Automated image analysis techniques for cardiovascular magnetic resonance imaging

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CHAPTER

4

Automated assessment of MR velocity maps of the ascending aorta: Evaluation of inter- and intraobserver variability in the determination of left ventricular stroke volume by automated and manual analysis methods

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Automated assessment of MR velocity maps of the ascending aorta: Evaluation of inter- and intraobserver variability in the determination of left ventricular stroke volume by automated and manual analysis methods
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ABSTRACT

Objective: An automated contour detection algorithm was developed for the objective and reproducible quantitative analysis of velocity encoded MR imaging studies of the ascending aorta.
Methods: The only user-interaction required is the manual definition of a center point inside the cross-section of the aorta in one of the available images. The automated contour detection algorithm detects an initial model contour in this image and subsequently corrects for motion and deformation of the aortic cross-section in each of the acquired images over the complete cardiac cycle using dynamic programming techniques. Integrating the flow velocity values for each pixel within the detected contour results in an instantaneous flow value. Next, by integrating the instantaneous flow values for each acquired phase over the complete cardiac cycle, left ventricular (LV) stroke volume (SV) measurement could be obtained. The results of the automated method were compared to results derived from manually traced contours in MR imaging studies from 11 healthy volunteers.
Results: An excellent agreement in SV measurements was observed: signed difference 0.61 ± 1.51%. Inter- and intraobserver variabilities were less than 2% for both manual and automated image analysis methods. Manual tracing of contours required in the order of ten minutes; the analysis time for automated contour detection was less than 6 seconds per study.
Conclusion: The present contour detection allows fast and reliable LV stroke volume measurements from velocity encoded MR imaging studies.

4.1 Introduction

Cine phase-contrast magnetic resonance (MR) flow velocity mapping has a proven clinical value in the evaluation of flow in the greater arteries[1-5]. This non-invasive imaging technique allows blood velocity measurements in vessel cross-sections at different points in the cardiac cycle at high temporal and spatial resolution. Application of this technique to the proximal portion of the ascending aorta allows the assessment of left ventricular (LV) systolic function. In the absence of mitral valve insufficiency, the LV stroke volume can be measured by integrating the instantaneous flow values over a complete cardiac cycle[4,6]. In addition, the presence and severity of aortic regurgitation may be quantified by comparing the antegrade and retrograde flow within a cardiac cycle[7,8].

For an accurate assessment of volume flow, contours describing the lumen of the vessel have to be traced in the images. Since the ascending aorta exhibits a significant in-plane and through-plane motion as well as changes in cross-sectional shape over the cardiac cycle, the user is required to trace the vessel border in each individual image of the multi-phase MR examination, thereby carefully avoiding the inclusion of flow in adjacent regions from other vessels. Since this is a time consuming and tedious procedure which introduces observer variabilities, the automation of this process would clearly enhance the clinical applicability of MR flow velocity mapping.

In this study an automated analysis algorithm is presented to be used for the automated detection of vessel boundaries in temporal series of MR flow velocity images of the ascending aorta. The required user-interaction is limited to the manual definition of an approximate center in one of the available images. The algorithm performs an automated detection of the vessel contours and corrects for motion and shape changes of the vessel cross section over the cardiac cycle. The contour detection algorithm was compared to manual tracings of the aortic contours by evaluating the derived stroke volume measurements in MR imaging studies of eleven healthy volunteers. Inter- and intraobserver analysis was performed for both manual and automated image analysis.

4.2 METHODS

4.2.1 Study subjects

The study population consisted of 11 healthy volunteers (4 men) with no history of cardiac disease. Mean age for this group was 37 years (range 25-63). All volunteers were in sinus rhythm during image acquisition and the mean heart rate was 66.6 ± 8.8 beats per minute.

4.2.2 MR examination procedure

MR examinations were performed on a 0.5 T MR scanner (Philips Medical Systems, Best, the Netherlands) using the body coil. Multi-slice spin-echo images orientated in the coronal plane were obtained to identify the orientation and course of the aortic arch. At a position 2-5 cm above the aortic valve, where the aorta was nearly parallel to the caudal cranial axis of the patient, a velocity map was acquired in the axial orientation using velocity encoded cine MR imaging. Imaging parameters for this MR scan were: echo time 8.7 ms, flip angle 45º, repetition time 25 ms, slice thickness 5 mm, field of view 200 x 140 mm, scan matrix 103 x 128, number of averages two, and the velocity sensitivity was set to 150 cm/s. Retrospective gating was applied to acquire images evenly spaced over a complete cardiac cycle resulting in twenty cardiac phases. The acquisition time was 2-3 minutes for the initial spin-echo scan and 2-3 minutes for the velocity-encoded scan depending on the heart rate of the study subject.

4.2.3 Contour detection software

During a velocity encoded MR imaging study, phase difference and standard gradient echo images are acquired at multiple points in the cardiac cycle. Since the gradient echo images show good contrast even in the absence of flow, the automated contour detection algorithm was developed to operate on these images (see Figure 4-1). The actual flow velocity calculations are based on the pixel data in the corresponding phase difference images.

Over a cardiac cycle the position and shape of a vessel cross section may change due to motion of the heart and pressure changes in the artery. In general, for the ascending aorta, the change in shape is relatively little compared to the in-plane motion of the cross-section. The present contour detection algorithm deals with these two types of motion in two separate processing steps. The motion component describes the in-plane motion of the vessel's cross section; the deformation component describes the change in shape or size of the vessel's cross section. To obtain a more general applicability of the detection software, the user may adjust the parameters describing the maximum motion and deformation to be

expected for a certain type of application. In the experiments described in this study the parameters which determine the maximum allowed motion and deformation were kept fixed.

The analysis procedure starts with the detection of a model contour in one of the available images. To this end the user has to indicate an approximate center point of the vessel. This is the only manual interaction required in the total analysis procedure. To detect the contours in the other images within the temporal series, first a correction for vessel motion is performed, followed by a correction for changes in contour shape. The three steps of the automated contour detection algorithm are explained in more detail in the following sections.

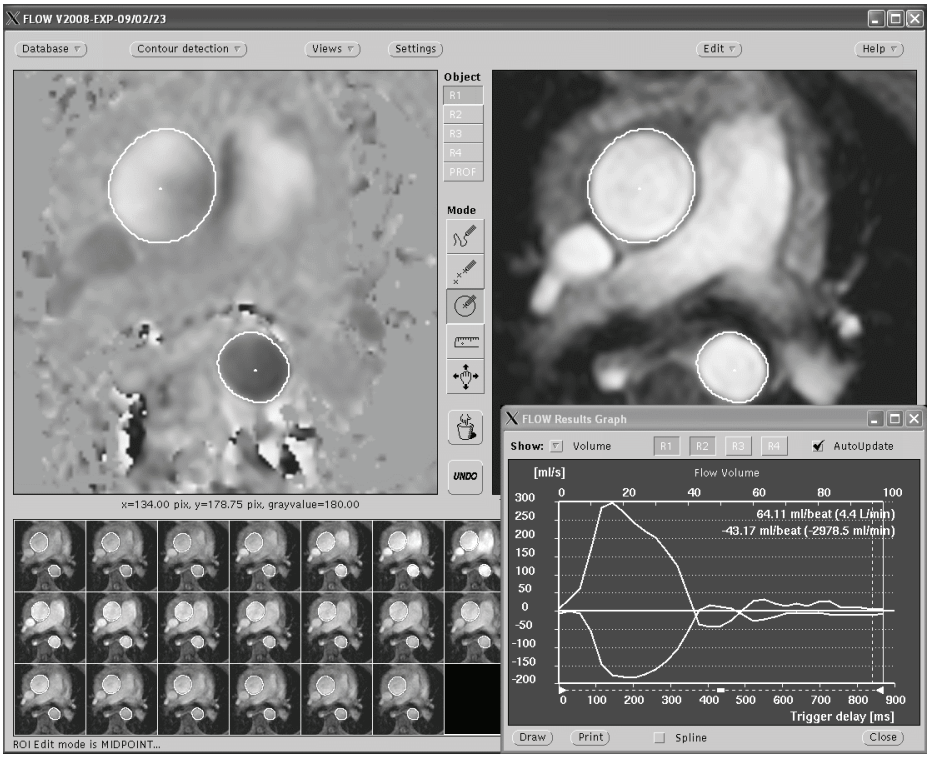


Figure 4-1. Screen lay-out of the software package FLOW. The upper panels display the phase and modulus images of one of the available cardiac phases. A volume graph is shown for the ascending and descending aorta derived from the automatically detected contours.

4.2.4 Detection of a model contour

To detect the vessel contours in a series of images, an initial model contour providing a rough approximation of the vessel boundary needs to be

detected in one of the available images. For this purpose the user has to select the phase in the cardiac cycle where the ascending aorta is shown with maximum contrast in the modulus images. In this image, the approximate center has to be indicated by the user. Following this step, radial scan lines are constructed at evenly spaced angular intervals starting at the center point. For each scan line the pixel with maximum edge strength is recorded. The search for edge pixels is limited to a distance of less than 20 mm from the indicated center point, representing a sufficiently large margin for most of the clinical cases. The size of this search distance can be adjusted manually in case the aorta appears to be enlarged. The mean gray value of these pixels is used as a threshold to obtain a rough estimation of the vessel's cross section. The contour surrounding this segmented area is used as a model contour for the vessel boundary in this image. This contour was deformed to fit on the edges in the image by applying a minimal cost contour detection algorithm, which is based on dynamic programming techniques[9,10]. The resulting contour was then used as a model contour for the next step in the automated contour detection procedure for the other phases of the study.

4.2.5 Motion detection

Given the model contour describing the cross-sectional shape of a vessel in a particular time frame within the cardiac cycle, the position of the same vessel at the other time frames was estimated by shifting the model contour in a limited circular region around the initial location and examining the median edge strength measured in the modulus image along the contour points. The edge strength at a contour point was measured by taking the first derivative in a direction perpendicular to the local contour direction. The computed edge strength values were stored in a motion matrix for each of the locations (dx, dy) evaluated. Entries in the motion matrix with relatively high edge strength values, suggest a probable contour translation. If however, for each time frame the contour would be translated according to the entry in the motion matrix with the maximum edge value, the resulting temporal series of contours would often show unrealistic position changes. The algorithm as depicted in Figure 4-2 was devised to detect a series of contours that move smoothly from phase to phase. For each time frame a two-dimensional cost array was created; each element in this matrix was assigned the inverse of the median edge strength for the corresponding contour location. For the time frame from which the model contour originated, the center element of the cost array was set at cost zero and the other elements at infinity. A closed path with minimal cumulative cost through the series of cost arrays was computed

using dynamic programming strategies[9]. This path represents the displacement of the vessel of all available time frames relative to the position of the model contour. The position of the contour was allowed to move two pixels from frame to frame and 10 pixels from the original model contour position at maximum.

4.2.6 Deformation detection

Given a contour which is an approximation of the vessel boundary resulting from the algorithm which was used for motion correction, a final optimized contour was detected by allowing small deformation of the model contour such that it would follow the edges in the modulus image. For this purpose a two-dimensional graph searching technique was used, often denoted as minimum cost contour detection[10]. In short, for each time frame a rectangular scan matrix is constructed by resampling the modulus image perpendicular to the model contour. From this matrix a cost matrix is constructed by taking for each line at each position the first derivative value at the corresponding position and line in the scan matrix. An optimum closed path with minimum cost through the cost matrix is found resulting in the final contour for the image. Since this contour connects the pixel positions with maximum edge strengths, the resulting contour was dilated by one pixel to account for partial volume effects and to be sure to encompass the complete region with flowing blood.

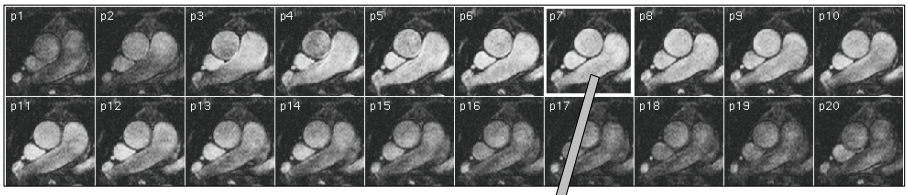
4.2.7 Analysis procedure

Manual and automated image analyses were performed by two independent observers (RvdG, AN). To assess intraobserver variabilities, the first observer repeated the automated and manual analyses after a two week interval. The flow quantification package ran on a commercially available SUN Ultra Sparc 1 workstation (Sun Microsystems, Mountainview, Ca).

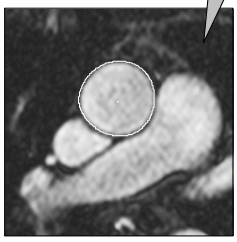
4.2.8 Manual analysis

During manual tracing of contours the observer was allowed to use the phase and modulus image simultaneously. A movie loop showing both phase and modulus images, with superimposed contours was used to facilitate the interpretation of the images. Great care was taken to include all visible flow in the phase images and to avoid regions of flow belonging to the inferior vena cava often lying adjacent to the aorta.

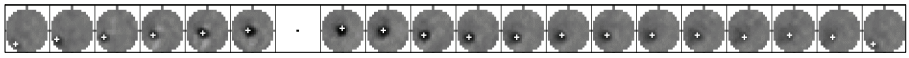
Modulus images for a complete cardiac cycle



Modulus image from phase 7 with model contour automatically detected from center point

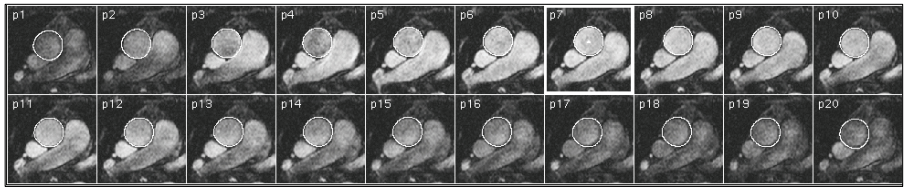


Motion matrices for phase 1 to 20 with optimal contour displacement indicated (+).



1 2 3 4 5 6 7 8 9 10 11 12 13 16 17 18 19 20

Result after motion detection



Result after deformation detection

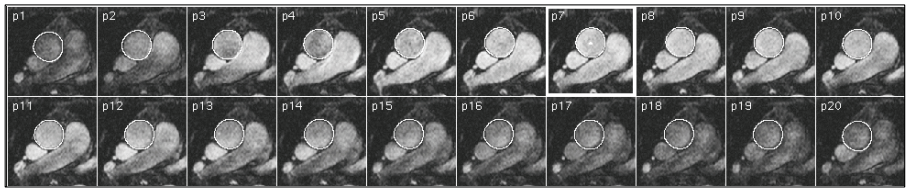


Figure 4-2. Graphical representation of the individual steps of the automated contour detection algorithm for a time-series of images of the ascending aorta.

4.2.9 Automated analysis procedure

The image analysis using the automated contour detection algorithm started with loading of all the phase and modulus images of a study into the flow quantification package. The time frame with optimal vessel depiction in the magnitude image was selected for display. In most cases this time frame corresponded to a cardiac phase in early diastole. In case the magnitude image showed overlap of the ascending aorta with the pulmonary artery, another time frame was selected. In the selected image an approximate center was manually indicated. From this center a contour was detected automatically. In case of a failure of the contour detection for this image, the next or previous time frame was selected to initiate the contour detection procedure. Subsequently, the motion detection algorithm was applied to estimate the translation of the vessel cross section for each time frame. The resulting contours were used as models for the deformation detection algorithm, resulting in optimally adjusted contours for each image. In no case were manual corrections allowed.

In the experiments described in this study, the model contour was allowed to deform ± 4 pixels (3.1 mm) in the radial direction. The maximum motion allowed was set to ± 10 pixels (7.8 mm). In one subject the amount of motion of the ascending aorta was much larger than this value and the maximum allowed motion for this case was set to ± 15 pixels. The total analysis time for automated analysis was approximately 6 seconds. The resulting contours and the location of the manually indicated center point of an analysis session were saved on disk.

4.2.10 Data analysis

From the vessel boundary contours which were manually traced by the first observer the actual in-plane motion and deformation of the vessel boundary over the cardiac cycle was evaluated within the study population. The center of gravity of each contour was computed and the maximum displacement of this point with respect to the end-diastolic time frame was computed. To study the shape changes of the vessel boundary of the aorta over the cardiac cycle, the relative increase of the contour area from the end-diastolic phase (time frame 1) to the time frame with maximum area was computed for each study.

From the automatically and manually determined contours, flow curves were constructed by computing instantaneous flow (i.e. the product of the contour area and mean flow velocity within the contour) at each time frame. From these curves, stroke volume measurements were obtained by integrating the flow over the complete cardiac cycle. Results derived from manual contour tracing were compared to results from automated contour

detection by computing the mean and standard deviation of the paired signed differences. The same measurements were performed to compare repeat measurements of the same observer and between the two observers for both manual and automated analysis. The mean and standard deviation of the paired differences of repeat measurements were expressed as a percentage of the mean. A Student t test was used to test the statistical significance of the differences observed between and within observers. Statistical significance was defined as $p<0.05$.

Table 4-1. Mean values and inter- and intraobserver variabilities for the assessment of stroke volume using either manual contour tracing or automated contour detection. The mean and standard deviations of paired differences of repeat measurements are presented

		Manual	Auto
Luminal area	Mean [cm^2]	7.26 ± 1.56	6.98 ± 1.40*
	Intraobserver difference [%]	-2.08 ± 6.28*	0.34 ± 2.15*
	Interobserver difference [%]	0.73 ± 5.43*	0.75 ± 2.56*
Stroke volume	Mean [ml]	88.14 ± 14.48	87.73 ± 14.49
	Intraobserver difference [%]	-0.23 ± 0.85	0.72 ± 1.25*
	Interobserver difference [%]	0.83 ± 1.59	0.55 ± 1.20

*: indicates statistical significant difference ($p<0.05$).

4.3 Results

On an average the maximum displacement of the contour center relative to the end-diastolic phase was 7.1 ± 2.1 mm (9.0 ± 2.6 pixels); range 5.1-12.0 mm (6.5-15.4 pixels). The average increase in cross-sectional area of the aorta relative to the end-diastolic phase was 24.0 ± 10.9%.

Figure 4-2 shows an example of flow curves of one of the study subjects, obtained by manual and automated contour detection. The mean stroke volume within the study population obtained by manual contour tracing was 88.1 ± 14.5 ml. Figure 4-3 demonstrates that the results of automated contour detection were in excellent agreement with the results from manual image analysis. The mean signed difference between the two methods of contour definition was found to be 0.41 ± 0.77 ml (0.61 ± 1.15%, p=NS), representing a small but non-significant underestimation in stroke volume by the automated contour detection method. In Table 4-1, the results of intra- and interobserver analysis are presented. For both manual and automated image analysis, the reproducibility of stroke volume measurement proves to be excellent. The signed differences and the standard deviations of the differences are always less than 2% of the mean values.

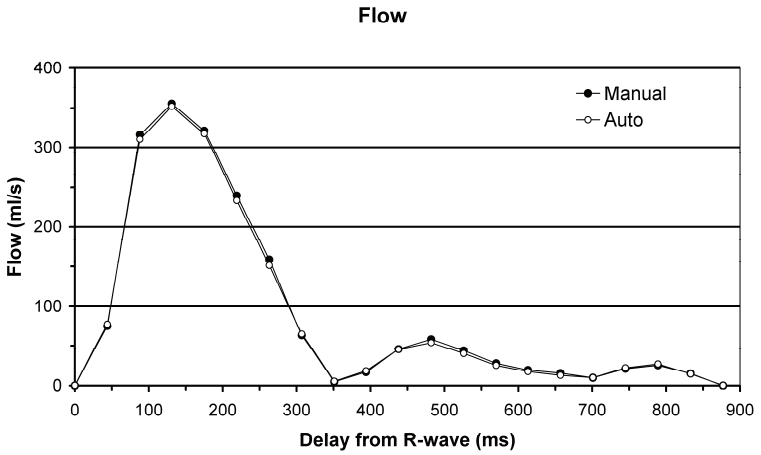


Figure 4-3. Example of flow curves obtained from manually (●) and automatically (o) determined contours. Along the horizontal axis the cardiac phase number is displayed; the vertical axis depicts the instantaneous flow rate.

4.4 Discussion

Velocity-encoded MRI of the ascending aorta provides a non-invasive means of assessing left ventricular stroke volume. The accuracy and reproducibility of this technique has been proven in various studies [6,7]. In the current study a conventional phase-contrast MRI technique was used, requiring 2-3 minutes of acquisition time. Faster imaging techniques which are based on echo-planar, or segmented k-space (multi-shot) techniques make it possible to reduce the imaging time to a single breath hold, and are becoming widely available on most modern MR scanners[11,12]. Quantification of stroke volumes from the acquired MR images requires an accurate delineation of the boundaries of the aortic lumen in each of the images, which is a time-consuming and tedious procedure when performed manually. From the measured maximum in-plane displacement of 7.1 ± 2.1 mm (25 ± 7.3% of the average ED diameter) and the maximum observed change in cross-sectional area of 24 ± 11%, corresponding to a change in diameter of about 3.19 mm or 4.1 pixels, it is evident that the contours to be drawn need to be repositioned and adjusted for each individual phase. In the current study the number of acquired phases was only twenty, but still the time required for manual contour tracing was in the order of ten minutes. If this technique is to be used for clinical purposes, it would be beneficial to reduce the analysis time. The present contour detection algorithm performs the automated detection in less than 6 seconds and is in excellent agreement with manually defined contours. Also, both the

inter- and intraobserver variabilities of the automated contour detection of less than 2% of the mean stroke volume demonstrate the robustness of this new approach. The variability of the measurements is explained by the required operator interaction, which is the manual definition of the aortic center in one of the available images. The contour automatically derived from this center, serves as a model (example) for the automated detection in the other phases within the study. Since in this study the operator was free in choosing in which image the center was indicated, some variability will be introduced in both the motion estimation step and the final deformation detection due to differences in the model contour.

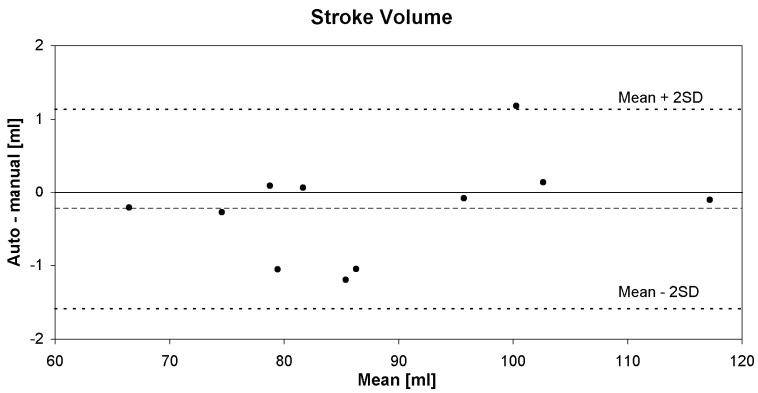


Figure 4-4. Differences in stroke volume measurements obtained by manual contour tracing and by automated contour detection as a function of the mean stroke volume.

4.4.1 Limitations

Since the automated contour detection algorithm exclusively uses the information from the gradient echo modulus images for the determination of the vessel boundaries, the detected contour may in some cases deviate from the visible flow in the corresponding phase image. Since the blood velocity near the vessel border normally is much lower than in the central region of the vessel, this resulted in no significant error in stroke volumes. Further improvement in the contour detection may be obtained by combining the information from the phase and modulus images simultaneously. However, in many clinical situations the information in the phase image may be misleading since the flow may be zero during most of diastole, and forward and backward flow may exist simultaneously.

In the present study, the new automated contour detection technique was compared to manual contour tracing by comparing just one quantitative parameter (SV). Additionally, no imaging studies from cardiac patients were included in the study subjects, and therefore the image

quality may have been relatively good. It is suggested that a more extensive validation study be performed on a wider variety of patients, including those with aortic regurgitation and aortic valve stenosis, to fully assess the clinical value of the presented automated contour detection algorithm.

The simple fact that a good agreement in SV was found does not necessarily mean that the manual and automated derived contours were nearly identical since the inclusion of regions of stationary tissue or blood in a contour does not affect the stroke volume measurements. However, in the case of the ascending aorta, many regions of flow in different directions are present in the neighborhood of the ascending aorta which definitely would have affected the results. By visual inspection of the manual and automatically detected contours in a cine mode, it could be noted that the automatically detected contours were somewhat more irregular in shape in those phases with little or no flow in the ascending aorta. Since in these phases the velocities in the neighboring regions are also close to zero, this introduced no significant errors in the stroke volume measurements.

4.5 Conclusion

In conclusion, an automated contour detection algorithm for the automated assessment of stroke volume from velocity-encoded MR of the ascending aorta has been presented and was validated on MR studies of 11 volunteers. When compared with results by manual analysis, the algorithm has proven to be 50 times faster, while the agreement was excellent and the inter- and intraobserver variabilities were less than 2%. Therefore, the described automated contour detection algorithm will contribute to the clinical applicability of MR velocity mapping in the evaluation of flow in the ascending aorta.

4.6 References

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