Otto-von-Guericke-University Magdeburg

Faculty of Electrical Engineering and Information Technology

Chair for Electromagnetic Compatibility

Degree Thesis



Characterization and Automated Alignment Detection of an Additively Manufactured Z-frame Marker to Process Signals for Robotic Control in Interventional MRI

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By: Parisa Parsanejad

Born on June 23, 1991

In Esfahan, Iran

**Abstract**

**Task of the Thesis in the Origin:**

**Declaration by the candidate**

I hereby declare that this thesis is my own work and effort and that it has not been submitted anywhere for any award. Where other sources of information have been used, they have been marked.

The work has not been presented in the same or a similar form to any other testing authority and has not been made public.

Magdeburg, August 01, 2018

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1. **Introduction**

**1.1 Motivation**

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1. **Literature Survey**
   1. **Magnetic Resonance Imaging (MRI)**

MRI is a non-invasive imaging technology generating three dimensional anatomical and functional images of the body without the use of ionizing radiation. It is particularly useful for neurological, oncological, cardiovascular, muscular and skeletal imaging. MRI employs a powerful magnetic field that forces the hydrogen atoms in the tissue being imaged to align with its axis. The additional radiofrequency fields are then used to stimulate the protons and alter the alignment of the magnetization. When the radiofrequency fields are turned off, the protons realign with the magnetic field and release the detectable energy by MRI scanner. It defines different MRI parameters such as repetition time and echo time which are the time between two consecutive excitation pulses and the time between the excitation pulse and the recording of the magnetization value, respectively.

**2.1 Pulse sequences**

A pulse sequence is the measurement technique and contains multiple parameters such as RF pulses, gradient pulses, and timing. Depending on the anatomical region under observation, the optimal pulse sequence has to be chosen in order to acquire the data in the desired manner. There are three general criteria that should be considered when modifying the measurement parameters of a pulse sequence through the user interface software: acceptable scan time, adequate spatial resolution, and the sufficient contrast between tissue relative to the background noise (contrast-to-noise ratio). Many pulse sequence parameters are available commonly which can be categorized by their effect on the MRI image as intrinsic and extrinsic parameters. Intrinsic parameters influence only the signal-producing portion of the image such as patient anatomy. Extrinsic parameters affect factors external to the tissue and the structure of the data collection such as voxel size.

Despite the non-ionizing radiation characteristics, MRI imaging experiences different imaging artefacts which generate difficulties for the detection and registration task.

**2.2 Artifacts in MR imaging**

In the MR images, artifacts refer to pixels that do not truly represent the anatomy being studied. Artefacts can be categorized into three groups, according to the cause of the signal misregistration. The first group refers to artefacts that are produced by the patient tissue movement during data acquisition. The second group is a consequence of the measurement technique and parameters. The last group are independent of the patient or measurement technique and are generated from an external source to the patient or scanner.

**2.2.1 Motion artifacts**

The motion artifacts are caused by tissue that is excited at one location, producing signals that are mapped to a different location during detection. The specific appearance of the artifact depends on the nature of the motion and the measurement technique. If the motion is periodic, the produced artefact is discrete, often referred to as “ghosts”. Moreover, the sensitivity of a measurement to tissue motion depends on the amount of frequency and phase variation that occurs between successive echoes due to tissue movement. In abdominal or lumbar spine imaging, respiratory motion is the most common causes of the motion artefacts.

**2.2.2 Sequence/Protocol-related artifacts**

In this class of artifacts, the specific measurement parameters cause the artifact in the image. The source of these artefacts is relatively constant during the measurement.

**2.2.2.1 Aliasing**

The acquisition matrix and FOV in the specific measurement technique are two important factors determining a different phase and frequency to each location in the image. Aliasing artifact occurs both in the readout and phase encoding directions. If the selected FOV in the readout direction is smaller than the anatomical slice, the tissue’s frequencies pass the Nyquist limit, and are mapped to a lower frequency. This situation happens in the readout direction and is called high-frequency aliasing. It can be avoided by enhancing the number of readout data while maintaining the same sampling time. Furthermore, when tissue outside the FOV in the phase encoding direction is excited, the protons experience phase changes and are transformed to a lower phase situation via the Fourier transformation. This is known as phase encoding aliasing that can only be eliminated by increasing the FOV in the phase encoding direction.

**2.2.2.2 Chemical shift artifacts**

Chemical shift artifacts emanate from the inherent 3.5 ppm frequency difference between fat and water protons under the effect of an external magnetic field. One consequence of this frequency difference is a misregistration of fat and water protons from a voxel that are mapped to different pixels. In MRI, the exact magnetic field is dependent on the location inside the magnet according to the equation (1?); consequently, the detected signal from a proton (better voxel) is localized based on its frequency:

Bi=B0+GT ri

Where B0 is the main magnetic field, Bi is the magnetic field at location ri and GT is the total gradient amplitude.

Due to the fact that fat and water protons have a different molecular structure, they resonate at a lower frequency than water protons when exposed to the same gradient field. Accordingly, they will be mapped to a lower frequency pixel in the readout direction by the same GRO within a voxel. This misregistration is not perceivable in tissues with a consistent fat-water content. However, at the borders between tissues with an indicatively different fat-water content chemical shift artifact is visible. The number of pixels resembling chemical shift artifact (CSA) can be estimated according to the equation 2?:

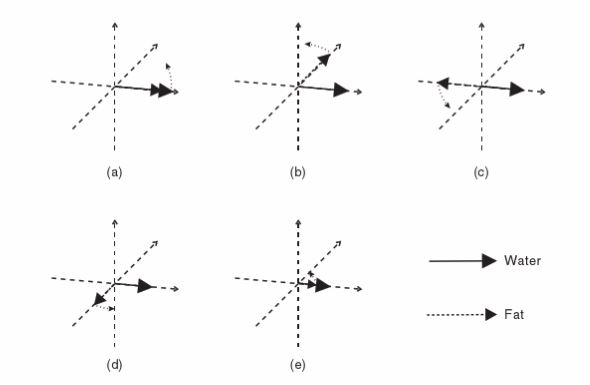
CAS =

Where represents the frequency difference between fat and water protons in a voxel, is the total receiver bandwidth, and shows the number of readout data points spanning the FOVRO.

**2.2.2.3 Phase cancellation artifact**

Phase cancellation artifact is the second type of chemical shift-based artifact which arise from the frequency difference between fat and water protons. This artifact is visualized in out-of-phase gradient echo images. As shown in Figure 2.1, the fat protons cycle in phase relative to the water proton precession at a rate linearly proportional to the measurement time after the initial RF excitation pulse. Combine the net search : http://www.revisemri.com/questions/artefacts/phase\_cancellation

**Figure 2.1** precession of fat and water protons. More explanation in book.



**2.1.0 Signal-to-Noise Ratio and Tradeoffs**

One of the most important characteristics of the MRI image data is the signal-to-noise ratio (SNR). The SNR in MRI images depends on the level of signal and the level of noise present in the data. There are different kinds of factors influence on the level of noise and signal in the MRI images. For instance, larger voxel size increases the SNR because the voxel contains more signal. Longer sampling time decreases the noise, and thus increases the SNR, the receive coil sensitivity and volume and the tissue. In addition, the receive coil volume and sensitivity contribute to the SNR. Furthermore, the tissue characteristics and its relaxation affect the SNR. These effects can be shown as follows:

SNR= V\*T1/2 \*R(B0, B1,…) \* Iseq (T1,T2,TE,TR,…) (1.1)

Where V is the voxel volume, T is the total sampling time for each voxel, R is a factor characterizing the effects of the main magnetic field, the receive coil sensitivity and so force, and Iseq is a factor characterizing the signal intensity form the pulse sequence and the tissue.

**Registration**

Registration is an important operation of surgical guidance and the accuracy of the guidance is based on the registration method. In fact, registration determines a geometrical transformation that aligns different views of an object. A “view” can be an image, such as CT or MR, or it can be the physical object itself. In this work, registration referred as “image-to-physical” registration, where

There are different types of registration. In this section

**Opposite registration approach**

The main strength of this registration system is the ability to achieve concurrent registration in desired target alignment from a single slice image. Conventional registration methods usually …. In robotically assisted surgery, the end-effector of a robot can be registered to the scanner using a rigid body fiducial frame attached to the end-effector. However, this technique has a common problem that robot moves the fiducial frame out of the field of view, causing the image slice to become incomplete for registration. Figure 3? Illustrates an image of Masamun’s device with an incomplete number of fiducial rods. Traditional methods cannot handle this problem without acquiring extra images, which costs time. (numerical algorithm for spatial ...).

The algorithm has following advantageous:

It is resistant to noise in the input data. One of the input to the registration algorithm I the two-dimensional coordinates of the fiducials marks in the cross-sectional image. This information is produced by an image processing program at I not the subject of this investigation. It is conceivable that the locations of fiducial marks are not exactly identified in the medical images, due to suboptimal imaging and image processing techniques. These errors are considered to be “noise” in the input to the registration algorithm.

The algorithms must also run reasonably fast, in order to be useful in intraoperative applications. Computation time longer than one second? Would be prohibitive.

**Mathematical background and notation**

1. **Methodology**

The implemented algorithm for automated alignment detection of the Z-frame comprises several stages for the successful accomplishment of the task. The first step is the pre-processing of the images for reducing the noise and enhancing the features of interest for the subsequent segmentation algorithm. This stage is developed in section 3.1, where 4? different procedures are compared. The next section discusses localization and registration techniques. All the considered procedures described in this chapter are implemented, and their results are shown and investigated in chapter 4.

* 1. **Image Pre-Processing and Segmentation**

After a single 2D image of the fiducial frame has been obtained, each individual ellipse of the frame should be segmented on the MR image. Therefore, to distinguish the fiducial frame from other anatomical structures, the following filtering steps are applied to the image.

* + 1. **Automatic detection of Z-frame using Faster R-CNN**

- CNN (Convolutional Neural Network) is needed to determine the marker alignment in different positions and! from different MRI-systems with different user interfaces and different imaging properties  -> non MRI-system-specific design

**3.1.1 Denoising MRI Images Using Gaussian Smoothing Filter**

The Gaussian smoothing filter is a type of image-smoothing filter that is commonly used to reduce noise in the image. The Gaussian smoothing filter uses a Gaussian function for calculating a transformation to apply to the image. The equation of a two dimensional Gaussian function is the product of two one dimensional Gaussian functions as:

G(x,y)= e-

Where x is the distance from the origin in the horizontal axis, y is the distance from the origin in the vertical axis, and σ is the standard deviation. The Gaussian filter works by using the 2D distribution as a point-spread function. This is achieved by convolving the 2D Gaussian distribution function with the image.

**3.1.2 Nonlinear Anisotropic Filtering**

The main drawback of the linear filtering is that the details in the original image will be destroyed during reducing the noise.

Anisotropic diffusion filtering proposed by Perona and Malic [reference] is a technique which reduces the image noise but preserves details and even enhances edges.

Perona and Malik [reference] proposed a technique, called anisotropic diffusion, which reduces the image noise but preserves or even enhances the feature in the image (e.g. edges, lines) which are of high interest in image processing tasks. The suggested filter can be expressed as a diffusion process which gives preference to intra-region instead of inter-region smoothing. The novelty is that the diffusive procedure is controlled by a variable diffusion coefficient, which limits the smoothing in areas of interest (edges, boundaries). The general mathematical formulation of the mentioned technique is given in eq. number, where c(x,y,t) is the diffusion coefficient, I(x,y,t) is the image intensity and div and are the divergence and the gradient operators. The spatial coordinates of the image are represented by x and y (in the 2D case), and t corresponds to the time parameter, which in discrete implementation is the iteration number.

the main difficulty is to choose the proper diffusion coefficient. It is defined as a positive monotonically decreasing function of the image gradient which, ideally, has to be 0 at edges and 1 when the filter is located at the interior of a region. Practically, c(x,y,t) has to encourage the forward diffusion inside smooth regions ( small variations like noise and useless texture have to be removed), and backward diffusion at high gradient locations (preserving and even sharpening the boundaries and the features of interest). Perona and Mike [source] proposed two mathematical functions for the diffusion coefficient, where the first one (eq. num) advantages the high contrast edges rather than the low contrast ones, and the second one (eq.4.7) favours the wide areas instead of narrow ones.

In eq. 4.6 and 4.7 k is called the conductance parameter and has to be chosen accordingly so the anisotropic diffusion process can distinguish between an edge an intensity value corrupted by noise. Usually it is selected empirically, or, when it is the case, it is defined using a noise estimator.

The numerical scheme which implements the eq.4.4 defines the intensity change at location (x,y) after one iteration as a sum of contributions of the neighboring pixels weighted by the corresponding directed flow components (defined in eq. 4.8), as shown in eq. 4.9.

**Equations come here**

It has to be mentioned that in eq. 4.8 and 4.9 dx and dy represent the pixel spacing in the intensity image accounting for the anisotropy of the procedure. This suggests that, at a certain location, closer pixels contribute more than ones located at a higher distance. Also, the aforementioned numerical scheme refers to a 4-pixels connectivity. For a better isotropy, it can be easily extended to 8-pixel connectivity, by adding the contribution of the diagonal neighboring pixels (placed at a distance ) or even to 26-pixel connectivity in the case of 3D image datasets. In eq. 4.9 the integration constant dt is introduced. For numerical stability reasons it has to be chosen with respect to a stability criterion. It depends on the number of neighboring pixels/voxels and a full list of integration constants, considering the connectivity structure, is provided in [24]

**3.1.3 Image Contrast Enhancement**

To adjust the intensity value in the image, the *imadjust* function in MATLAB is implemented, while the range of the input values and the output values are specified in two vectors that pass to the *imadjust* as arguments. The first vector specifies the low- and high-intensity values that must be mapped and the second vector specifies the scale over which the values of the first vector should be mapped to.

**3.1.4 Image Binarization**

In this step, the filtered image is binarized with a threshold, and the cylindrical markers are extracted from the background. So any pixel for which is bigger than a threshold labeled 1 and corresponds to the fiducial maker; otherwise, the pixel corresponds to the background and labeled 0.

**3.1.5 Binary Image Mask**

At this stage, each segment which is detected as a fiducial marker, is examined based on its volume and dimensions. If the volume in a given segment is lower than a pre-defined range, it does not belong to the Z-frame structures, and should be removed from the image. Therefore, function *bwareaopen* in MATLAB is used to create a binary mask and remove small segments from the image.

* 1. **Localization and Registration**

This section is dedicated to the localization and registration of the Z-frame marker in the MR image space.

* + 1. **Center Detection of the Fiducial Ellipses**

Once all 7 ellipses with the size close to the physical size of the marker were detected, the function *regionprops* in MATLAB is used to calculate the center of each fiducial marker. The result is a 7×2 matrix where the first column represents x coordinate and the second column represents y coordinate of the center of the fiducial mass region. The resulting 7 centroids of the ellipses were ordered as illustrated in Fig.2

* + 1. **Rotation Angle Calculation of the Oriented Z-frame Marker**

Depending on the attachment of the Z-frame on the robotic device, the frame has different position, scale, and orientation in the MR images.

The ordered set of fiducial point coordinates *P’* are then applied to compute the 6-DOF pose of the Z-frame with respect to the image plane. Finally, the computed frame position and orientation is used to compute the required motion of the robotic device to reach the target pose.

* + 1. **Marker Alignment Calculations**
    2. **Angle Calculation of the MR Image Plane and Z-frame Marker**

1. **Results**
   1. **Implementation**

The automatic alignment detection algorithm which integrates the methods describes in Section 4 has been implemented using a Graphical User Interface in MATLAB R2017a. The use of GUI makes the interaction of the user with the processing steps much easier. A brief description is given here for each of the buttons and panels, for a better understanding of the offered capabilities.

* *Import image data* panel. It includes a popupmenu button for setting the path of the image volume to be analyzed. The user has three options: importing the image from a folder containing the images of the scan, or triggering a webcam or a capturing device.
* *Input parameters* panel:
* *Z-frame segmentation* panel:
* *Image pre-processing* panel: it implements the noise removal, contrast enhancement, binarization, and small object removal.
* *Center detection* panel: Using the binarized image produced in the previous step, an algorithm for computing
* *Calibration* panel: it implements the rotation of the 7 fiducial points detected in the last step and transforming the points to mm by using pixel spacing parameter. In addition, in this step the coordinate system of the points is transformed to the center point of the Z- marker.
* *Alignment detection* panel: in this step, the angles of the Z-marker and image plane is calculated as explained in section?? .
  1. **Test**

In this section, we have investigated the accuracy of the software algorithm for detection markers in data sets used for image-guided surgery.

For this part look at the article “automated fiducial marker detection for patient registration …”.

**Placement of Fiducial markers**

**Data acquisition**

**Marker detection**

**Influence of marker size**

**Influence of Threshold**