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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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**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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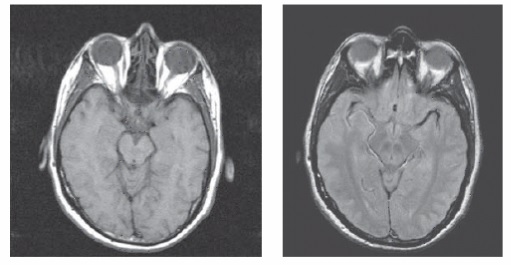
MRI is a non-invasive imaging technology generating three dimensional anatomical and functional images of the body without the use of ionizing radiation. It is particularly useful for neurological, oncological, cardiovascular, muscular and skeletal imaging. MRI employs a powerful magnetic field that 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.

**2.1 Pulse Sequences**

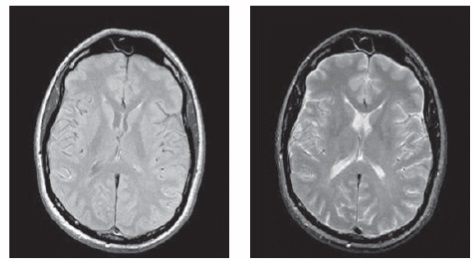
A pulse sequence is the measurement technique which contains multiple parameters such as RF pulses, gradient pulses, and timing. Depending on the anatomical region under observation, the optimal pulse sequence has to be chosen in order to acquire the data in the desired manner. There are three general 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.

**Intrinsic Parameters**

Repetition time, TR, determines the amount of T1 weighting contributing to the image contrast. A longer TR produces images with less T1 weighting, and therefore less contrast.



Echo time, TE, determines the amount of T2 weighting for spin echo images and T2\* weighting for gradient echo images.



**2.1 Measurement parameters and image contrast**

**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.

* + 1. **Artefacts in MR Imaging**

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.

**3.1.0 Automatic detection of Z-frame using Faster CNN**

**3.1.0 Cropping the Z-frame by Using Graphical User Interface**

The use of GUI makes the interaction of the user with the segmentation process much easier. A brief description is given here for each of the buttons and panels, for a better understanding of the offered capabilities.

* *Load dataset* panel. It includes a browsing button for setting the path of the image volume to be analyzed. The user can import the dataset from a folder containing DICOM files of the scan, by loading and stacking the slice scans into a 3D array. IMAGE
* *Cropping image.*

**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, 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**