Image Encoding and Reconstruction with Wearable MRI Hardware

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Motivation

MRI is a powerful imaging modality capable of resolving high resolution, diagnostically relevant images from a non-invasive data acquisition procedure. Conventional MRI is performed on high field systems (1-7 Tesla) which cost millions of dollars. Recently there has been a push towards low-field MRI scanners with hopes of improving accessibility and enabling bed-side scanning [2].

While imaging at lower field strengths is far more cost effective and accessible, **the low field regime is fundamentally limited in SNR. As a consequence of imaging at low SNR, temporal averaging must be done in order to achieve reasonable image quality, leading to extremely long scan times**. These long scan times require subjects to remain immobile throughout the scan (>30min), which is infeasible for several populations.

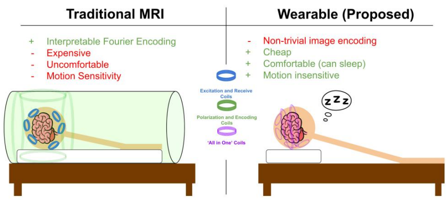

Figure 1. Comparison of a conventional MRI system (left) with the proposed wearable system (right).

Inspired by [1], **our proposed solution is to build a comfortable wearable scanner which can perform continuous imaging during the human sleep cycle**. Since the hardware is wearable, the subject is free to move as the wish, which allows for a comfortable way to perform temporal averaging. Wearable encoding coils produce non-linear magnetic fields, which requires an entirely new MRI forward model. **We will explore how feasible it is to reconstruct images using a wearable MRI system in the presence of noise and non-linear encoding fields**.

Image Encoding Model

MRI Physics

Suppose that K loop coils are placed on the subject's head at several known locations, orientations, and diameters. Each coil is used both for generating encoding magnetic fields and receiving data. If one were to apply currents $I_1, ..., I_K$ to each coil, the magnetic field induced by all of these coils $\mathbf{B}(\mathbf{r}) : \mathbb{R}^3 \to \mathbb{R}^3$, which is varying over spatial coordinates $\mathbf{r}$, is simply a linear combination of the unit field response $\mathbf{B}_k(\mathbf{r}) : \mathbb{R}^3 \to \mathbb{R}^3$ per coil

$$\mathbf{B}(\mathbf{r}) = \sum_{k=1}^{K} I_k \mathbf{B}_k(\mathbf{r}). \tag{1}$$

With some simplifying assumptions about a pre-polarizing field, and using Faraday's law, the received temporal signal $s_k(t)$ for the k^{th} coil can be well modeled as

$$s_k(t) = -\frac{\partial}{\partial t}\int_{\mathbf{r}} \mathbf{B}_k(\mathbf{r}) \cdot \left(\mathbf{R}_{\hat{\mathbf{B}}(\mathbf{r})}(\gamma||\mathbf{B}(\mathbf{r})||_2 t)\mathbf{m}_0(\mathbf{r})\right) d\mathbf{r}, \tag{2}$$

where $\mathbf{R}_{\mathbf{v}}(\theta) \in \mathbb{R}^{3\times 3}$ describes a rotation about $\mathbf{v}$ by angle θ and $\mathbf{m}_0(\mathbf{r}) : \mathbb{R}^3 \to \mathbb{R}^3$ is the initial magnetization that we would like to estimate.

Linear System Formulation

Upon vectorizing the multi-coil received temporal signals $s_k(t) \to \mathbf{b} \in \mathbb{C}^M$ and the desired magnetization vectors $\mathbf{m}_0(\mathbf{r}) \to \mathbf{m} \in \mathbb{C}^N$, we can combine (1) and (2) to arrive at a much simpler linear system, where the forward model depends on the applied currents $I_1, \cdots, I_K$:

$$\mathbf{A}(I_1, \cdots, I_K)\mathbf{m} = \mathbf{b} + \mathbf{n}, \tag{3}$$

where $\mathbf{A} \in \mathbb{C}^{M\times N}$ is the discretized forward model described in (2), and $\mathbf{n} \sim \mathcal{N}(0, \sigma^2 I)$ is Gaussian noise induced by resistance in the coils and subject. Since we are analyzing the analytic signal in each coil via IQ demodulation, the received data is complex. A visual representation of $\mathbf{A}$ for some fixed set of currents is shown in Figure 2.

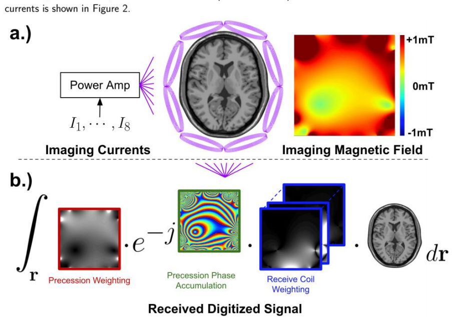

Figure 2. a.) Currents $I_1, \cdots, I_8$ are applied to their respective coils to generate an imaging magnetic field (right). b.) The digitized signal induced on each coil is given by a linear function of the image. The encoding functions are dependent on the imaging magnetic field, and hence they are also dependent on the currents $I_1, \cdots, I_8$. The same set of $K = 8$ coils are used for encoding and receiving.

Image Reconstruction

In order to have sufficient encoding to invert (3), and battle the relatively large levels of Gaussian noise at low field, we can repeat our imaging experiment with a set of N currents $\{(I_1^{(i)}, \cdots, I_K^{(i)}\}_{i=1}^N$, yielding N measurement vectors $\{\mathbf{b}^{(i)}\}_{i=1}^N$, and solve the following regularized linear inverse problem

$$\min_{\mathbf{m}} || \underbrace{\begin{bmatrix} \mathbf{A}(I_1^{(1)}, \cdots I_K^{(1)}) \\ \vdots \\ \mathbf{A}(I_1^{(N)}, \cdots I_K^{(N)}) \end{bmatrix}}_{\mathbf{A}} \mathbf{m} - \begin{bmatrix} \mathbf{b}^{(1)} \\ \vdots \\ \mathbf{b}^{(N)} \end{bmatrix} ||_2^2 + \mathcal{R}(\mathbf{m}). \tag{4}$$

In our experiments, we solve (4) with no regularization (Conjugate Gradient), and several denoising regularizers using ADMM. In all experiments we choose the currents $I_k^{(i)} \sim \mathcal{U}(-I_{\max}, I_{\max})$. The received signal is acquired over a 2ms period with a sampling time of 4μs. We repeat this over $N = 200$ measurements.

Results

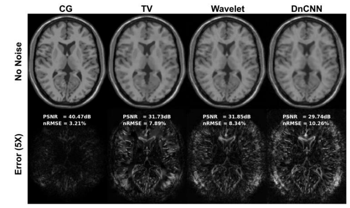

Figure 3. Different regularized and unregularized reconstructions without any noise

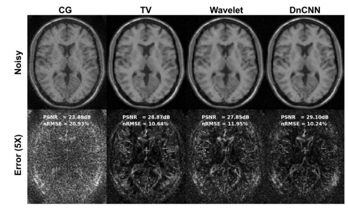

Figure 4. Different regularized and unregularized reconstructions without moderate Gaussian Noise

Iterative reconstructions are performed with (Figure 3) and without (Figure 4) Gaussian noise. Conjugate Gradient (CG) reconstructions were run for 500 iterations, and an ℓ_2 regularization was used to combat noise amplification. The Total Variation (TV), Wavelet, and Deep Learning based denoising (DnCNN) priors are enforced using 100 iterations of the ADMM algorithm with optimized hyper-parameters.

Conclusion

In this work we show how a naive strategy of controlling comfortable, non-linear, MRI coils can produce images with reasonable quality. Various forms of regularization are shown to assist with the problematic noise experienced in low field systems.

References

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