Model-based Catheter Shape Reconstruction for Interventional MRI

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Abstract. In interventional MR, 3D visualization of catheters is hampered by low spatial and temporal resolution as conventional image reconstruction requires data sampling at or above the Nyquist rate.
We propose a model-based reconstruction technique to fit a parameterised model of a catheter to acquired MRI data by minimising the ℓ_2-norm in Fourier space. A modelled image of the catheter shape is transformed to the Fourier space and then registered to the acquired data. Thereby the number of independent degrees of freedom to be solved for in reconstruction is reduced. Using computer simulations and phantom data it is demonstrated that the catheter shape can be reconstructed from highly undersampled data indicating the potential of the method for 3D imaging of catheters.

Keywords: Interventional MRI, catheter imaging, model-based reconstruction, curve fitting

1 Introduction

Image guided interventions are an important technique for treatment of a variety of cardiovascular diseases and interventional devices like catheters are visualized using mainly X-ray fluoroscopy. The use of real-time MRI for guidance of intravascular interventions is a rapidly developing research field given its superior soft-tissue contrast and the lack of hazardous radiation. In a recent study, the feasibility of MR-guided cardiac catheterization in a series of patients was demonstrated for the first time [1]. However, for a widespread use of MR-guided interventions, reliable and high-contrast visualization of the catheter is an indispensable prerequisite. In the past, a wide range of techniques has been proposed for these purposes; they are usually classified into active and passive techniques. Active techniques use locally sensitive receive coils for localization [2, 3] or for visualization [4-6].

Passive techniques visualize the catheter either by including paramagnetic materials [7-9], by including contrast agents into the catheter [10] or filling the catheter with substances at a different resonance frequency such as F-19 [11]. The latter technique has the advantage of providing the functionality of an active catheter without any of the safety concerns associated with radio-frequency heating and mechanical modifications.

In order to visualize the entire length of such a catheter, a three dimensional (3D) image of a complete volume is necessary. Unfortunately, temporal resolution of conventional MRI is poor hampering 3D display of catheters at high update rates.

Recently, a method for highly accelerated 3D imaging of active catheters was proposed employing Compressed Sensing to sample far below the Nyquist rate [12]. The framework is a two-stage approach of accelerated MR imaging in conjunction with a curve fit to extract a parameterized curve description in 3D. However, frame rates of this technique remain moderate as the method relies on general imaging principles and does not consider the particular nature of the problem of reconstructing a curve in 3D. Recently, a reconstruction method for MR-perfusion imaging was proposed that incorporates a parametric model to improve the reconstruction fidelity from undersampled data [13]. This approach allows fitting of the model in Fourier domain, in which the MR-data is obtained.

In this work we propose a model-based reconstruction technique to fit a parameterised model of a catheter onto acquired MR-data by minimising the ℓ_2-norm in Fourier space. To this end, a modelled image of the catheter shape is transformed to the Fourier space and then registered to the acquired data. Thereby the number of independent degrees of freedom to be solved for in reconstruction is reduced. Using computer simulations and phantom data it is demonstrated that the catheter shape can be reconstructed from highly undersampled data indicating the potential of the method for 3D imaging of catheters.

2 Theory

Registration of two objects refers to optimizing a similarity function between the target object $y(t)$ and the transformed source object $\hat{x}(t)$, e.g. by minimising the ℓ_2-norm of the difference signal as a function of the parameter set $\boldsymbol{p}$ describing the transform:

$$\left| y(t) - \hat{x}_{\boldsymbol{p}}(t) \right|_2 \underset{\boldsymbol{p}}{\rightarrow} min \qquad \textbf{(1)}$$

It has been well-recognized that many transformations can be performed in the Fourier domain [14], i.e. a rigid translation in the image domain yields a phase shift in the Fourier domain while a rotation in the image domain results in rotation by the same angle in the Fourier domain. Parseval's theorem states that the ℓ_2-norm in both domains is identical:

$$\int_{-\infty}^{+\infty} |s(\tau)|^2\, d\tau = \int_{-\infty}^{+\infty} |S(f)|^2 df \qquad \textbf{(2)}$$

This means that the ℓ_2-norm in Eq. (1) may as well be solved in the Fourier domain.

2.1 Modelling Fourier domain representation of a wire

In MRI, data is acquired in the Fourier domain. Hence, the signal acquired at time point $\boldsymbol{t}$ serves as the target signal $\boldsymbol{y_t}$. In order to perform a least-squares fit, the representation of the source object in the Fourier domain has to be modelled (Fig. 1). The problem of fitting a parameterized curve in 3D to MR data can be written as

$$\left|\boldsymbol{WFI}\left(\boldsymbol{c}(\boldsymbol{p}_t)\right) - \boldsymbol{y}_t\right|_2 \underset{\boldsymbol{p_t}}{\rightarrow} min \qquad \textbf{(3)}$$

where $\boldsymbol{p}_t$ is a parameter vector describing the curve $\boldsymbol{c}$ at time point t, $\boldsymbol{I}$ is the image model of the given curve. The MR encoding process is modelled by applying the Fourier transform $\boldsymbol{F}$ and sampling pattern $\boldsymbol{W}$.

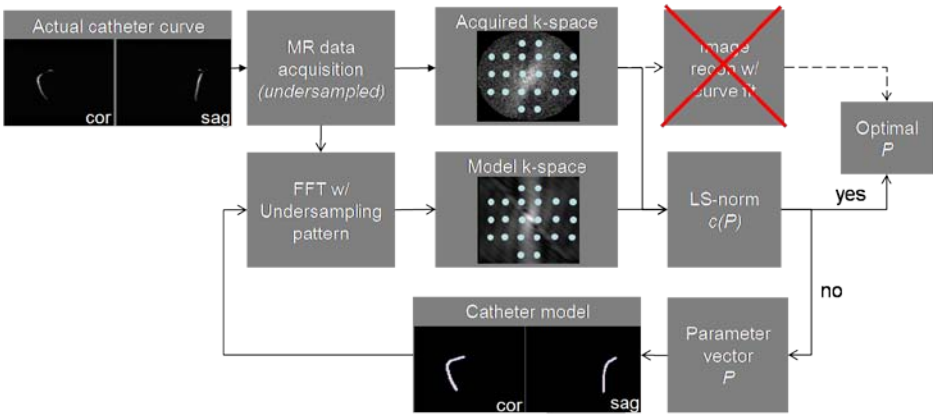


Fig. 1. Catheter modeling and fitting scheme. Instead of curve fitting after image reconstruction (crossed block), the parametric catheter model is fitted in k-space. From a set of parameters $\boldsymbol{p_t}$ at time point $\boldsymbol{t}$ a parameterized curve and a binary image of the catheter model is calculated. Subsequently, its Fourier domain representation is derived and its similarity to the acquired k-space data measured using the least-squares norm.

For curve description, Catmull-Rom splines were used as they are continuously differentiable, go exactly through the nodes and are easy to compute. As they are described by N nodes only, the parameter vector $\boldsymbol{p}$ is of length $(3 \cdot N)$.

An image representation of the curve is obtained by creating a binary image using the field-of-view (FOV) settings of the acquired data (Fig. 1).

2.2 Constrained fitting

The movement of the catheter is incremental and limited between time frames. Furthermore, the length of the wire is known. This allows for bounds of the search space for the parameter vector $\boldsymbol{p_t}$ by modifying Eq. (3) to

$$\left|\boldsymbol{WFI}\left(\boldsymbol{c}(\boldsymbol{p_{t-1}} + \Delta\boldsymbol{p_t})\right) - \boldsymbol{y}\right|_2 \underset{\Delta\boldsymbol{p_t}}{\longrightarrow} min$$

$$s.t. \quad \left|\frac{\text{arc length}\left(\mathbf{c}(\boldsymbol{p_{t-1}} + \Delta\boldsymbol{p_t})\right) - L}{L}\right| < \varepsilon, \tag{4}$$

$$|\Delta\boldsymbol{p_t}| < \mathrm{d}$$

where $\Delta\boldsymbol{p_t}$ is the incremental change relative to the last position, L is the known length of the curve and d the maximum displacement between time frames.

3 Materials and Methods

A phantom experiment was carried out to assess the proposed technique. Data processing was performed on a standard PC using MATLAB. Data were acquired on a 3T MR scanner (Philips, Best, The Netherlands) using a standard head coil to achieve a homogeneous sensitivity profile. As the system was not tuneable to non-proton imaging, a tubing of 14 cm length (inner ø 2.5 mm) was filled with Gd-DPTA-doted water to decrease T1-relaxation time. The tube was placed in open space to mimic selective fluorine imaging (Fig. 1 left). For data acquisition, a 3D balanced SSFP sequence (acquisition matrix 200x100x100mm^3, resolution 1x1x1mm^3, TR/TE/flip 2.8/1.36/45°) was acquired and undersampling of phase encodes was performed retrospectively.

3.1.1 Fit of a simplified model to complex data

A binary image of the acquired dataset was created by thresholding to obtain a simplified model having an accurate shape of the actual catheter. By translating and rotating this binary model (Fig. 2), the behaviour of the cost function regarding translation (± 10 mm) and rotation ($\pm 10°$) was assessed for equidistant undersampling with undersampling factors of 1, 91 and 134 equivalent to acquisition of 7840, 87 and 56 phase encoding steps (within a conventional elliptical k-space shutter).

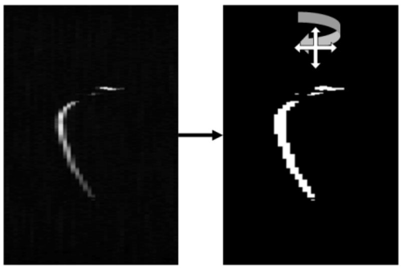


Fig. 2. Binary image from original image via thresholding (maximum intensity projection). The derived rigid model is then used to calculate the ℓ_2-norm for translation and rotation.

3.1.2 Fit of parameterized model onto complex data

The shape of the catheter in the acquired position was modelled using parameterised Catmull-Rom splines with four nodes equidistantly distributed along the curve. From this correct position the distal end of the catheter was bent by angling the last node away from its optimal position, hence modelling a catheter with a flipping head (Fig. 4. Fitting parameterized model to acquired complex data. The distal node was angulated away from its optimal position and the ℓ_2-norm calculated for angulations around the x- and y-axis. a,b) into two directions.

4 Results

Fig. 3 shows the ℓ_2-norm for rotation and translation in two directions between the model derived by thresholding and the complex catheter data itself for the applied undersampling. The cost function has a global minimum at the optimal position. Interestingly, Fig. 3(lower right) for translation in y direction shows a change of the cost function as the undersampling increases.
Fig. 4. Fitting parameterized model to acquired complex data. The distal node was angulated away from its optimal position and the ℓ_2-norm calculated for angulations around the x- and y-axis.
a,b shows the centerline of the parameterized catheter model with a bent head in relation to the actual acquired position (volume rendered). Fig. 4. Fitting parameterized model to acquired complex data. The distal node was angulated away from its optimal position and the ℓ_2-norm calculated for angulations around the x- and y-axis.
c shows the ℓ_2-norm for angulation around x- and y-direction, respectively, for full sampling while Fig. 5 shows the cost function for angulations around both directions and different undersampling factors. For all orientations and undersampling factors, the cost function shows a single global minimum.

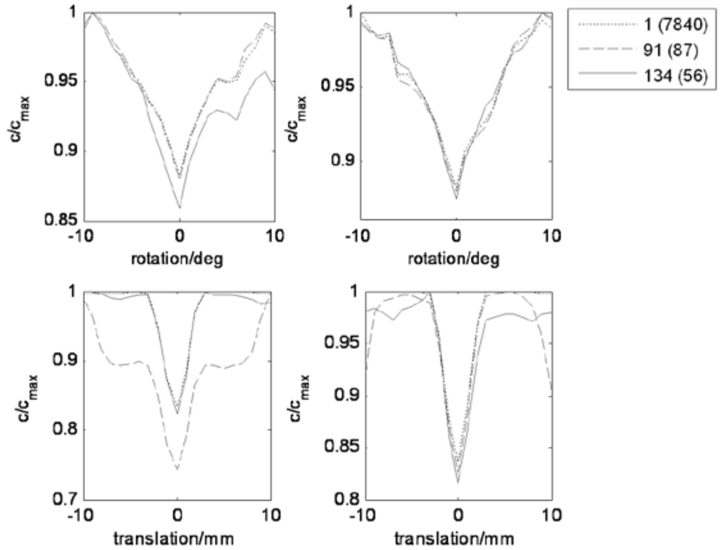
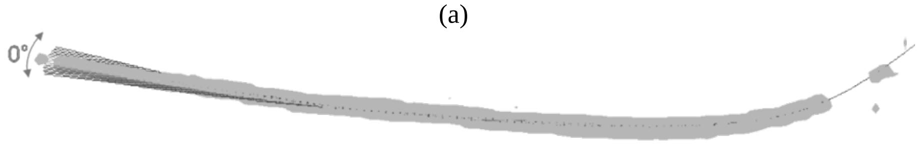
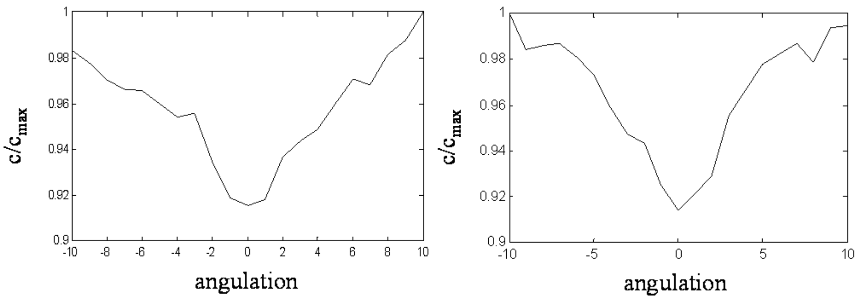


Fig. 3. Least-squares norm between the acquired data and the model gained through thresholding for translation and rotation around two major axes (left: X, right: Y).

(a)



(b)



(c)

Fig. 4. Fitting parameterized model to acquired complex data. The distal node was angulated away from its optimal position and the ℓ_2-norm calculated for angulations around the x- and y-axis.

5 Discussion

The simulations and phantom experiments prove the general feasibility of model based reconstruction for 3D catheter shape recovery from sparse MRI data. It was shown that a simple binary model of the catheter curve can be fitted onto acquired complex data in the Fourier domain thus allowing for high undersampling of k-space.
Of great importance is the choice of the applied undersampling pattern (see Fig. 3 lower right) to assure a continuously decreasing cost function towards the global minimum. Setting bounds for the search space based on length- and displacement-constraints may help to promote termination of the least-squares solver in the global minimum.

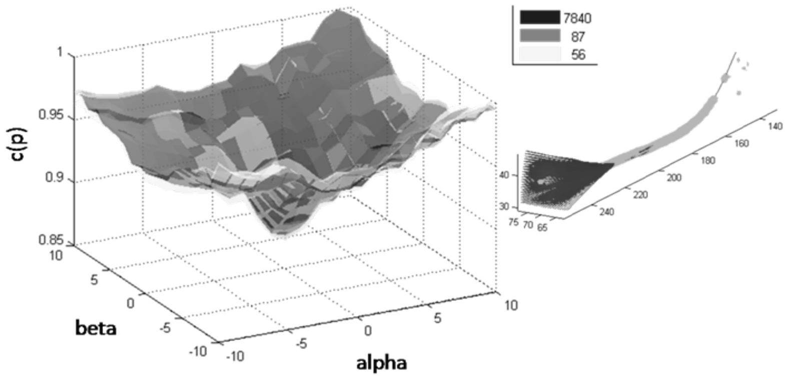


Fig. 5. Cost function for parameterized model bent around both major axes for different undersampling factors

In general, MR images are altered by the sensitivity profile $\boldsymbol{S}$ which is given by the orientation of the imaging coil and location of the acquired FOV. When imaging with an active catheter comprising a built-in receiver coil, $\boldsymbol{S}$ also depends on the wire geometry and has to be included in the model. In case of imaging of a non-proton based passive catheter, such as fluorine-filled catheters, the receiver coil is a standard imaging coil strapped on the patient and has a fixed sensitivity profile which can be derived prior to the intervention, thus rendering modelling of the catheter representation in the frequency domain easier.
Furthermore, the sampling scheme $\boldsymbol{W}$ should be chosen in a way that occurrence of eddy currents is reduced, e.g. by equidistant undersampling.

6 Conclusion

A method for model-based catheter shape reconstruction has been presented allowing for highly undersampled MR data acquisition. The simulations and experiments show the potential for passive catheter imaging in 3D at high spatial and temporal resolution and experiments with more complex catheter orientations will be carried out in the future.

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