FEASIBILITY OF 3D MODEL CONSTRUCTION IN MRI COLONOSCOPY USING A MICRO-COIL ANTENNA AND THE PULL-BACK METHOD

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Abstract: It is possible to incorporate a pull-back concept from intravascular ultrasound (IVUS) imaging into internal MR imaging techniques. The method stacks 2D images by using plane position and orientation information to create a 3D volume. This information is accessible in MRI by the use of micro-coils and tracking software. Data acquired from an inverse Helmholtz coil that can be mounted on a colonoscope is presented here and volume creation using micro-coil tracking and the pull-back method is demonstrated.

Introduction

Colorectal cancer causes the third highest mortality of all cancer deaths in both men and women in the UK [1]. Colonoscopy procedures are used for detection of cancer, inflamed tissue, abnormal growths, and ulcers in the colon and rectum. Conventional colonoscopy relies on the relay of an optical image from the tip of the instrument, the information provided is therefore purely superficial. While virtual colonoscopy methods exist using computed tomography (CT), they are not interventional methods as such. Using MRI to supplement colonoscopy may lead to improved diagnosis, earlier detections for example, as well as an enhanced ability to apply and evaluate the requisite therapy because the imaging can provide detail of structures in and beyond the colon wall.

MR colonoscopy can be performed with a micro-coil used to augment conventional MRI through image-combination techniques. Improved signal to noise ratios can be achieved at localised regions of interest during minimally invasive procedures [2]. That is, images acquired internally are combined with those acquired with the conventional external coil.

An inverse Helmholtz imaging coil and active tracking coil have been fitted on a MR compatible colonoscope for this study. The imaging micro-coil is well suited for high-sensitivity 2D applications that do not require superior SNR for penetration depths greater than about 4.5 coil radii [2].

However, the coil does not inherently lend itself to imaging in 3D. To overcome this, it is demonstrated that a 3D model of the subject can be constructed by stacking the 2D images taken along the path traversed, perpendicular to the tip of the scope, provided that the orientation and location of each plane is known.

A similar construction method has been applied to intravenous 2D ultrasound imaging [3,4]. In these cases, the images were automatically acquired using a catheter mounted ultrasonic transducer. A 3D volume can be created by effectively stacking the ultrasound images, which are disk-like and perpendicular to the catheter. This achieves an improved method of viewing atherosclerotic plaques. However, there are factors that impede a real-time implementation of this method. Specifically, knowledge of the plane location, orientation, and the path traversed is constrained in IVUS and can lead to error in the constructed 3D volume [5]. Such information must be derived with the use of pull-back rate control or measurement, or by fusion with biplane angiography [6].

Geometrical similarities between IVUS and MR colonoscopy make the pull-back method a good candidate for creating a 3D model from the inverse Helmholtz coil data. Unlike IVUS, position tracking in MRI can be realised in a fast and robust manner by using active MR tracking coils [7]. With the availability of imaging plane data, the application of the pull-back method to procedures involving internal MR coil imaging is simplified and may be feasible in real-time.

Materials and Methods

The pull-back method in MRI is significantly dependant on the ability to simultaneously image and track the location and orientation of the imaged planes. Imaging was successfully performed with an inverse Helmholtz coil previously used in MR colonoscopy studies [2]. Tracking was performed with a 15 turn, 32 gauge tracking coil with an outside diameter of 3.0 mm. The coil was internally loaded with vegetable oil for increased tracking SNR. Both coils were tuned to 21.28 MHz, which is the resonant frequency of hydrogen at the main field strength, 0.5 T. While the imaging coil is situated on the outer surface of the colonoscope, the tracking coil is situated in the distal tip of the working channel, and thereby located within the sensitivity volume of the imaging coil, Figure 1. This configuration was chosen in order to decrease the overall cross-section penalty incurred from the additional hardware attached to the colonoscope.

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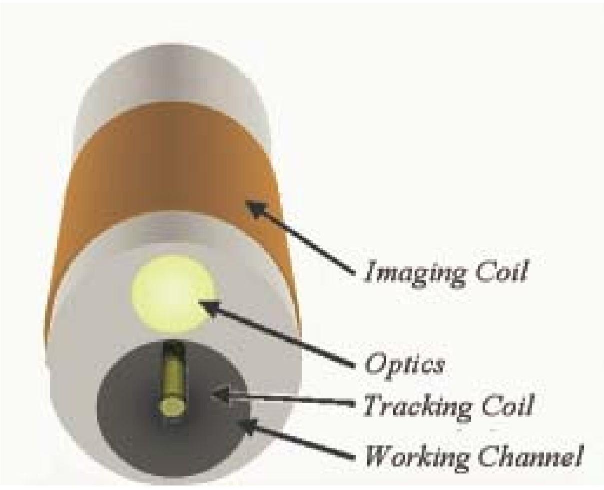

Figure 1: Schematic of the MR compatible colonoscope with fitted imaging coil and tracking coil.

For testing purposes, the coil was mounted on a Perspex rod with outer and inner dimensions similar to that of the colonoscope. The synthetic colonoscope will be referred to as *the colonoscope* throughout the paper for readability.

The test phantom consisted of two concentric cylinders with a primarily gelatine solution between them, giving it a uniform MR signal. It contained a small orange in the gelatine filled cavity between the cylinders. The intention was to mimic a structure within the colon wall.

The diameter of the inner cylinder of the phantom is 19.0 mm, while the outside diameter of the imaging coil is 17.0 mm. This leaves enough clearance to accommodate movement of the colonoscope, yet allows for significant signal penetration. The sensitivity range of the coil is approximately 4.5 radii, or 38 mm.

A linear 10 cm path was traversed, starting below and ending above the test target. Fast spin echo (FSE) sequencing was used during imaging. A standard GE birdcage body coil was used for RF transmission and the colonoscope coil for reception. For comparison, the phantom was also imaged using both the body coil and the head coil independently.

The experiment was performed on a Signa SP/i MRI scanner (GE Healthcare, Milwaukee, WI), equipped with the potential to perform tracking. The system allows for the tracking of up to four coils, and for the automatic selection of the scan plane such that it contains up to three of them. This property can be exploited to select a slice that is in the current prescribed orientation (such as sagittal) and contains the first tracking coil. Additionally, a second coil can be used to define the normal of the desired plane that contains the first coil. Figure 2 shows the use of singular and dual coil tracking.

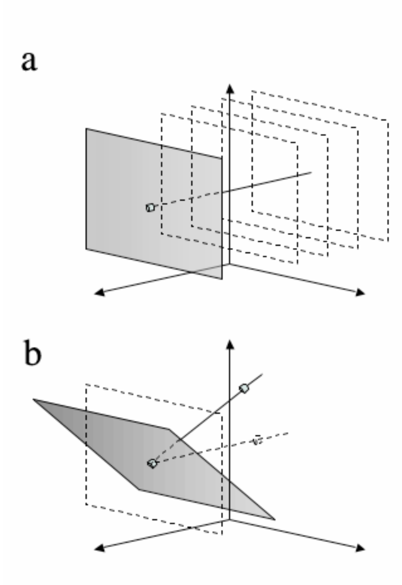

Figure 2: a) Plane definition with one coil allows for transverse tracking. b) Slice definition with two coils allows for plane rotations, where the plane normal is derived by the out of plane, second coil.

Future work will introduce a second tracking coil into the working channel of the colonoscope, so that changes in orientation will be detected and used for prescribing the subsequent excitation plane. At present, the slice selection was based on a single coil. The slice plane was chosen such that it was oriented perpendicular to the colonoscope and located at the tracking coil position, at the distal tip of the scope, along the gradient axis. While the colonoscope orientation was not incorporated into this pull-back study, the use of two tracking coils was investigated during stationary imaging.

Results

Images obtained from a standard birdcage body coil and from the colonoscope coil are shown in Figure 3a) and Figure 3b) respectively. Bilateral filtering and brightness correction have been applied to the colonoscope images. The brightness correction algorithm accounts for the fall off of signal from the antenna, while bilateral filtering reduces noise in the image. The slice suffers from some ringing and ghosting artefact.

In both images, a cross section of the orange in the phantom is visible. The size and location of the orange is comparable between the images, however there is a higher SNR in the colonoscope slice, making the segments more clear.

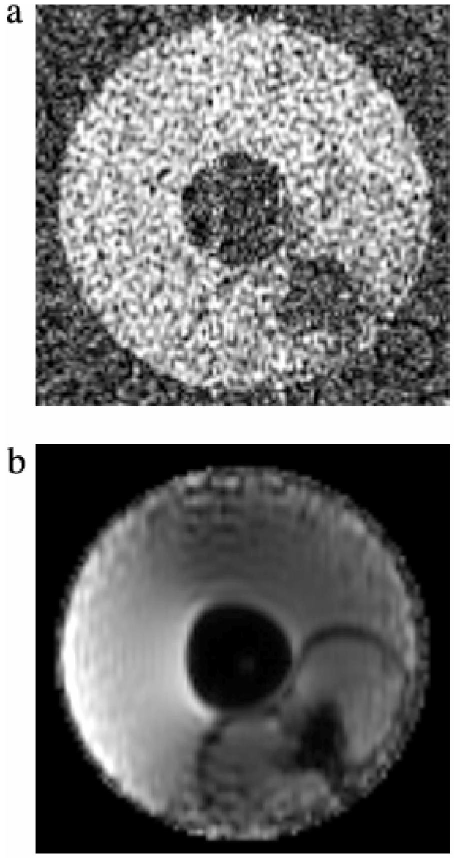

Figure 3: a) Example FSE slice of the phantom from the body coil. b) Example FSE slice of the phantom from the colonoscope coil.

A stack of 2D images with a spacing of 2.5 mm was created from the colonoscope data by the application of spline interpolation between the unequally spaced slices. 3D surfaces were created from the stack by using the rendering capability of the OsiriX v. 1.7.1 software **[8]**, Figure 4a) and 4b).

Manual segmentation was performed on the colonoscope data to separate the noise from the orange structure in the 3D model. Figure 4b) shows the orange volume rendered separately, based on the segmented images, and superimposed on the phantom data for illustrative purposes.

Figure 4c) shows the 3D volume created by the head coil data, which is essentially a gold standard for the phantom reconstruction, but this type of coil is not geometrically viable for colonoscopy. The orange volume is easily identifiable due to the lack of noise when imaging with the head coil. The test target in each phantom reconstruction have similarities in that some landmarks can be identified in each, such as the inner core, the surface, and the segmented internal structure.

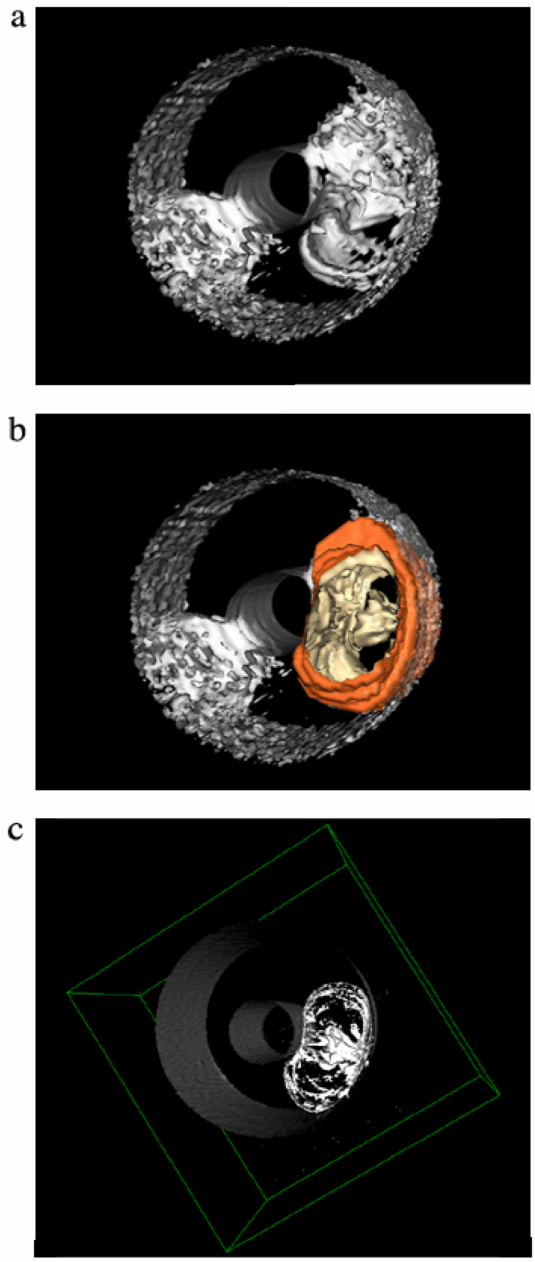

Figure 4: a) Pull-back volume from the colonoscope data. b) The orange in the phantom is highlighted by segmentation. c) Volume created from the head coil data.

Discussion

The data that comprise the 3D models in Figures 4a) and 4b) were acquired with the pull-back method, thus incorporating a concept used in IVUS imaging. However, unlike US imaging, MRI lends itself to rapid image plane determination, based on the location of tracking coils. The pull-back method described in this paper is along a linear path, with uneven spacing between images, and was also performed at a varied rate.

The processing time for the phantom reconstruction was minimal, as the images only required brightness

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correction, filtering and some interpolation before being viewed with a common medical data viewing software package, OsiriX.

Further segmentation was shown to produce a 3D volume with improved visual clarity.

Future experimentation should employ additional reconstruction computation and tracking coil hardware in order to apply the method to non-linear path traversals. Also, the volume building time between 3D MR acquisition and the pull-back method should be compared. Regardless of subject reconstruction time, this method demonstrates that 3D volumes can be created using internal coils, which may not be suitable for traditional 3D acquisition MRI, due to their localised sensitivity.

Clinically, this method may provide a way to visualise elongated, cylindrical physiological structures that are suitable to image from the centre in a radial direction, such as the bowels or arteries.

Conclusion

The pull-back method is highly applicable to internal MRI imaging with coils that image perpendicular to the direction of motion, such as this case of an inverse Helmholtz coil on a colonoscope. MR imaging planes can be determined based on the location and orientation of tracking coils, a feature which modalities such as IVUS and CT lack. This provides enough information to perform the pull-back method in a way unique to MRI. The method presented here is an effective means of creating a 3D volume during internal MR imaging procedures.

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