil match circuits and the preamplifier inputs as required for maximum preamplifier detuning. The central 20 cm of the total cable length for each channel was used to construct a solenoid trap (18) to block common mode currents and reduce cross talk between cables. In addition, bazooka baluns (18) were made by placing copper braided sleeves over the coax cable with one end shorted and one end with a capacitive termination to the coax ground shield to create shorter balun lengths. These were placed between each coil match circuit and the solenoid trap to further reduce common mode currents at the coil and limit the coupling between the cables of the array. RF connectors (LEMO Rohnert Park, CA, USA) were placed 10 cm from the preamps allowing

each of the NSS coils to be interchangeable with the same preamp housings. Low input impedance 3T preamplifiers (Siemens Medical Solutions, Erlangen, Germany) were used to amplify the received signal. A short multi-channel cable with a solenoid balun in the ground shield was used between each preamp housing and the MRI table plug. A circuit diagram is shown in Figure 3.

Coil elements were labeled from patient right to left side as R4-1 to L1-3 for the 7ch coil and R5-1 to L1-4 for the 9ch coil. The center coil element was R1 for both NSS coils. The preamp housings were plugged into two unused coil ports on either side of the standard head coil on the Siemens PRISMA patient table. Coil elements R1-R4 (R1-R5) of the 7ch (9ch) were attached to the patient-right preamp housing while coil elements L1-L3 (L1-L4) of the 7ch (9ch) were attached to the patient-left preamp housing. Separate coil selection script files for each coil array selected the proper coil files that accounted for the number of channels required and prevented the sampling of unused channels during a scan.

S21 measurements, made with common mode current probes, with and without the trap and balun circuitry between the probes, were used to assess the effectiveness of the solenoid traps and bazooka balun. The solenoid trap measurements were made directly on the coil while the bazooka balun measurements, which could not be made on the coil, were made using an equivalent test configuration.

Phantom Studies

Two cylindrical phantoms of 11.5 cm and 14.5 cm diameter, both filled with $3.75\text{g NiSO}_4 \times 6\text{H}_2\text{O} + 5\text{g NaCl}$ solution per 1000g H_2O distilled, were used to acquire the data needed for inverse geometry factor maps (1/g-factor) (19). These maps were calculated from 2D gradient echo (GRE) images with a TE/TR = 4.0/500 ms, flip angle = 90° , matrix size = 320×320 , FOV = 300×300 mm, bandwidth = 260 hertz/pixel and a noise-only image using the same sequence with the RF transmit voltage set to 0 volts and TR = 50 ms. The 11.5 cm diameter phantom was used to obtain 1/g-factor maps for the 7ch+OEM neck/spine coil while the 14.5 cm diameter phantom was used to obtain 1/g-factor maps for the 9ch+OEM neck/spine and OEM neck/spine only coils.

Human Studies

Institutional IRB approval was obtained for the human imaging studies and informed consent was obtained from each volunteer. Three volunteers were scanned to acquire SNR, noise correlation, or head/neck images. All images were acquired using either the 7ch or 9ch NSS coils in conjunction with the OEM coils.

The SNR maps were calculated (20) from data acquired using a 2D GRE sequence (TE/TR = 4.0/500 ms, flip angle = 90° , matrix size = 320×320 , FOV = 300×300 mm, bandwidth = 260 hertz/pixel) and a noise-only image using the same sequence and parameters, except TR = 50 ms and the RF transmit voltage was set to 0 volts—the shorter TR just reduces the scan time but should have no effect on the measured noise. These same noise only image parameters were used to acquire the data used to calculate the noise correlation plots (20).

A 3D T1 weighted Sampling Perfection with Application optimized Contrast using different flip angle Evolutions (SPACE) (21) sequence with Delay Alternating with Nutation for Tailored Excitation (DANTE) (22) flow suppression was performed with the following parameters: coronal plane, isotropic voxel dimension = 0.73 mm, TE/TR = 22/800 ms, and scan time = 5 minutes.

A hi-res 3D Ciné Reconstruction Method Using Retrospective Ordering and Compressed Sensing (Ciné-ROCS) (23) DANTE-prepare fast low angle shot (DANTE-Prep FLASH or DASH) (24) image covering the head and neck vasculature was acquired with the following imaging parameters: coronal plane, isotropic voxel dimensions = 0.63 mm, TE/TR = 2.5/8.0 ms, FOV = 240×240 mm, matrix size = 384×384×96, DANTE prep = 150 ms, flip angle = 8.0°. The scan time for the 3D Ciné-ROCS DASH images with two averages was 8 minutes 20 seconds. Two spatial saturation RF pulses were applied left and right of the imaging location to suppress the signals from the shoulders. A chemical-shift fat saturation RF pulse was applied to eliminate perivascular fat.

2D Transverse T1 THIN images were obtained with the following parameters: voxel dimensions = 0.3125×0.3125×2.5 mm, TE/TR = 12/425 ms, FOV = 160×160 mm, matrix size = 512×512, flip angle = 180°.

The NSS arrays were also used to image 25 patients with carotid artery disease as part of a research protocol. Of those 25 patients, 14 patients were imaged with the 9ch and 11 patients with the 7ch. Comparison studies were not conducted on these patients, instead we made general observations of coil performance and fit. An additional 25 subjects were scanned with these coils, but are not included because the specific coil used (7ch or 9ch) was not recorded.

Results

7ch and 9ch Design Results

Preamp, passive, and active detuning measurements for the 7ch and 9ch coils were measured to be better than 25.3 dB, 45 dB, and 45 dB, respectively. All detuning measurements were calculated as the difference between the resonant and detuned state of the loop. These S21 measurements were made using a double loop probe lightly coupled to the coil loop being tested. When tuned, the solenoid traps and bazooka baluns reduced common mode currents by at least 20 dB and 10 dB, respectively.

Phantom Results

Inverse geometry-factor maps (1/g-factor) are shown in Figure 4. These maps show that acceleration can be done in both the Anterior/Posterior and Left/Right phase encoding direction with significant improvements over the OEM neck coil.

Human Results

SNR measurement results are shown in Figure 5. The 7ch and 9ch coils demonstrate SNR gains of approximately 4× and 3× over the OEM neck coil at the carotid arteries for these

volunteers (~1.75 cm and ~4 cm below the surface of the skin respectively). The OEM head coil was not used in this SNR comparison

Images of the element by element noise correlation between NSS coils and the OEM head/neck/spine coils are shown in Figure 6. Maximum correlation between any of the elements of the 7ch+OEM head/neck/spine as well as the 9ch+OEM head/neck/spine did not exceed 0.34. The maximum correlation between the 7ch and OEM head/neck/spine coils was 0.17 and maximum correlation between the 9ch and the OEM head/neck/spine coils did not exceed 0.22.

Figure 7 shows 3D T1w SPACE images comparing images from the 9ch (Fig 7: A and C) and the OEM neck coil (Fig 7: B and D). The 9ch coil has significantly better vessel depiction than the clinical coil and the 7ch coil gives similar results. Also shown in Figure 7 are clinical hi-res 3D Ciné-ROCS DASH images of the head and neck, acquired simultaneously with the NSS+OEM head coils (Fig 7: E and F) to compare against images acquired simultaneously with the OEM head/neck/spine coil (Fig 7: G).

2D Transverse T1 THIN images comparing the 7ch+OEM neck with the OEM neck coils are shown in Figure 8. The 7ch coil provided higher image quality at the level of the vocal cords and arytenoids (Fig 8: A) as well as better visualization of the cricoid cartilage (Fig 8: C). These structures are important landmarks in the diagnosis of laryngeal pathology, and serve as relevant structures in the staging of laryngeal squamous cell carcinoma.

When imaging the 50 patients with the NSS coils we found that the 7ch coil fit all patients with a medium neck well. A well-fitting coil was defined as the coil being able to be comfortably placed against the neck as intended without any rotation or manipulation. The neck length of all medium necked patients was long enough to accommodate the 7ch coil. The 9ch wrapped further around the neck than the 7ch resulting in a poor fit for some patients. For some imaging studies, the 9ch coil could not be positioned properly because of interference with the padding behind the neck or with the shoulders. Some patients had a buccula (“double-chin”) which required an undesirable rotation about the Left/Right axis for the 7ch coil to fit against the neck surface. The 9ch coil was designed with the buccula taken into account and didn’t require the same Left/Right axis rotations for proper positioning. For some patient studies the coils would get nudged off the neck by patient motion or because of fitting problems. This resulted in unwanted gaps between the surface of the neck and the coils, and consequently, less than ideal coil performance. We found that this problem could be solved using a strap to hold the coil in position on the patient neck. In all completed imaging studies, regardless of coil fit, the NSS coils outperformed the OEM neck coils in SNR.

Discussion

This paper has demonstrated the feasibility of developing a form fitting neck-shape-specific coil that achieves very high SNR over the anterior neck that can be used simultaneously with the OEM high performance head coils. Using the resulting composite coil, pulse sequences

that achieve high resolution images over the entire head and neck simultaneously with no gaps in signal sensitivity can be applied.

The interchangeable coil concept using the Limo connectors worked well. There was no observable degradation in performance from switching the different NSS coil formers into the same preamp housing. Whereas typical Siemens coils switch the output of the preamp into the same plugs, the NSS coils switch the coils on the different NSS coil formers into the same preamp housing.

Although SNR and parallel imaging performance might be further improved with more complicated coil layout and more RF channels, the ladder array provides a good compromise between substantial SNR gains over the commercial neck coil and simplified robust coil construction.

Coil fit was affected primarily by neck girth and length. In addition, the amount of fat around the neck and the angle of the neck/torso junction were important factors in obtaining a good NSS coil fit. Generally, the 7ch coil fit the 11 different patients well. Patient studies were approximately an hour and there were no complaints of patient discomfort from the NSS coils for any of the patients that were scanned.

The NSS coils are very easy to place and can easily be fastened down with a strap. Placement of the NSS coils is much easier to accomplish than placement of a bilateral coil and generally requires less time. Overall, placement of the NSS coils required very little time (only a minute or two) and the increased benefit provided by the coil was deemed to outweigh the extra cost in time, especially since the coil never requires repositioning due to the large coil coverage. There were some studies where the 7ch NSS coil was selected for a patient only to find out that the 9ch would be required.

As can be seen from Figure 4 the NSS coils provide significantly increased g-factor performance when compared to the OEM neck coil. This acceleration could be increased even further with higher density coil layouts. The relative importance of parallel imaging on human subjects remains to be demonstrated.

Although the center coil element in each NSS array has little effect on carotid vessel SNR, having a hole in the former for the laryngeal prominence increased patient comfort and reduced coil movement for medium sized neck volunteers. The central loop also provided high SNR for imaging the anterior structures of the neck such as the larynx and thyroid. This center coil provided continuous anterior neck coverage, which can be lacking when a bilateral coil design is utilized.

Many coils are designed with the preamps located close to the coil elements which increases the strength of preamplifier decoupling and overall SNR. Without the requirement that the NSS former needs to be thin to fit under the anterior portion of the OEM neck coil; we could place the preamps for the NSS design on the coils themselves. This would provide higher SNR over the current design.

Unintentionally, all neck elements of the OEM head/neck coil and the 7ch coils were used to acquire images in A and C of Figure 8. Normally, when imaging with the NSS arrays, the OEM anterior neck elements are turned off. Images were not re-acquired, however, because previous tests had demonstrated no visible difference between images acquired with and without the OEM anterior neck elements when using the 7ch coil due to negligible coupling between the two arrays.

Future work will investigate the minimal number of formers needed for high SNR imaging of the anterior neck, with their correct shape and size to fit the general patient population. Then the optimal coil layout will be calculated for optimal SNR and parallel imaging performance for those specific former shapes. Future work will also investigate the SNR gains over the OEM neck coil as a function of patient neck size and shape; more specifically, neck shapes that may be too large to achieve any SNR gains using the NSS coils, as compared to the OEM neck array will be determined.

Conclusion

We have demonstrated the interchangeable NSS coil concept by constructing two interchangeable anterior neck coils for both medium and large neck volunteers. These interchangeable anterior neck coils supplemented the OEM head/neck/spine coils to provide simultaneous hi-res head/neck imaging with high SNR where the SNR of these commercial coils is low in the anterior neck otherwise. NSS coils are easier to position, and do not require repositioning. Additional NSS coils will provide a close fitting coil for a large variety of neck shape and size. The NSS coil concepts overcome the coil related barriers that have prevented high-resolution neck imaging from inclusion in standard clinical practice.

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Figure 1.

Models generated from three patients with carotid artery disease show the variation in neck habitus. The NSS coils presented in this paper will fit patients with medium and large necks, but some patients, such as patient 3 above, would require an additional NSS coil. Models were generated from CTA data using 3D Slicer (<https://www.slicer.org>).

Figure 2.

The Siemens 20 channel head/neck coil with the A) 7ch and the B) 9ch. The integrated neck coil on the Siemens 20 channel head/neck coil limits the thickness of the NSS coils for some larger patients since these formers are designed to be placed on the neck at the same time as the Siemens neck coil. Also shown are the C) connectors allowing the carotid coils to be interchangeable with the preamp housing. The white colored arrows point to the bazooka baluns while the black arrows point to the solenoid traps.

Figure 3.

Circuit diagram for both the 7ch and 9ch coils. The bazooka balun was 25 cm long for the central coil of each array and 20 cm for all the other coils of the array. The diagram represents the 60 cm cables that were used for the non-central coil elements. The solenoid trap required 20 cm of cable leaving 20 cm of cable between the solenoid trap and phase shifter circuit. The capacitor values for each loop varied to achieve resonance at the Larmor frequency for each of the different sized coil elements.

Figure 4.

Inverse g-factor maps for the 7ch+OEM neck/spine, 9ch+OEM neck/spine, and OEM neck/spine only coils in the anterior/posterior and left/right directions. The maximum g-factor map values are also displayed below each inverse g-map.

Figure 5. SNR comparison between a small neck volunteer (A, C, E) and medium neck volunteer (B, D, F) using the A) 7ch+OEM posterior neck coil, B) 9ch+OEM posterior neck coil, C) and D) OEM neck coils. E) Ratio between the 7ch+OEM posterior neck (A) and OEM neck coils (C). F) Ratio between the 9ch+OEM posterior neck (B) and OEM neck coils (D). Also, the anterior SNR of the OEM neck coils is lower than the posterior due to the gap between anterior coil elements and volunteer, further justifying the need for better anterior neck coils.

2D GRE sequence, TE/TR = 4.0/500 ms, flip angle = 90°, matrix size = 320×320, FOV = 300×300 mm, bandwidth = 260 hertz/pixel.

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Figure 6.

Correlation plots between the Siemens 32 channel spine (superior elements only), Siemens 20 channel head/neck coil (head and posterior neck elements only) and the 7ch and the 9ch. Maximum correlation between NSS coil elements is less than 0.34. Maximum correlation between NSS coils and Siemens coils is less than 0.22.

Figure 7.

3D T1w SPACE images comparing image quality between the A,C) 9ch and the B,D) OEM neck coil. Images A,B) are coronal slices while C,D) are transversal slices. Image results for the 7ch are similar to the 9ch. 3D T1w SPACE sequence with DANTE flow suppression: isotropic voxel dimension = 0.73 mm, TE/TR = 22/800 ms, and scan time = 5 minutes. Also shown are 3D Ciné-ROCS DASH images acquired simultaneously with the E) 7ch+OEM head coils, F) 9ch+OEM head coils and G) OEM head/neck/spine coils. Imaging parameters for the 3D Ciné-ROCS DASH image: isotropic voxel dimensions = 0.63 mm, TE/TR =

2.5/8.0 ms, FOV = 240×240 mm, matrix size = 384×384×96, DANTE prep = 150 ms, flip angle = 8.0°, and scan time = 8 minutes 20 seconds.

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Figure 8.

2D Transverse T1 THIN images acquired simultaneously with the A,C) 7ch+OEM head/neck coils and B,D) the OEM head/neck coils on a medium volunteer. Imaging parameters for the 2D transverse T1 THIN images: voxel dimensions = $0.3125 \times 0.3125 \times 2.5$ mm, TE/TR = 12/425 ms, FOV = 160×160 mm, matrix size = 512×512 , flip angle = 180° . A and B arrows point to the arytenoids while C and D arrows point to the cricoid cartilage.