A FAST IMAGE REGISTRATION TECHNIQUE FOR MOTION ARTIFACT REDUCTION IN DSA

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ABSTRACT

In digital subtraction angiography (DSA), patient motion is the primary cause of image quality degradation. The motion correction algorithms developed so far were not sufficiently fast so as to be suitable for integration in a clinical setting. In this paper we describe a new image registration technique for motion artifact reduction in DSA which is fully automatic, effective, and computationally very efficient. Using an image content driven control point selection mechanism and modern graphics hardware for image warping, the algorithm requires less than one second per DSA image (on average). Preliminary experiments on cerebral DSA images illustrate the applicability of the technique.

1. INTRODUCTION

Digital subtraction angiography (DSA) is a well-established technique for visualization of blood vessels in the human body. With this technique, a sequence of X-ray projection images is acquired during the passage of a bolus of injected contrast material through the vessels of interest. By subtracting an image acquired prior to arrival of the contrast medium (the mask image), background structures in the contrast images are largely removed. However, due to patient motion, DSA images often show misregistration artifacts that may hamper proper diagnosis. An example of this type of artifacts in a cerebral DSA image is provided in Fig. 1.

Currently, the only available tool for motion correction on commercial DSA devices is manual pixel shifting, which only allows for correction of global translational motion. Moreover, achieving a correction manually may be quite cumbersome. Therefore, many semi- or even full automatic correction techniques have been developed over the past two decades [1]. However, these were not sufficiently fast so as to be suitable for integration in a clinical setting.

In this paper we describe a new image registration technique for the reduction of motion artifacts in DSA, which is fully automatic, effective, and computationally very efficient. The applicability of the technique is illustrated by preliminary experiments with cerebral DSA images.

2. PRELIMINARIES

It is important to note that the individual images of angiographic X-ray image sequences are in fact 2-D projections of 3-D anatomical structures. In the attempt to correct for motion artifacts in the subtraction images, the use of a 2-D image registration technique is justified only when it can be shown, at least theoretically, that it is possible to construct a 2-D geometrical transformation that completely accounts for the projective effects of 3-D patient motion.

The existence of 2-D geometrical transformations between X-ray projection images was studied by Fitzpatrick [2]. He showed that (subject to a few conditions, which are easily met in DSA imaging), given two X-ray images, I_0 and I, taken at times t_0 and t respectively, there always exists a one-to-one 2-D mapping $\Psi : \mathbb{R}^2 \rightarrow \mathbb{R}^2$ that transforms points $\mathbf{x}_0$ in I_0 into their corresponding points $\mathbf{x} = \Psi(\mathbf{x}_0)$ in I, and that the associated change in grey value is given by

$$I(\mathbf{x}) = J_\Psi^{-1}(\mathbf{x}_0) I_0(\mathbf{x}_0) \tag{1}$$

where J_Ψ^{-1} in the inverse of the Jacobian of the mapping Ψ. As argued by Fitzpatrick [3], the factor J_Ψ^{-1} will be finite and larger than zero for all transformations describing physical motion.

However, in most practical cases it will be impossible to retrieve a transformation that exactly satisfies Eq. (1). This is primarily due to the aperture problem, the inherent dissimilarities as caused by the contrast material, and the inevitable use of neighborhood operations for the computation of local displacements [1,4].

In *IEEE International Conference on Image Processing – ICIP'99*,
IEEE Computer Society Press, Los Alamitos, CA, vol. III, 1999, pp. 435–439.

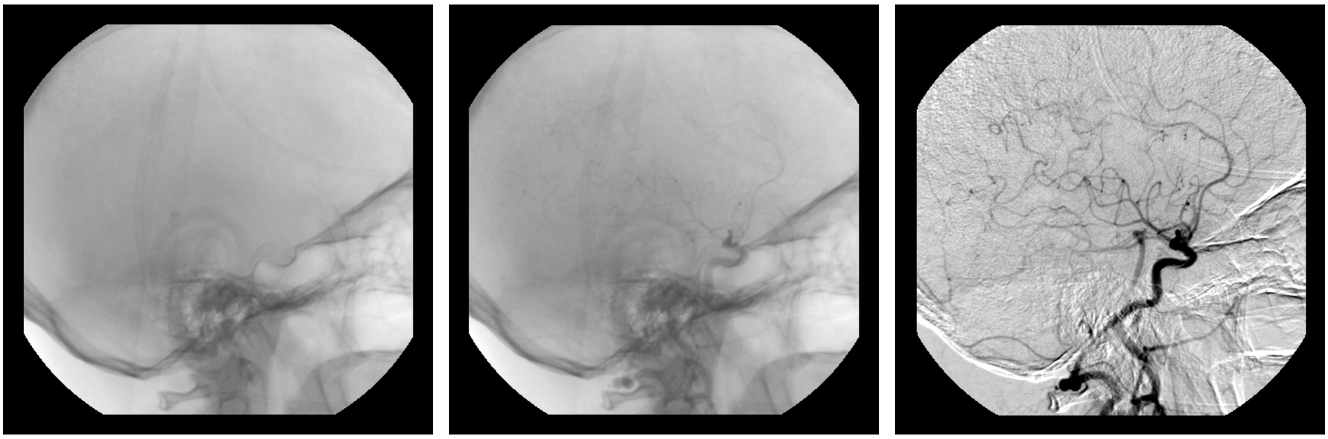

Figure 1: An example of motion artifacts in a cerebral DSA image (right), obtained by subtraction of the mask image (left) from the contrast image (middle), and subsequent image-contrast enhancement. The artifacts appear as black and white structures.

These factors will limit the accuracy of any registration technique for this type of images.

3. REGISTRATION TECHNIQUE

In this section we briefly describe the individual steps in the registration approach, as summarized in Fig. 2. For an in-depth discussion of the technique we refer to a more elaborate paper [4].

3.1. Control Points Selection

Even with today's modern workstations, it is computationally too expensive to compute the displacement explicitly for every individual pixel in images of 512×512 or 1024×1024 pixels (the usual dimensions of digital angiographic images). In order to reduce the required computation time to a clinically acceptable level (in the order of seconds), a frequently used approach is to consider only a limited set of so called control points. The overall displacement vector field can then be obtained by interpolation.

In many of the techniques reported in the literature, control points are chosen on a regular grid [5–9], without taking into consideration the underlying image content. However, since motion artifacts can only appear in regions that contain significant grey-level variations, and because these regions can be matched better than more homogeneous regions, the selection of control points should be image content driven.

In our approach, potential artifact regions are extracted by applying Canny's edge detection scheme [10] to the mask image, I_0. Next, control points are selected at local maxima of the gradient magnitude $||\nabla I_0||$ (provided that these maxima exceed a predefined edge-threshold), while constraining the minimum and maximum distance between these points [4].

3.2. Displacement Computation

Techniques for the computation of local displacements in images can be divided into gradient based optic flow techniques, and template matching based techniques. Pilot experiments [4] have indicated that the latter are more suitable for digital angiographic images.

A crucial aspect of any template matching algorithm is the similarity measure that is used to determine the amount of correspondence of regions in successive images. In a recent paper [1] we have summarized and discussed the measures and comparative evaluations of many authors, from which it was concluded that histogram based measures are most suitable.

In our approach, we use the so called energy measure [11–13], which is computed as

$$\mathcal{M}(\mathbf{d}) = \sum_g \mathcal{H}^2(g; \mathbf{d}) \qquad (2)$$

where $\mathcal{H}$ denotes the normalized histogram of the grey-value differences, g, of two regions to be matched, as a function of the displacement, $\mathbf{d}$. This measure is insensitive to mean grey-level offsets and local dissimilarities caused by contrasted blood vessels. In addition, it is computationally cheap and leads to very smooth match surfaces, which allows for efficient optimization by means of hill climbing. In our current implementation, displacement vectors are computed with an accuracy of 0.1 pixel.

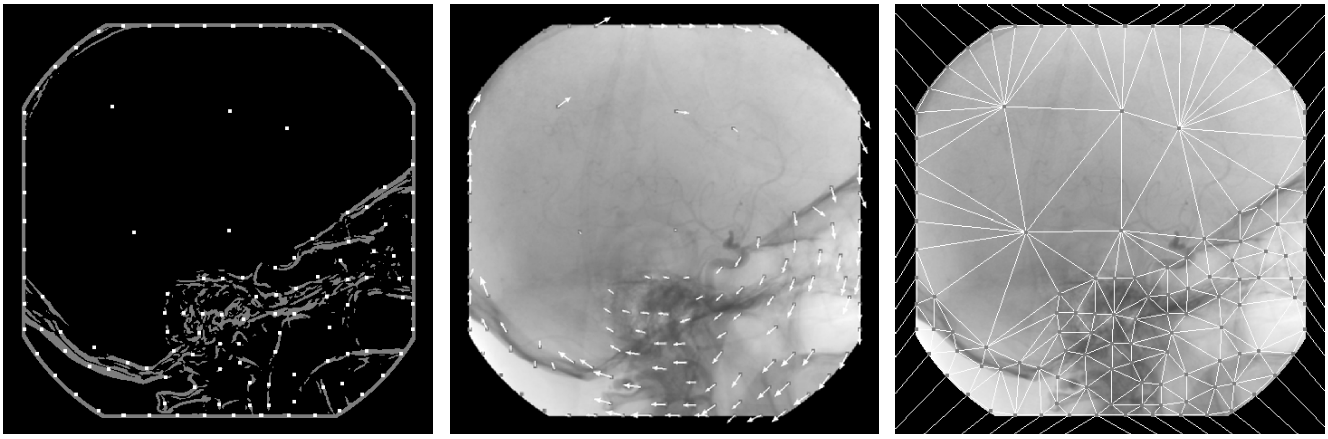

Figure 2: Overview of the successive stages and operations of the image registration technique. Starting with the mask image (left image in Fig. 1), the potential artifact regions are extracted by applying Canny's edge detection scheme. Next, control points are selected (left), followed by the computation of the corresponding displacement vectors (middle). The total displacement vector field is obtained by triangulation of the set of control points (right) and linear interpolation. Finally, the corrected DSA image is obtained by warping the mask image according to this vector field.

3.3. Mask Image Warping

The final operation in the registration approach is the warping of the mask image. Therefore, a complete description of the displacement vector field is required. As mentioned in Section 3.1, the displacement at an arbitrary point in the image can be obtained by interpolation of the computed displacement vectors of the control points. In order to minimize computation time, we use linear interpolation. To this end, the set of control points is tessellated into a Delaunay triangulation, for which we use the incremental approach proposed by Watson [14]. The warping of triangular meshes can be carried out real-time by using the graphics hardware of modern workstations.

4. RESULTS & DISCUSSION

The algorithm was implemented in the C++ programming language, utilizing the Open Graphics Library. The experiments were carried out on a relatively low cost Silicon Graphics O2 workstation (one 180 MHz R5000 processor, 64MB/512kB main/cache memory, and hardware support for OpenGL instructions).

Pilot experiments have indicated that the algorithm is most effective in cerebral and peripheral DSA images, since in general they do not contain considerable over-projections of independently moving structures [4]. In order to illustrate the performance of the algorithm for the first type of images, eight cerebral angiographic image sequences were acquired on an Integris V3000 C-arm imaging system (Philips Medical Systems, Best, the Netherlands). For selected mask-contrast pairs from these sequences (in which the blood vessels were sufficiently filled with contrast material), the results of applying the automatic image registration technique are shown in Fig. 3.

As can be appreciated from the first six examples in Fig. 3, the algorithm was capable of removing virtually all motion artifacts. Even though, in some cases, the artifacts could also be removed by manual pixel shifting, the advantage of the described registration approach is that it is fully automatic and requires less than one second per DSA image (on average). The second last example in Fig. 3 shows an extreme case, in which the motion artifacts were very serious. Although some artifacts could be removed, this example illustrates that in extreme cases, the performance of the algorithm is limited. This is due to the fact that in the case of very large displacements, the hill-climbing optimization procedure (see Section 3.2) is more likely to become trapped in local optima that do not correspond to the actual displacement. The last example in Fig. 3 was included to show that in the case of complete absence of motion artifacts, images are left intact by the algorithm.

5. CONCLUSIONS

In this paper we have described a new image registration technique for the reduction of motion artifacts in

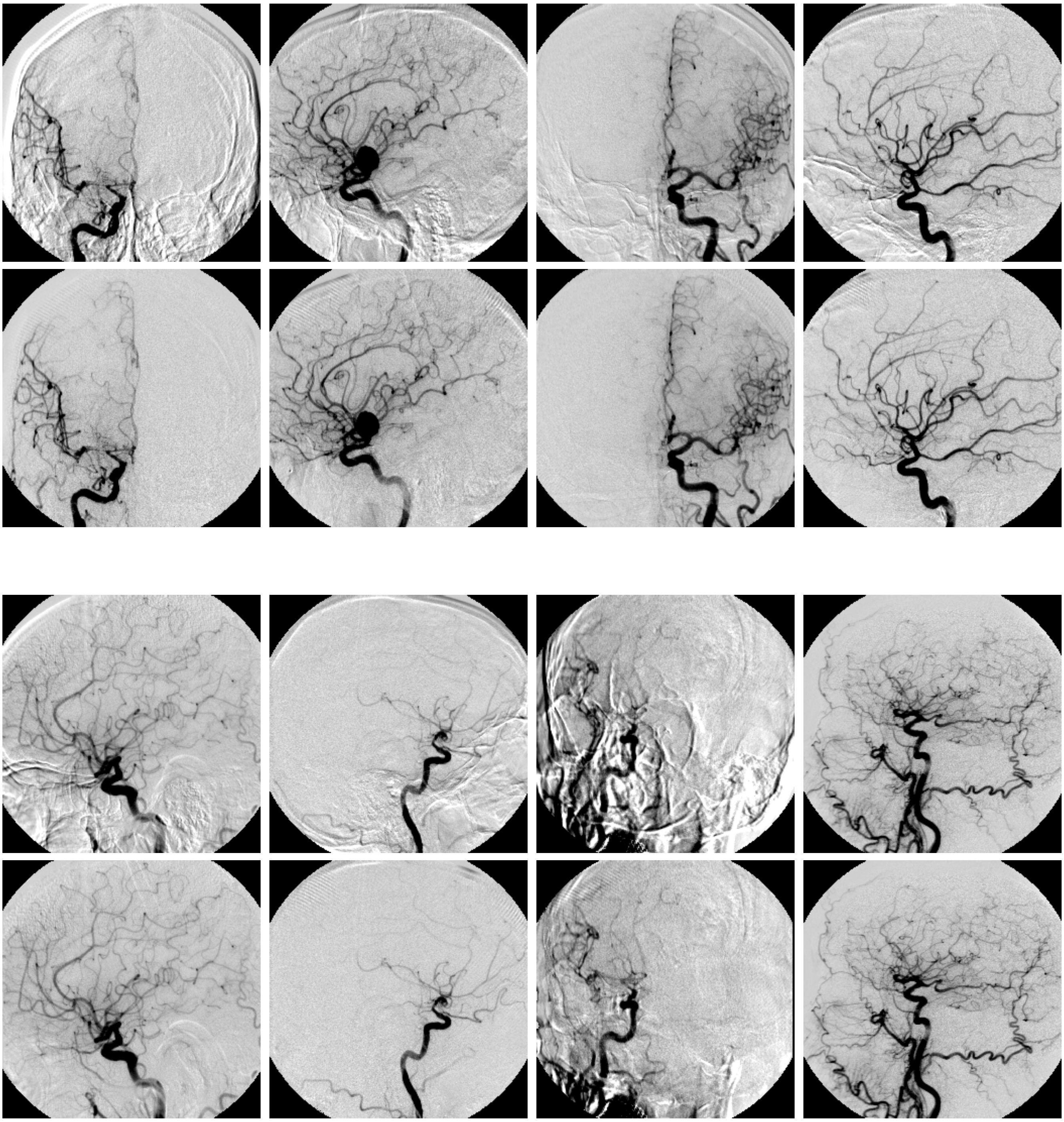

Figure 3: Results of preliminary experiments with cerebral DSA images. First and third row: the original (uncorrected) DSA images. Second and fourth row: the corresponding corrected DSA images resulting from the automatic image registration technique described in this paper. See Section 4 for a discussion.

DSA images. The algorithm is fully automatic, effective, and computationally very efficient. Thanks to an image content driven control point selection mechanism and the use of modern graphics hardware for image warping, the algorithm requires less than one second per image (on average). The results of a preliminary evaluation on cerebral DSA images have clearly illustrated the applicability of the technique.

6. REFERENCES

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