A Method for Correcting Inter-Series motion in Brain MRI for Auto Scan Plane Planning

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Introduction: Patient motion during Magnetic Resonance Imaging (MRI) is a major degradation factor of MR image quality. In many cases, patient motion moves regions of interest prescribed by the operator out of the field of view and creates sub-optimal views for diagnosis. In this abstract, we propose an algorithm that relies on the known position and orientation of the patient anatomy at the beginning of a scanning session, and uses fast three-plane (3P) localizers to update patient position/orientation just before an image is acquired. The algorithm finds a rigid transform that best aligns the 3P localizers to the full volumetric localizer with known anatomy, which is acquired at the beginning of the scanning session. This rigid transform is then used to compensate for the patient motion and produce patient-centric scan plane prescription. This algorithm is shown to be effective in correcting for small motion, which is often observed during brain MR scans. We have incorporated this approach in a clinical MR system and demonstrated its usefulness in automatically obtaining consistent imaging planes in brain exams, even with the presence of patient motion.

Methods: In our previous work [1], we have developed an algorithm for automatically determining the brain anatomy orientation from a volumetric MR image. The algorithm is used in a system for automatic scan plane planning for repeatable data acquisition, and has been shown to be effective in prescribing anatomically relevant, rather than scanner relevant, scan planes. We refer to the rigid transform relating the scanner coordinate system to the desired scan plane in the initial patient position as $\mathbf{T}^*$, and the initial volumetric localizer as $I_O(x,y,z)$.

To address the patient motion that occurs after the initial registration step, we obtain a fast sparse set of localizer images in three orthogonal planes, $I_A(x,y,z)$, $I_S(x,y,z)$, and $I_C(x,y,z)$. The axial, sagittal, and coronal orientations of these images are defined in the scanner coordinate frame. A registration step is used to find a rigid transform **R** that aligns $I_O(x,y,z)$ to $I_A(x,y,z)$, $I_S(x,y,z)$, and $I_C(x,y,z)$ simultaneously. The cost function for a given **R** is defined as:

$$\varepsilon(\mathbf{R}) = d\big(\mathbf{R}\circ I_O(x,y,z), I_A(x,y,z)\big) + d\big(\mathbf{R}\circ I_O(x,y,z), I_S(x,y,z)\big) + d\big(\mathbf{R}\circ I_O(x,y,z), I_C(x,y,z)\big), \quad (1)$$

where $d(\bullet,\bullet)$ is the disparity measure between two images, and $\mathbf{R}\circ I_O$ is image I_O under transform $\mathbf{R}$ evaluated in appropriate spatial locations. In this work, we use the negative of mutual information [2] as the image disparity measure. The solution $\mathbf{R}^*$ that minimizes Eq. (1) relates the current patient position to the initial patient position. The composition of $\mathbf{R}^*$ and $\mathbf{T}^*$ is the transformation between the scanner coordinate frame and the patient anatomy and is used for scan plane prescription. This algorithm is implemented using the Insight Toolkit [3]. It is integrated on the scanner with a simple user interface, where the operator can initiate the update step whenever a patient motion is suspected. In our current implementation, it takes 5 seconds to acquire the localizer images and about 7 seconds for the algorithm to compute the new transformation. This is significantly faster than a 30 second whole head localizer volume acquisition and 25-30 seconds full registration as presented in [1].

Results: We tested the algorithm on two volunteers for brain MR scans under an IRB approved study. For each volunteer, the following experiment was conducted: a whole-head 3D axial localizer image was acquired at the beginning of the scan session using a fast spoiled gradient echo sequence (TR/TE 5.7/1.2 ms, Flip 11˚, Matrix 112x112, FOV 28cm, slice thickness 3mm, 64 slices) (Fig. 1a). This whole head localizer image was used to determine the initial position and orientation of the patient, $\mathbf{T}^*$ using the method in [1]. A set of “standard” axial images (defined in the patient anatomical space) was acquired using this transform (Fig. 1b). Then, the volunteer was instructed to move his head. A set of three-plane localizers [FGRE sequence with 5 slices in each of Sag, Cor, Ax orientations, with FOV 30 cm, slice thickness 5 mm, and slice spacing 5 mm, matrix 256x128] was acquired. A new set of whole head axial localizer was also acquired in this new head position (Fig. 1c). Fast registration was performed as described above between the 3P localizer after head motion and the original whole head volume localizer to estimate the patient motion $\mathbf{R}^*$. The composition of $\mathbf{R}^*$ and $\mathbf{T}^*$ was used to acquire a “standard” axial scan using the fast method as described above (Fig. 1d). For comparison purposes, a full registration was also performed using the whole head volume localizer data of 1c giving images in Fig.1f. Excellent match can be seen in Fig 1b,d,f showing that consistent scan planes can be obtained by using this 3P localizer based fast registration.

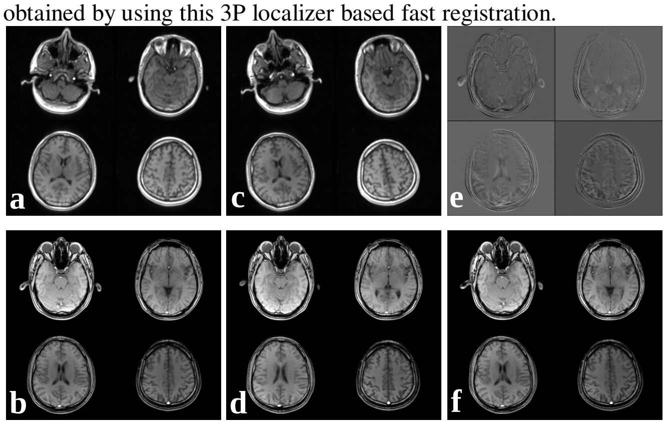

Fig. 1: Example results from one subject: (a) Representative axial images of original whole head localizer; (b) “Std Axial” images in original head position; (c) Axial scout images in new head position; (d) “Std axial” images using fast registration from 3P localizer data; (e) difference images between (d) and (b); (f) “Std axial” images using full localizer data in (c).

Conclusion and Discussion: We have presented an algorithm for compensating motion during a brain MR imaging session. The algorithm relies on the known position and orientation of the patient anatomy at the beginning of a scanning session, and uses fast three-plane (3P) localizers to update patient position/orientation just before an image is acquired. Initial experiments demonstrated that the algorithm is effective and accurate. Future work includes more clinical study and quantification of the accuracy and the limits of the method.

References: [1] ISMRM submission# 808, 2009. [2] *IEEETrans Med Imag.* 22(1), 120-128; Jan 2003. [3] http://www.itk.org.