Phased Array Coil for Implementing Parallel MRI in Intravascular Imaging: A Feasibility Study

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ABSTRACT: The use of intravascular MRI imaging of vessels will provide high quality images for improved diagnosis and treatment. An intravascular MR receive coil provides high SNR because of proximity to the tissue but respiratory and cardiovascular motion may affect the quality of acquired images. Parallel MRI provides a framework to reduce the scan time using multiple coils. This article presents a novel design of a Phased Array MR receive coil for intravascular imaging which can implement Parallel MRI. Cartesian and Radial GRAPPA are used as the main reconstruction algorithms. We demonstrate that Moving Segment Radial GRAPPA provides good image reconstruction with minimum artifacts when data is under-sampled by a factor of 16. © 2015 Wiley Periodicals, Inc. Concepts Magn Reson Part A 00: 000–000, 2015.

KEY WORDS: parallel MRI; intravascular imaging; SENSE; GRAPPA; radial GRAPPA

INTRODUCTION

Magnetic Resonance Imaging (MRI) is a highly flexible imaging modality with many control parameters and provides robust soft tissue contrast. The ability of

MRI to obtain high resolution images can be utilized in intravascular imaging for the identification and characterization of vascular diseases e.g., atherosclerosis and may improve their treatment. A key component for intravascular imaging is a suitably designed receive coil that provides a high signal-to-noise ratio (SNR) when in close proximity to the anatomy of interest. The receive coil can also provide visual tracking of the catheter in real time without using any other imaging technique (e.g., fluoroscopy) thus increasing the accuracy of the diagnosis (1–3).

The feasibility of a catheter-based MRI was first shown by Kantorin 1984 (4) and since then many research groups have developed intravascular MRI

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Figure 1 The schematic diagram of the proposed coil system.

coils (5–10). However, respiratory and cardiovascular motion may badly affect the SNR and quality of the image in catheter based imaging (11). The reduction in MR scan time offered by parallel imaging may help in minimizing these motion artifacts.

Parallel MRI provides a framework to reduce the data acquisition time but it requires multiple coils for the data acquisition followed by a complex computation algorithm (e.g., SENSE, GRAPPA) (12). The spatial information inherent in each coil element is used to reconstruct the final image. This paper presents a novel design of a catheter based MR receiver coil system for intravascular imaging. This phased array coil is composed of two coil elements, a simple loop coil and an inverse solenoid coil. We have successfully implemented the phased array coil in parallel MRI and demonstrate that the scan time can be reduced by a factor of 16 using non-Cartesian techniques.

THEORY

The main function of an MRI receiver coil is to maximize the signal detection while minimizing the noise. An MRI receiver coil consists of a conductor, which is tuned and matched to receive the signals of only a particular frequency range. The proposed design of the MRI receiver coil system consists of two coil elements mounted on a catheter. One coil element is an inverse solenoid coil and the other is a simple loop coil. The coils are mounted on top of each other to achieve geometric decoupling (12) as shown in Fig. 1. Both coils are intrinsically decoupled because of the different designs and the sensitivity profiles of both coils are different (a fundamental requirement for the Parallel MRI). This particular phased array system is designed to support Parallel MRI and can accelerate the data acquisition process in intravascular imaging.

Figure 2 The final design of the receiver coil system.

Figure 3 The model for the imaging experiments using MRI scanner.

Parallel MRI reconstruction algorithms can be classified into two main categories: (1) Image domain algorithms e.g., SENSE (13), (2) k -space algorithms e.g., GRAPPA (14). The discussion about these methods is beyond the scope of this paper but the details can be found in (12). However, the Parallel MRI methods which require the coil sensitivity information a priori may be difficult to implement in some situations e.g., in intra-vascular imaging when the imaging plane is continuously changing and the blood flow and respiratory movements (15) make it harder to acquire a good estimate of the receiver coil sensitivity. In the literature (16–18) there are suggestions for the use of k -space based Parallel MRI techniques [e.g., GRAPPA (14)] for intra-vascular MRI, where prior coil sensitivity information is not required to implement Parallel MRI (19–21).

GRAPPA (Generalized Auto-calibrating Partially Parallel Acquisitions) (14) is a k -space method for Parallel MRI image reconstruction where the linear combinations of the coil sensitivity variations of the receiver coil elements (in the receiver array) are used to estimate some of the gradient (or phase) encoding steps (14) thus allowing to skip some of the phase encoding steps in the MRI data acquisition. A weighted linear combination of the acquired points (source points) is used to generate the missing k -space points (the target points) (14), as shown by the following equation:

Figure 4 The phased array receiver coil system.

Figure 5 The designed uniform vascular phantom with tubes of four different diameters (outer diameters: 6 mm, 8 mm, 10 mm, 12 mm).

Figure 6 The vascular phantom with different geometric features, used to compare the quality of MR images using different reconstruction algorithms (Top) Schematic Diagram, (Bottom) Phantom.

$$\vec{S}(k_y + m\Delta k_y) \approx \hat{W}_m \cdot \vec{S}(K_y)$$

Here $\vec{S}(k_y)$ represents the signal in k -space at the location k_y and $\vec{S}(k_y + m\Delta k_y)$ shows the signal located at $k_y + m\Delta k_y$. The elements of the matrix $\hat{W}_m$ contain the GRAPPA weight set for each of the coil combinations, where the subscript m signifies that different weight sets are required for each missing k -space line (14).

MATERIALS AND METHODS

We have developed a phased array MRI receiver coil system which consists of two coils: (1) simple loop coil, (2) inverse solenoid coil. Both coils are mounted on a catheter to be used for intravascular imaging. The tuning and matching is performed remotely because of limited space on the catheter. A $\lambda/2$ (1.613 m) length of micro-coaxial cable (Nexans, S.A., Paris, France) is used to connect the tuning and matching circuit to the receiver coils.

The Quality Factor (Q-Factor) is used to quantify how well tuned and matched the coils are. The coil system is encased in a polymer shrink tubing (Fig. 2).

The coils are mounted on a catheter. A 50 Ω thin coaxial cable is used to connect the receiver coils to the tuning and matching circuit, which connects to the impedance analyzer/MRI scanner via a BNC cable as shown in Fig. 3. The phased array coil system is shown in Fig. 4.

Two vascular phantoms are designed for the imaging experiments. The first phantom is a uniform vascular phantom made from Perspex as shown in Fig. 5. The phantom has four tubes of different diameters (outer diameters of the tubes are: 6 mm, 8 mm, 10 mm,

Figure 7 Comparison of the coil sensitivity in Axial direction (T_2 weighted Fast Spin Echo sequence): (a) Simple Loop Coil (b) Inverse Solenoid Coil.

Figure 8 Comparison of the coil sensitivity in Coronal direction (T_2 weighted Fast Spin Echo sequence): (a) Simple Loop Coil (b) Inverse Solenoid Coil.

12 mm) and can accommodate the receiver coils of varying diameters. The phantom has been filled with copper sulphate solution of 4 mM concentration. This phantom is used to estimate the sensitivity profiles of the interventional receiver coils.

The second phantom is designed with some geometric features in it so that the quality of the reconstructed image (using different reconstruction algorithms) can be evaluated. The phantom is built in the form of a Perspex cylinder of 60 mm outer diameter. A tube of 10 mm diameter passes through the centre of the phantom (to hold the receiver coil) and protrudes out at both ends. Two Perspex tubes of different diameters (4 mm and 6 mm) are attached to the central 10 mm tube. The

4 mm tube contains Copper Sulphate solution of 10 mM concentration and is sealed at both ends, while the 6 mm tube contains the Copper Sulphate solution of 1 mM concentration. The different concentrations of the solutions in the 4 mm and 6 mm tubes are aimed at providing different T_1 and T_2 relaxation times thus resulting in different contrasts in the MR images. Two nylon screws with 1 mm pitch and 2 mm pitch (M6 and M10, respectively) are also attached to the central 10 mm tube in the axial as well as longitudinal directions. These screws act as resolution bars and are placed to quantify the resolution capability of the different Parallel MRI reconstruction algorithms. The 60-mm cylinder is filled with the 4 mM Copper Sulphate

Figure 9 Cartesian GRAPPA reconstruction, each image obtained after the sum-of-squares reconstruction of the individually reconstructed coil images: (a) AF 2, AP = 0.069, (b) AF 3, AP = 0.552.

Figure 10 Standard Radial GRAPPA reconstructions for different acceleration factors (AF): (a) AF = 2, AP = 0.014, (b) AF = 8, AP = 0.024.

solution. The central 10-mm tube along with all the associated structures is placed inside the 60-mm cylinder and the system is sealed with Perspex glue. The phantom is shown in Fig 6.

The MRI data is acquired using GE MR750, 1.5T MRI scanner at St. Mary's Hospital London using Fast Spin Echo sequence with the following parameters: Slice Thickness 3 mm, Matrix Size 256×256 , Flip Angle 90° , TR 520 ms, TE 15 ms, FOV 55 mm. First, a fully sampled dataset is acquired in Cartesian space and then converted to the Radial trajectory space using the modified version of the Matlab code provided by Nicole Seiberlich at ESMRMB Parallel MRI workshop in Wurzburg, Germany in 2010 (22).

Figure 11 A plot of the artifact powers (%) for different acceleration factors using Standard Radial GRAPPA.

RESULTS

In this paper Q-factor changes from 14 in the unloaded case to 13.69 in the loaded case for the simple loop coil, which is similar to a value reported by Homagk et al. (23). The inverse solenoid coil has a Q-factor of 15 in the unloaded case and 15.2 in the loaded case, indicating a good tuning and matching condition. The cross coupling S21 of the coil system is -29 dB at 63.86 MHz (the tuning frequency for 1.5 T) which indicates a good decoupling.

Figures 7 and 8 show the fully sampled axial and coronal slices through the uniform vascular phantom obtained using the intravascular phased array coil. The images show the difference between spatial sensitivity of both the coils which helps to achieve better reconstruction results in Parallel MRI (13).

Cartesian and Radial GRAPPA (12) reconstructions are performed for different acceleration factors. The reconstruction quality is evaluated using artifact power

Table 1 Artefact Power (AP) for Different Reconstruction Algorithms and Acceleration Factors (for Uniform Vascular Phantom Images)

Acceleration Factor	Cartesian GRAPPA	Radial GRAPPA	Radial GRAPPA
		(Still Segment)	(Moving Segment)
2	0.069	0.014	0.009
4	–	0.019	0.009
8	–	0.024	0.0093

Figure 12 Moving Segment Radial GRAPPA reconstructions for different acceleration factors (AF): (a) AF = 2, AP = 0.009 (b) AF = 8, AP = 0.0093.

(AP). AP is a measure of accuracy and calculates the “Square Difference Error” between the reconstructed image and the reference image (24). Figure 9 shows the reconstruction results using Cartesian GRAPPA for acceleration factors 2 and 3. The results indicate that Cartesian GRAPPA provides a good reconstruction for AF = 2 with a low AP of 0.069 ($\sim 7\%$ error) but fails to provide a good reconstruction for AF=3 (artifact power 0.552).

The reconstructed images using Standard Radial GRAPPA (25,26) for different acceleration factors are shown in Fig. 10. A visual inspection of these images indicates the deteriorating quality of the reconstructed images at higher acceleration factors. Figure 11 shows a plot of the Artifact Power (AP) in

the reconstructed images for the uniform vascular phantom using Standard Radial GRAPPA. As the acceleration factor increases, the artifact power increases as well, thus indicating a greater reconstruction error for higher acceleration factors e.g., the AP for acceleration factor 2 is only 1.4% but this increases to 3.7% for an acceleration factor 16 in this case (Table 1).

Figure 12 shows the reconstruction results for different acceleration factors using Moving Segment Radial GRAPPA (25,26). The reconstructed images show a smaller artifact power even at higher acceleration factors. A plot of the AP vs. acceleration factors for Moving Segment Radial GRAPPA is shown in Fig. 13. The plot indicates an almost consistent reconstruction

Figure 13 A plot of the artifact power vs. acceleration factors for Moving Segment Radial GRAPPA.

Figure 14 A plot of the SNR vs. acceleration factors using Moving Segment Radial GRAPPA.

Figure 15 Cartesian GRAPPA reconstruction, each image obtained after the sum-of-squares reconstruction of the individually reconstructed coil images: (a) Acceleration Factor 2, $AP = 0.256$, (b) Acceleration Factor 3, $AP = 0.439$.

quality with an AP of 0.9%. There is a decrease in SNR in Moving Segment Radial GRAPPA as acceleration factor gets higher e.g., the SNR decreases by 5% in this example when the AF changes from 2 to 16 in Moving Segment Radial GRAPPA and this is shown in Fig. 14.

Figure 15 shows the reconstruction performed for Cartesian GRAPPA using the under-sampled data of the geometric vascular phantom. Cartesian GRAPPA for $AF = 2$ provides apparently noisy reconstruction in this case with an error of 25.6% ($AP = 0.256$) but the reconstruction results are even worse for $AF = 3$ with 43.9% error ($AP = 0.439$).

The reconstruction results using Moving Segment Radial GRAPPA for the geometric vascular phantom are shown in Fig. 16. The images show an error of only 2.26% and 2.32% for the acceleration factors of 2 and 8, respectively. The 1 mm threads of the Nylon screw used as the resolution object are also visible in the reconstructed images.

DISCUSSION

The reconstruction results using Cartesian and Radial GRAPPA have been presented above in Section 4. The

Figure 16 Moving Segment Radial GRAPPA reconstructions for different acceleration factors, each image obtained after the sum-of-squares reconstruction of the individually reconstructed coil images: (a) $AF = 2$, $AP = 0.0226$ (b) $AF = 8$, $AP = 0.0232$.

AP (24) is used to quantify the quality of reconstruction. Two objects (uniform vascular phantom and a geometric vascular phantom) are used in the imaging experiments. Cartesian GRAPPA seems to provide acceptable reconstruction results for acceleration Factor 2 but the reconstruction error (AP) becomes considerably higher for $AF = 3$ i.e., an error up to 43.9% (Fig. 9). It is important to note that the receiver coil system used in these experiments consists of only two coils (a simple loop coil and an inverse solenoid coil). The theoretical limit of acceleration for standard Cartesian reconstruction in Parallel MRI as described by Preussmann (13) is equal to the number of the coil elements in a parallel receiver array if each coil element has a distinct sensitivity profile (two coils in this case). Furthermore, the coil sensitivities shown here are different for both the coil elements as shown by the fully sampled images in Figs. 7 and 8 yet not necessarily fully independent. This demonstrates that the phased array coil can provide acceptable reconstruction results for $AF = 2$ but not for $AF = 3$.

The results also indicate that Radial GRAPPA helps in achieving higher acceleration factors as compared to Cartesian GRAPPA. This is because of an oversampling of the centre of k -space even at higher acceleration factors and the central part of the k -space contains most of the image information. However the Moving Segment Radial GRAPPA provides more consistent and better reconstruction results with minimum AP even at higher acceleration factors as compared with Standard Radial GRAPPA e.g., the maximum artifact power for Moving Segment Radial GRAPPA is only 2.32% for an acceleration Factor 8. In Moving Segment Radial GRAPPA a new segment is defined centred at each missing point in the under-sampled k -space and the corresponding weight set is applied to estimate the missing k -space signal (25). In this way, the segments here more accurately reflect the acceleration distance thus improving the quality of reconstruction as compared to Standard Radial GRAPPA where all the missing points in a segment are estimated using the same segment.

Moving Segment Radial GRAPPA is computationally expensive because a new segment is to be defined and weights to be calculated for each missing k -space point (26) in the under-sampled data. For a 256×256 image with an acceleration factor of 2, Standard Radial GRAPPA takes approximately 4 seconds on an Intel(R) Core™ 2 Duo, 2.66 GHz system having 3.48 GB RAM, but Moving Segment Radial GRAPPA takes ~24 s. Moving Segment Radial GRAPPA performs better but it is computationally more expensive (six times more expensive in this case) as compared to Standard Radial GRAPPA.

Some improvements in the design of the coils can improve the reconstruction results. The micro-coaxial cable connecting the tuning and matching circuit to the receiver coils may cause signal loss and thus limits the maximum achievable SNR. A better alternative will be to have the tuning and matching circuit integrated on the coil itself. This will minimize the signal losses and will improve the SNR. Although the current design of the coil system does not allow the capacitors to be placed on the coil or catheter, however, the use of miniature capacitors may help to achieve this in a future design.

One major safety concern regarding the use of active imaging catheters is the possibility of unwanted RF coupling which may lead to thermal injury owing to local RF heating effects (27,28). Many techniques have been proposed to avoid RF induced heating (29,30) and this aspect is beyond the scope of the current study. Future work involving clinical experiments may also require detailed investigation of the RF heating effects.

CONCLUSIONS

A new catheter-based phased array receiver coil is developed. The coil is composed of two coil elements (a simple loop coil and an inverse solenoid coil) and is capable of implementing Parallel MRI. The Q -factor calculations show good tuning and matching for both the coils in the unloaded as well as loaded conditions. The phased array MR receiver coil has been tested using different phantom objects. The imaging experiments are performed with 1.5 T GE scanner. Three different Parallel MRI algorithms (Cartesian GRAPPA, Standard Radial GRAPPA and Moving Segment Radial GRAPPA) are applied on the under-sampled data for different acceleration factors. The quality of the reconstructed images is quantified by AP. The results show that Moving Segment Radial GRAPPA provides better results with minimum artifacts as compared with the other two methods for the proposed phased array intravascular receiver coil.

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BIOGRAPHIES

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