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Automatic determination of the dynamic geometry of abdominal aortic aneurysm from MR with application to wall stress simulations

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Abstract. The current surgical intervention criterion for abdominal aortic aneurysm is based on the maximum transverse diameter of the aorta. Recent research advances indicate that a better rupture predictor may be derived from the wall stress, which can be computed with the finite element method. An essential prerequisite for this modelling is an accurate description of the geometry of the aneurysm. We developed an automatic method to derive the dynamic patient-specific aneurysm geometry from non-contrast enhanced MRA balanced turbo field images. The slices of our 2D-scanned volumes are registered onto 3D-scanned volumes to restore spatial coherence. The resulting images are noise-filtered and the enddiastolic volume is segmented with an active objects technique (deformable models). The resulting geometrical model is propagated to the remaining phases using the correlation between grey value profiles on the surface as an external force for the active object. From our segmentations we derived tetrahedral finite element meshes which were used as the input for finite element wall stress simulations. © 2005 CARS & Elsevier B.V. All rights reserved.

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1. Introduction

An abdominal aortic aneurysm (AAA) is a permanent dilation of the human aorta which is seriously life threatening in the event of a rupture. In current clinical practice, surgical intervention is considered once the maximum transverse diameter exceeds 55 mm. Recent advances in research on AAA intervention criteria have shown that wall stress simulations, based on the finite element method (FEM) may provide better intervention criteria for AAA [1]. In order to assess the wall stress, we are working towards patient-specific FEM models based on both computed tomography angiography (CTA) and magnetic resonance angiography (MRA) [2]. Most methods for segmentation of AAA are based on CTA images, which may provide the luminal geometry and the outer vessel wall location [1,3,4]. With MRA, we can obtain the dynamic geometry of the AAA and measure hemodynamic input parameters such as the quantitative flow, without having to submit the patient to harmful radiation. For truly patient-specific wall stress simulations for AAA, further input such as vessel wall thickness and material properties are necessary. In this work, we present a method to reconstruct the dynamic geometry of AAA from MRA, which is especially well suited for FEM mesh generation. We also present the results of wall stress simulations we performed for six individual AAA patients.

2. Material and methods

2.1. Imaging

2D balanced turbo field echo (B-TFE) images have been acquired on a Philips Gyroscan Intera 1.5T MR scanner (Rel. 10.4) for six male AAA patients who were scheduled for surgical repair (age: min. 60, max. 77, av. 69) for 25 slices and 12 cardiac phases (SENSE cardiac coil, TE/TR = 2.14/4.28 ms, flip angle 50°, FOV = 300 mm, matrix scan 224, voxel dimensions $1.2 \times 1.2 \times 6.0$ mm, slice gap 0 mm, no breath-holding, non-contrast-enhanced). 3D B-TFE images were acquired with similar parameters for 50 slices with an overlap of 3 mm.

2.2. Image registration

Between the acquisition of the 3D B-TFE and the 2D B-TFE images, the patient may have moved. Furthermore, the 2D B-TFE images are acquired slice by slice in an interleaved manner, which means that these slices may also be slightly dislocated with respect to each other. For these reasons, image registration is necessary. In the first step, we register the 2D and 3D B-TFE volumes using a rigid motion model (rotations and translations with respect to the three principal axes) using the normalised mutual information as similarity measure. We have assumed that further transformations of the individual slices of the 2D B-TFE scan are well described with an in-slice translation and a rotation around the patient's transversal axis. With this model we registered the individual 2D B-TFE slices onto the 3D B-TFE volume, again with the normalised mutual similarity measure, to further refine the registration. Reconstruction of the final registered images can thus be performed with in-slice interpolation only.

2.3. Segmentation initialisation

First, the 2D B-TFE images are locally anisotropically filtered, tangentially to edges, in order to reduce noise without affecting the edge location [5]. To delineate the AAA region of interest, the user selects a point in the AAA lumen below the renal arteries and a second

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