A FAST 3-D MEDICAL IMAGE REGISTRATION ALGORITHM BASED ON EQUIVALENT MERIDIAN PLANE

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ABSTRACT

For the rigid registration of multi-modality medical images, mutual information (MI) technique is unsuitable to clinical diagnose because of high computational cost and low robustness. In this paper, a new concept of equivalent meridian plane (EMP) is proposed, and the EMP and other two normal feature planes are determined using principal component analysis (PCA); the rough registrations of those 2D planes are to be realized at six freedom degree; finally, the refine registrations can be completed using MI in a small neighboring region. This method is called as EMP based MI registration technique. The accuracy and robustness of EMP-MI approach can be verified by applying it to the simulated and real brain image data (CT, MR, PET, and SPECT). The experimental results indicate that the proposed algorithm reduces computational time distinctly and is a global optimal strategy.

Index Terms—equivalent meridian plane, principal component analysis, mutual information

1. INTRODUCTION

The geometric registration of multimodality images is an essential and fundamental task to clinical diagnosis. The reason for this is clear: there are numerous applications to diagnostic as well as treatment settings that benefit from integrating the complementary characters of multimodal images. Because of the challenges it pose, biomedical registration remains an active research endeavor.

Various registration methods were presented [1], and can be basically classified into two categories: feature-based and intensity-based. A feature-based [2]-[4] method requires the extraction of common features in both images including points, edges, shapes and surfaces to use them to estimate the transformation. Feature-based approaches have the advantage of greatly reducing computational complexity. However, these approaches depend on the feature extraction so as to be highly sensitive to the accuracy of the extracted features. In contrast, intensity-based registration techniques optimize an objective function to measure the similarity of all geometrically corresponding voxel pairs, and then obtain the transformation between the entire intensity images. The key advantage of intensity-based [5]-[7] methods is to avoid the difficulty of the feature extraction stage. But these approaches need expensive computational cost and are unsuitable to clinical diagnose.

We extend the meridian plane (MP) to 3D rigid medical image registration. A new concept of equivalent meridian plane is proposed, and the EMP is determined using PCA; the rough registrations of the 2D plane is to be realized at six freedom degrees; finally, the refine registrations can be completed using MI in a small neighboring region. This method is called as EMP based MI registration technique, which combines the feature and intensity information.

This paper is organized as follows. In section II, we originally propose the EMP concept. In section III, the implementation of the EMP based MI registration algorithm is described in detail. The results in both simulated and clinical medical images are illustrated in section IV. In section V, we present our conclusions and give some direction for further work.

2. EQUIVALENT MERIDIAN PLANE

As is well known, a meridian plane is arbitrary plane perpendicular to the celestial equator, which passes through the earth's axis of rotation. For three-dimensional medical image, it is necessary to propose a new EMP concept since estimating the meridian plane is not always feasible in practice.

Definition:

For a three-dimensional irregular volume, a set of orthogonal principal axes can be always found, by which a family of orthogonal planes can be determined. One of these planes, containing the first and the second principal axis, is just EMP.

The next problem is how to find EMP. We suggest using classical PCA method [8] for determining EMP. PCA produces a single best line that satisfies the following condition: the sum of the squares of the perpendicular distances from the sample points to the line is a minimum. The variable defined by the line of best fit is the first principal component which indicates the greatest amount of variation. The second principal component is the variable defined by the line that is orthogonal with the first. The center of the data set is the intersection of the two axes. The vector orthogonal to the line of best fit, together with the line of best fit, defines a plane of best fit, namely, the EMP. The sum of squares of perpendicular distances of points from the plane is a minimum. The uniquely defined plane is an optimal representation of the mass distribution.

PCA approach describes object by forming vectors from the coordinates of the object. Each pixel in the object is treated as a 3-D vector $X = \{(x_i, y_i, z_i)^T \mid i = 1, \cdots, n\}$, where T indicates transpose, n is the total number of points and xi ,y i , z i are the coordinate values of that pixel with respect to the x-, y- and z-axes. The mean vector of the population and covariance matrix can be estimated by $u = \frac{1}{n}\sum_{i=1}^{n} X_i$ and $C = \frac{1}{n}\sum_{i=1}^{n} X_i X_i^T - uu^T$. If E is a matrix whose rows are the eigenvectors of C and E is arranged in the order so that the first row of E is the eigenvector that corresponds to the largest eigenvalue of C and the last row corresponds its smallest eigenvalue, then $Y = E(X - u)$ is the PCA transform. The effects of the PCA transform on the set of points of a given image are both a translation and a rotation. The centroid of the volume is translated to the origin of the global coordinate system after translation. The principal axes of the volume will be coincident with the x-, y- and z- coordinate axes after rotation by the eigenvector matrix. By equating the eigenvector matrix of *C*

$$E = \begin{pmatrix} e_{11} & e_{12} & e_{13} \\ e_{21} & e_{22} & e_{23} \\ e_{31} & e_{32} & e_{33} \end{pmatrix}$$

to the rotation matrix:

$$R = R_x(\theta_x) \cdot R_y(\theta_y) \cdot R_z(\theta_z)$$

where $\theta_x, \theta_y, \theta_z$ are rotation angles with respect to the x-, y- and z - axes, respectively. We find that

$$\begin{aligned} \theta_y &= \arcsin(e_{31}) \\ \theta_x &= \arcsin(-e_{32} / \cos\theta_y) \qquad (1) \\ \theta_z &= \arcsin(-e_{21} / \cos\theta_y) \end{aligned}$$

3. EMP BASED MI IMAGE REGISTRATION

3.1. Mutual Information

Mutual information is defined as:

$$MI(A,B) = H(A) + H(B) - H(A,B) \qquad (2)$$

where H(A), H(B) and H(A,B)are individual entropies and joint entropy respectively. As a similarity measure, MI has enjoyed a great deal of success, particularly in the medical imaging domain [9]. It is robust to outliers and efficient to calculate. MI generally provides smooth cost function which is used for optimization. Many studies [10], [11] have compared various measures of voxel similarity and concluded that MI is the most accurate and robust measure for 3-D image registration.

3.2. Registration based on EMP

The primary drawback of the optimization-based approach is that it may fail unless the two volumes are misaligned by a moderate difference in rotation and translation. In order to address this problem, we bring the volumes into approximate alignment by utilizing PCA transform. The geometric effects of the transformation on the set of points of a given volume are both, a translation and a rotation, so that the centroid of the volume is translated to the origin of the global coordinate system and its equivalent meridian plane is positioned on the *XY* coordinate plane. These effects eliminate the problems that arise as a consequence of translations and rotations. A rough estimate of registration is realized in relatively little time using PCA transform, which is subsequently refined using the equivalent meridian plane. The EMP is coincident with *XY* coordinate plane; therefore we can fix one volume as target the other as source and make small correction of the source volume then compute the mutual information of the *XY* coordinate plane. When the volumes are correctly registered, the MI will reach maximum. The optimization is performed using Powell's method [9] which is extensively used in medical image registration. Powell's method exhibits fast convergence and is highly accurate if the initial point is close to the optimum.

The new registration technique consists of main steps follows as:

Step1. Performing binaryzation for both volumes and forming three-dimensional vectors form the coordinates of the object;

Step2. Getting centroid and covariance matrix;

Step3. Computing the PCA transformation;

Step4. Setting original volumes into a canonical coordinate frame utilizing PCA transform;

Step5. Refining registration by maximizing MI between EMP in the target and the coordinate plane (*XY*) in the source.

4. EXPERIMENTS

To test the performance of our proposed EMP-MI, it is necessary to compare the computation cost and accuracy of EMP-MI with ones of MI-based method. The platform was Matlab 7.0, Windows XP SP2 with 3.0 GHz Intel Pentium 4 CPU and 512M RAM.

4.1. Data Set Description

In experiment 1, the images were obtained from the BrainWeb Simulated Brain Database. The clinical images used in experiment 2-3 were obtained from the Biomedical Imaging Resource. The image pairs were aligned by clinical experts using interactive registration method. In experiment 4-6, the image files were provided by General Hospital, Tianjin Medical University, China. Imaging and data acquisition was performed on a novel combined PET–CT in-line system (GE Medical Systems), combining the ability to acquire CT images and PET data from the same patient in one session. The CT and PET images were acquired simultaneously. Thus, the CT and PET images are registered intrinsically. This intrinsic registration gives us an ideal method to evaluate the registration accuracy.

4.1. Experiment Setup

To generate randomly misregistered pairs, one volume was rotated sequentially along the *x*-, *y*- and *z*- axis in three different angles. The rotated image was then translated to a new position. These translations and angles had a uniform distribution over a specified range. In every experiment, a set of 100 random misregistration volume pairs were registered. An experiment is considered successful if the translation error is less than one pixel, and the rotation error is less than one degree.

4.2. Registration Results and Discussions

1) Registration of Simulated MRI Brain Volumes.
The translations and angles were uniformly distributed over [-10, 10] pixel or degree. Tables I summarized the registration results for MI and EMP-MI algorithm, including mean error of the rotation and translation, compute time and success rate. The average computational speed of EMP-MI approach is higher than one of MI (i.e., 3.6935 minutes versus 10.1407 minutes).
2) Registration of T1 and T2 MRI Brain Volumes
The translations and angles are uniformly distributed over [-20, 20] pixel or degree. The results are summarized in Table II. The success rate of the EMP-MI approach is slightly better than the MI approach. Most of the failed registrations by EMP-MI approach are due to the noise in the T2 volume.
3) Registration of MR-SPECT Brain Volumes
The rotation angles were uniformly distributed over [-10, 10] degrees and the translations were uniformly distributed over [-20, 20] pixel. The statistics of the registration are shown in Table III. It is clear that the advantages of our algorithm over MI-based method are execution time and accuracy. Our method is much faster than the MI-based algorithm, by a factor of 4 or more.
4) Registration of CT-PET Cardiac Volumes
The rotation angles were uniformly distributed over [-20, 20] degrees and the translations were uniformly distributed over [-10, 10] pixel. The difference of the resulting registration parameters and the true ones were then statistically analyzed in Table IV. Cardiac image registration is a more complex problem than brain image registration [30], particularly because of the nonrigid and mixed motions of the heart and the thorax structures. The success ratio of MI-based method is only 63%, and one of EMP-MI is 81%, which is lower than the previous four experiments (>90%).
5) Registration of CT-PET Brain Volumes
The rotation angles were uniformly distributed over [-10, 10] degrees and the translational offsets were uniformly distributed over [-20, 20] pixel. The difference of the resulting registration parameters and the true ones were then statistically analyzed in Table V.
6) Registration of CT-PET Brain Volumes
The rotation angles were uniformly distributed over [-20, 20] degrees and the translational offsets were uniformly distributed over [-10, 10] pixel. The difference of the resulting registration parameters and the true ones were then statistically analyzed in Table VI.

5. CONCLUSIONS AND FUTURE WORK

In this paper, we defined a new concept of EMP and presented a two-stage global optimization registration for medical images. All of the experiments, including simulated data and clinical data, show that EMP-MI is not only more robust and faster, but also quite automatic. In practice, a lot of clinical experts highly appreciate and use it. For some special cases, we may have a sort of incomplete data (such as, the structures in one modality image absent in another modality one). We are developing a new method based on EMP-MI to solve this problem.

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6. REFERENCES

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TABLE I
REGISTRATION RESULTS FOR SIMULATED MR-T1 AND MR-PD VOLUME PAIRS
(MR-T1 OR MR-PD VOLUME SIZE: 181×217×181)

Algorithm	t_x	t_y	θ_z	θ_x	θ_y	t_z	*Time*	*Success*
MI	*0.062*	*0.079*	*0.081*	*0.108*	*0.095*	*0.070*	*10.14*	*100%*
EMP-MI	*0.100*	*0.112*	*0.117*	*0.102*	*0.007*	*0.003*	*3.69*	*100%*

TABLE II
REGISTRATION RESULTS FOR MR-T1 AND MR-T2 VOLUME PAIRS
(MR-T1VOLUME SIZE: 256^2×62 AND MR-T2VOLUME SIZE: 256^2×124)

Algorithm	t_x	t_y	θ_z	θ_x	θ_y	t_z	*Time*	*Success*
MI	*0.210*	*0.175*	*0.102*	*0.122*	*0.130*	*0.460*	*25.16*	*94%*
EMP-MI	*0.192*	*0.141*	*0.028*	*0.107*	*0.070*	*0.386*	*7.04*	*96%*

TABLE III
REGISTRATION RESULTS FOR MR AND SPECT IMAGE PAIRS
(MR VOLUME SIZE: 256^2×124 AND PET VOLUME SIZE: 256^2×124)

Algorithm	t_x	t_y	θ_z	θ_x	θ_y	t_z	*Time*	*Success*
MI	*0.152*	*0.868*	*0.416*	*0.203*	*0.570*	*0.113*	*36.60*	*75%*
EMP-MI	*0.107*	*0.854*	*0.412*	*0.133*	*0.552*	*0.106*	*8.43*	*92%*

TABLE IV
REGISTRATION RESULTS FOR CARDIAC CT - PET IMAGE PAIRS
(CT VOLUME SIZE: 512^2×35AND PET VOLUME SIZE: 128^2×35)

Algorithm	t_x	t_y	θ_z	θ_x	θ_y	t_z	*Time*	*Success*
MI	*0.247*	*0.126*	*0.072*	*0.127*	*0.224*	*1.062*	*5.06*	*63%*
EMP-MI	*0.208*	*0.121*	*0.062*	*0.053*	*0.141*	*0.483*	*2.01*	*81%*

TABLE V
REGISTRATION RESULTS FOR BRAIN CT - PET IMAGE PAIRS
(CT VOLUME SIZE: 512^2×35 AND PET VOLUME SIZE: 128^2×35)

Algorithm	t_x	t_y	θ_z	θ_x	θ_y	t_z	*Time*	*Success*
MI	*0.475*	*0.422*	*0.430*	*0.642*	*0.343*	*0.363*	*4.84*	*70%*
EMP-MI	*0.119*	*0.094*	*0.094*	*0.340*	*0.283*	*0.311*	*2.10*	*89%*

TABLE VI
REGISTRATION RESULTS FOR BRAIN CT - PET IMAGE PAIRS
(CT VOLUME SIZE: 512^2×35AND PET VOLUME SIZE: 128^2×35)

Algorithm	t_x	t_y	θ_z	θ_x	θ_y	t_z	*Time*	*Success*
MI	*0.498*	*0.123*	*0.308*	*0.230*	*0.211*	*0.426*	*4.98*	*65%*
EMP-MI	*0.365*	*0.117*	*0.290*	*0.206*	*0.202*	*0.409*	*1.74*	*88%*