Estimation of distensibility of the aorta using non-rigid registration of cine MR images

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INTRODUCTION

The assessment of the elasticity of the great vessels is important to evaluate cardiovascular disease and performance since it reflects aging and atherosclerosis [1,2]. This elasticity assessment is usually accomplished using compliance or distensibility measurements, which are wall properties of great vessels. A common approach is to use measurements of area changes to calculate compliance or distensibility. Manual delineation of the great vessels is usually performed in order to compute the area, however it is time consuming and there is a significant intra- and inter-observer variability, which make the results inconsistent. We therefore propose a method to estimate the area changes of the aorta from MR images automatically.

METHOD

Image registration is a technique where the geometrical relation between two images is described by a transformation that maps points in one image to its corresponding point in another image. To model local deformations we are using the non-rigid transformation algorithm proposed by Rueckert *et al.* [3] that is based on free-form deformations and B-splines. The algorithm manipulates control points allowing deformation of the shape of the 2D or 3D object producing a smooth and continuous transformation. This is achieved by maximization of a voxel-similarity measure. Since the intensities across time are closely coupled we are using normalized cross-correlation as similarity measure.
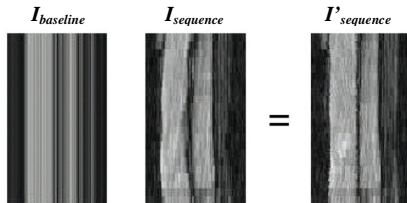

Figure 1. Registration of the image sequence $I_{sequence}$ with the baseline image $I_{baseline}$

In order to obtain the deformation fields we use the 2D image sequence as a 3D (2D + time) image, where *x, y* describe the cross-sectional position and *t* the time during cardiac cycle. The goal of the spatio-temporal registration is to find the optimal transformation $T: (x, y, t) \rightarrow (x, y, 0)$ which maps all points in the 3D image $I_{sequence}(x, y, t)$ to their corresponding anatomical location in the 2D baseline image $I_{baseline}(x, y)$. The image sequence $I_{sequence}(x, y, t)$ is then registered to the baseline image $I_{baseline}$ to find the optimal transformation. Fig. 1 shows an example of the transaxial view baseline image $I_{baseline}$ (first time frame) of the aorta as well as a transaxial view of image sequence that includes all the time frames. Subsequently, to calculate regional area changes the determinant of the Jacobian of the deformation field is integrated over the area of interest to quantify differences between registered images and the reference frame. This would describe the global expansion or contraction ratio of the aorta at that particular time frame. The segmented area in the first time frame is then multiplied by this ratio to obtain the area in each time frame. In this way, not only the actual expansion/contraction ratio values but also the absolute area changes of the vessel throughout a complete cardiac cycle can be visualised and used to estimate the elasticity of the vessel.

EXPERIMENTS AND RESULTS

Images were acquired at Royal Brompton Hospital, London, UK, using a Siemens Sonata 1.5T scanner with a TrueFisp sequence. Images of the proximal, mid and distal parts of the ascending aorta were obtained from six different subjects with a resolution of 192×256 and pixel size around 0.6×0.6 mm with a slice thickness of 6 mm.

The images were taken perpendicular to the aorta. For each subject between 19 and 31 time frames were acquired, with a time resolution of approximately 50msec. The aortic area was computed from the manually segmented sequences and compared to those obtained with the proposed method. Fig. 2 shows the area manually estimated by the expert and by the proposed approach. Table 1 summarises the maximum difference between the manual and automatic estimates of the area. This maximum difference is 5% on average. For volunteer 6 the maximum difference between both areas is 10%, however, the source of this error is due to an inconsistent manual segmentation of the second time frame as can be seen in Fig. 2.

Table 1 Differences between the area estimated manually and automatically for each curve of Fig. 2 divided by the area manually estimated

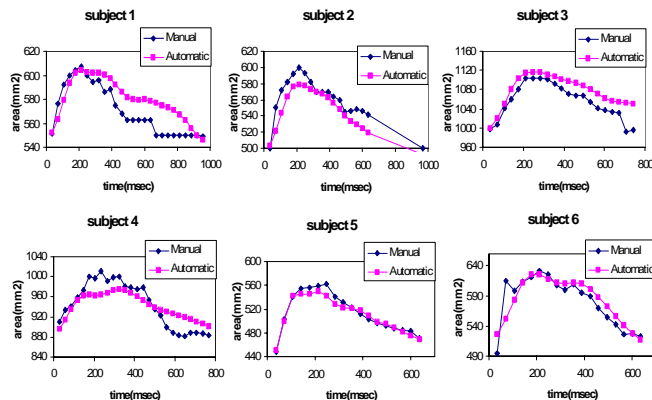| Subject	1	2	3	4	5	6
Max. diff.	4.45%	4.81%	5.51%	4.62%	3.62%	9.92%
Avg.diff.	2.03%	2.27%	2.00%	2.14%	1.09%	2.16%

Figure 2. Example of the manual and automatic estimation of areas

CONCLUSIONS

We have presented a procedure to estimate the transverse area in MR images of the aorta. The 6 examples in Fig.2 show that the area-time curve estimated by the proposed approach is smooth while the manual estimate of the area-time curve is noisy and has some abrupt changes due to the different criteria of the expert to segment the aorta in different time frames. The manual method was very time consuming with 60-100 measurements per section and took approximately one hour while the automatic method took 5 minutes once the program was set up. The automatic contours were consistent with no variation, contrary to the manual double determinations, and appeared more reliable. The method relates the images in time and the results are promising for physiological and clinical studies in the assessment of elasticity of the aorta. At present only one time frame per sequence needs to be segmented manually. Future work will involve the automatic segmentation of the first frame in order to make the approach fully automatic and measurements in other vessels will be performed.

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