2D/3D Image Registration on the GPU

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Abstract—We present a method that performs a rigid 2D/3D image registration efficiently on the Graphical Processing Unit (GPU). As one main contribution of this paper, we propose an efficient method for generating realistic DRRs that are visually similar to x-ray images. Therefore, we model some of the electronic post-processes of current x-ray C-arm-systems. As another main contribution, the GPU is used to compute eight intensity-based similarity measures between the DRR and the x-ray image in parallel. A combination of these eight similarity measures is used as a new similarity measure for the optimization. We evaluated the performance and the precision of our 2D/3D image registration algorithm using two phantom models. Compared to a CPU + GPU algorithm, which calculates the similarity measures on the CPU, our GPU algorithm is between three and six times faster. In contrast to single similarity measures, our new similarity measure achieved precise and robust registration results for both phantom models.

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1. INTRODUCTION

As mentioned in [1], image registration is a very common problem in medical image processing, and, thus, automatic image registration is a very important component in current medical imaging systems [2]. In 2D/3D image registration, a preoperative volume is registered with an intraoperative x-ray image. Thus, the preoperatively acquired volume can be used for intraoperative therapy guidance [3]. Intensity-based methods for the 2D/3D image registration consist of two parts, namely the DRR generation and the computation of a similarity measure between the DRR and the x-ray image. We restrict ourselves to rigid 2D/3D image registration in which six parameters (three translations and three rotations) must be estimated.

Because of the increasing performance of Graphical Processing Units (GPUs) in the last fifteen years, the GPU has become an attractive target for image registration algorithms. In [4] the regularized gradient flow algorithm has been implemented on the GPU for non-rigid 2D/2D-registration. This algorithm has been extended in [5] for 3D/3D image registration, likewise by using the GPU. Several approaches were developed to perform the 2D/3D registration on the GPU. The 2D/3D-registration consists of the generation of digitally reconstructed radiographs (DRRs) and the computation of similarity measures [6]. Some of these approaches like [7] and [8] generated only the DRRs on the GPU, while the similarity measures were still computed on the CPU.

Lately, some approaches like [9] implemented both parts of the 2D/3D registration on the GPU and achieved huge performance speedups.

In this paper we present a method that performs the rigid 2D/3D image registration efficiently on the GPU. Both parts of our registration algorithm, i.e., the DRR generation and the computation of the similarity measure, are executed on the GPU. Our method for generating DRRs is based on the GPU-based volume rendering algorithm that was published in [10]. We also use the improvements for this algorithm that were mentioned in [11]. As one main contribution of this paper, we extend the algorithm in [11] so that it produces realistic DRRs that are visually similar to x-ray images. In this paper we use the term realistic or visually similar if the intensities of the DRR and the x-ray image are nearly the same. By increasing the similarity of DRRs and x-ray images, the image registration will be simplified. The similarity can be increased because we model some of electronic post-processes of current x-ray C-arm-systems in our method. After accomplishing the DRR generation, the GPU is used to compute eight intensity-based similarity measures between the DRR and the x-ray image in parallel. Instead of using only single similarity measures for the registration, we use a combination of the eight similarity measures to ensure precise and robust registration results.

In Section 2 we describe our 2D/3D image registration method that is executed on the GPU. We describe the realistic DRR generation and the parallel computation of eight similarity measures on the GPU in this section. In Section 3 we show and discuss the results of our implementation. Finally, in Section 4 we conclude and present some ideas for our future work.
2. 2D/3D IMAGE REGISTRATION ON THE GPU

Image registration is a very common problem in medical image processing. In medical settings images of the same modality or of different modalities are often needed in order to provide precise diagnoses. However, a meaningful usage of different images is only possible if the images were correctly aligned before. Therefore, an image registration algorithm is deployed.

In 2D/3D image registration, a preoperative volume (e.g., CT or MRT) is registered with an intraoperative x-ray image. In this paper we restrict ourselves to rigid image registration where the volume can only be translated and rotated according to three coordinate axes. This transformation is given by the parameter vector \( x = (t_x, t_y, t_z, r_x, r_y, r_z) \). Thereby, the parameters \( t_x, t_y, t_z \) represent the translation in mm along the \( X-, Y-, \) and \( Z- \) axis, whereas the parameters \( r_x, r_y, r_z \) belong to the Rodrigues \([12]\) vector \( r = (r_x, r_y, r_z) \).

As mentioned before, the image registration algorithm consists of two parts, i.e., the DRR generation and the computation of a similarity measure between the DRR and the x-ray image that should be aligned to the volume. After translating and rotating the volume, the DRR \( I_{\text{DRR}}(x) \) is produced from the volume according to the current transformation \( x \) of the volume. The DRR is generated by the method that is described in Subsection 2.1. Afterwards, the DRR \( I_{\text{DRR}}(x) \) is compared to the x-ray image \( I_{\text{FULL}} \) by applying a similarity measure \( S \) according to (1). The used x-ray image \( I_{\text{FULL}} \) was acquired by an x-ray C-arm-system earlier.

\[
S(I_{\text{FULL}}, I_{\text{DRR}}(x)).
\]  

By using a certain optimization technique, the similarity between the DRR and the x-ray image will be increased according to (2) until the optimal parameter vector \( x^* \) is found. The method for computing the similarity measure is described in Subsection 2.2. Our optimization technique consists of a combination of the global optimizer Adaptive Random Search \([13]\) and the local optimizer Best Neighbor \([14]\).

\[
x^* = \arg\max_x S(I_{\text{FULL}}, I_{\text{DRR}}(x)).
\]

2.1. Realistic DRR Generation

Intuitively, it is easier to align the DRR and the x-ray image, if they are visually similar. If the DRR is not similar to the x-ray image, either the DRR or the x-ray image must be adapted to the other image by executing postprocessing algorithms. Instead of costly postprocessing algorithms, we develop a method for the realistic DRR generation. In this method we model some of the electronic postprocesses that are executed in current x-ray C-arm-systems. However, we restrict ourselves to those electronic postprocessing algorithms that significantly affect the intensities and the contrast of the x-ray image. Hence, in our method the DRRs are created in a similar way as in current x-ray C-arm-systems and, thus, the DRR will be visually similar to x-ray images, without the need for postprocessing algorithms.

**Theoretical background.** The value of a pixel in an x-ray image depends on two factors, namely the intensity of the incident x-rays on the detector and the following electronic postprocesses \([6]\). In contrast to \([6]\), some of these electronic postprocesses are also integrated in our method for the realistic DRR generation.

The intensity \( J \) of the incident x-rays depends on three factors, i.e., the initial intensity \( J_0 \), the distance between the x-ray source and the detector, and the attenuation of the x-rays by the material through which the x-rays pass \([6]\). The attenuation coefficients \( \mu_i \) describe the percentage by which the x-ray is attenuated in the passed material. For instance, water has the attenuation coefficient \( \mu_{\text{Water}} = 0.25 \text{ cm}^{-1} \) \([15]\). In the computer tomography sector, Hounsfield Units (HU) \([15]\) are used in order to specify the attenuation of the passed tissue. HU values are often in the domain \([-1000; 7000]\) \([15]\). For instance, water has 0 HU. They can be computed from the attenuation coefficients \( \mu \) by

\[
\text{HU} = 1000 \frac{\mu - \mu_{\text{Water}}}{\mu_{\text{Water}}}. \quad (3)
\]

If the x-ray passes the distance \( d \) through a homogeneous object with the attenuation coefficient \( \mu \), the initial intensity \( J_0 \) is attenuated to the intensity \( J \)

\[
J = J_0 e^{-\mu d}. \quad (4)
\]

However, the human body is not a homogeneous object. It can contain several kinds of tissue with different attenuation coefficients \( \mu_i \). Additionally, each kind of tissue has a certain thickness \( d_i \). The attenuated intensity \( J \) can be computed correctly and approximately

\[
J = J_0 e^{-\sum \mu_i d_i} = J_0 e^{-\sum \mu_i d_i}. \quad (5)
\]

If the volume consists of an equidistant grid of discrete voxels and each voxel has the same size, the computation of the attenuated intensity \( J \) can be simplified to

\[
J = J_0 e^{-\sum \mu_i d_i}. \quad (6)
\]

After the x-rays reached the detector, the intensity \( J(x, y) \) can be measured at each point \((x, y)\) of the detector. Hereon, the measured intensities \( J(x, y) \) will be converted into gray level values by executing some of the electronic postprocesses of the C-arm-system. Thus, in the following we model some of these electronic postprocessing steps of current C-arm systems. Firstly, the measured intensities \( J(x, y) \) must be normalized, so that their mean value corresponds to a given desired value \( J_{\text{desired}} \). Thereto, in the inner circle or rectangle \( \Omega \) of the DRR, the average of \(|\Omega|\) gray values in \( \Omega \) is computed.

\[\text{average of } |\Omega| \text{ gray values in } \Omega \text{ is computed.}\]