

Ewa Pietka

Jacek Kawa

Wojciech Wieclawek *Editors*

# Information Technologies in Biomedicine, Volume 3

# **Advances in Intelligent Systems and Computing**

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Janusz Kacprzyk, Polish Academy of Sciences, Warsaw, Poland  
e-mail: kacprzyk@ibspan.waw.pl

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Ewa Pietka · Jacek Kawa  
Wojciech Wieclawek  
Editors

# Information Technologies in Biomedicine, Volume 3

 Springer

*Editors*

Ewa Pietka  
Silesian University of Technology  
Faculty of Biomedical Engineering  
Gliwice  
Poland

Wojciech Wieclawek  
Silesian University of Technology  
Faculty of Biomedical Engineering  
Gliwice  
Poland

Jacek Kawa  
Silesian University of Technology  
Faculty of Biomedical Engineering  
Gliwice  
Poland

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# Preface

In clinical application we deal with problems that have to be solved in a fast and objective way. However, human observation is influenced by internal (coming from the observer) as well as external (often independent from the observer) impacts. The objectivity of classification is restricted by the receptivity of human senses which are influenced by the experiences or level of training, psychological conditions (tiredness, haste, etc.), as well as external conditions (lighting, destructive noise, etc.) A failure in perception questions the entire recognition process. The recognition process itself, influenced also by the above mentioned conditions, may cause a slowdown and/or lead to a false diagnosis.

New computerized approaches to various problems have become critically important in healthcare. Computer assisted diagnosis has been extended towards a support of the clinical treatment. Mathematical information analysis, computer applications together with medical equipment and instruments have become standard tools underpinning the current rapid progress with developing Computational Intelligence. We are witnessing a radical change as technologies have been integrated into systems that address the core of medicine, including patient care in ambulatory and in-patient setting, disease prevention, health promotion, rehabilitation and home care. A computerized support in the analysis of patient information and implementation of a computer aided diagnosis and treatment systems, increases the objectivity of the analysis and speeds up the response to pathological changes.

This book aims to present a variety of state-of-the-art information technology and its applications to the networked environment to allow robust computerized approaches to be introduced throughout the healthcare enterprise. Image and signal analysis are the traditional parts that deal with the problem of data processing, recognition and classification. Bioinformatics has become a dynamically developed field of computer assisted biological data analysis. Patients' safety and shortening of the rehabilitation time requires a more rapid development of minimally invasive surgery supported by image navigation techniques. Home care, remote rehabilitation assistance, safety of the elderly require new areas to be explored in telemedicine and telegeriatics.

This book set is a continuation of a book series. This set contains two volumes. *Information Technologies in Biomedicine, Volume 3* discusses Image analysis techniques and their applications in healthcare, as well as some Bioinformatics issues. *Information Technologies in Biomedicine, Volume 4* consists of six parts including Computer Aided Surgery, Telemedicine, Telegeriatics,

We would like to express our gratitude to the authors who contributed their original research papers as well as the reviewers for their valuable comments.

Ewa Pietka

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Part I

Image Analysis and  
Applications

# Applications of Ray-Casting in Medical Imaging

Maciej Borzęcki<sup>1</sup>, Adam Skurski<sup>1</sup>, Marek Kamiński<sup>1</sup>, Andrzej Napieralski<sup>1</sup>,  
Jarosław Kasprzak<sup>2</sup>, and Piotr Lipiec<sup>2</sup>

<sup>1</sup> Department of Microelectronics and Computer Science, Lodz University  
of Technology, Łódź, Poland

mborzecki@dmcs.pl

<sup>2</sup> Department of Cardiology, Medical University of Lodz, Łódź, Poland

**Abstract.** The authors present applications of ray casting as segmentation and analysis method for processing of medical imaging data. The first application features ray casting based image segmentation for extraction of a region enclosing heart structures from a series of CT scans. Proposed method yields significant gains in reduction of the data set size, that are of importance in applications such as Transesophageal USG simulations on mobile devices or web platforms.

Another application, utilizes ray casting determining location of characteristic points of *left ventricle* (LV). The points are used as reference during automatic fusion of ECHO Automated Function Imaging output with a 3D model of LV.

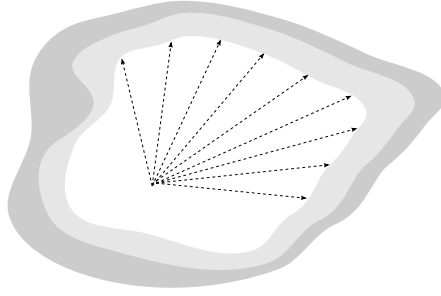
**Keywords:** ray casting, image segmentation, computed tomography.

## 1 Introduction

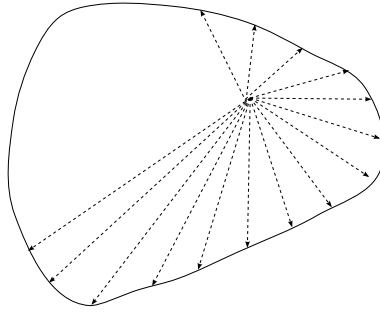
Ray casting is a method that has seen wide use in a plethora of applications in different fields. Object picking within a 3D scene or collision detection are examples of traditional uses of ray casting [1, 2]. The method has also been successfully applied in the field of image processing for segmentation and extraction purpose. Example applications as described in [3, 4], indicate a level of success when method is applied for segmentation of medical imaging.

The core idea behind use of ray casting in image segmentation or feature location is largely unchanged from the typical approach. Virtual rays are emitted in a number of directions from a single origin point. For each iteration, a boundary condition is evaluated, indicating whether for the current location of the tip of the ray, the propagation can continue. In case of image segmentation, the boundary condition is typically a function testing if given point is within the desired data set. The concept is briefly visualized in Fig. 1.

Once the points where the ray propagation has ceased are established further steps are taken in order to create a mask or an outline of the identified structure. Typically employed methods are neighbor contour tracing [5] or construction of polygon by connecting the end points.



**Fig. 1.** Ray casting for image segmentation



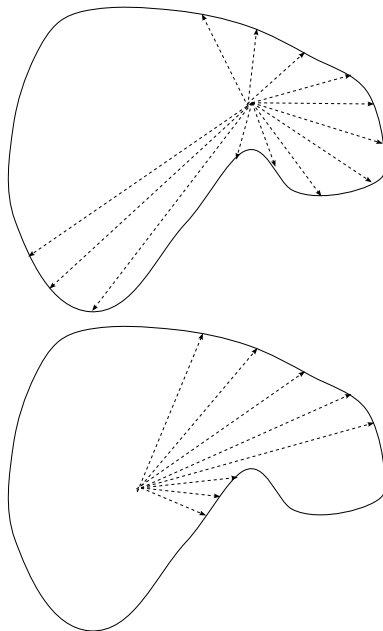
**Fig. 2.** Ray casting within a convex shape

Ray casting as a method is a subject to a number of problems that need to be tackled. The first problem is the shape of the object under consideration. As shown in Fig. 2, convex shapes usually yield good results.

For concave shapes, such as one shown in Fig. 3, location of ray origin point poses a problem. Misplaced origin will amplify the shadowing, the effect of which needs to be taken into consideration.

If the structure being extracted is relatively large, one needs to consider how many rays need to be cast in order to obtain a reasonably good set of points for analysis. The problem is caused by the fact that only a certain number of rays is practical for most applications. In case of image segmentation, too few points may negatively affect the process of establishing object contour.

In the following sections the authors discuss ray casting as a method for processing medical imaging data, where Section 2 describes application in image segmentation, and Section 3 covers automated location of features in a 3D model.



**Fig. 3.** Ray casting origin location for concave shapes

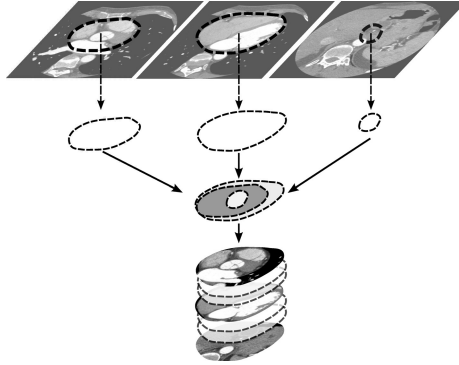
## 2 Heart Region Extraction for USG Simulations

The problem of ultrasound simulation has gained wide interest starting from fundamental work based on physical models [6], towards medical applications such as [7, 8]. With advent of cheap, multi-core processing the simulation algorithms have been successfully applied for GPGPU as in [9] and [10]. At the same time new simplified, but effective, processing algorithm such as [11] were proposed. Applications such as [12, 13] conform to the trend of utilizing simulations as a training tool in medicine that right now is considered a necessity [14, 15].

Simulation of Transesophageal Echocardiography commonly makes use of medical imaging collected during patient examination. A TEE simulator applies online algorithms transforming an input data set obtained in CT examination into a simulated USG image. This section discussed a ray casting based input data preprocessing method that provides a significant size reduction of the input data.

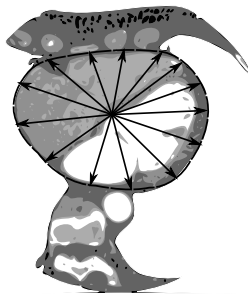
Given the physical properties of the USG imaging, the input needs to contain only the heart image and the position of the esophagus. Neighboring organs can be discarded without losing the educational value of the program. A raw data set of size 1-2GB is common for CT scans of a chest. Inclusion of temporal information, so that depiction of heart's cycle is possible, increases it's size significantly. Reduction of the input set size enables latency and processing limited applications to gain wider use.

The authors propose a ray-casting based method for segmentation of CT scan image and *Region of Interest* (ROI) extraction for the purpose of TEE simulation [16]. The method is based on generation of ROI mask for each of the relevant images in the input data set. The next step is superpositioning of all the masks, so that an aggregate mask enclosing all identified ROIs is obtained. The aggregation step allows for generating a 3D volume that is effectively encloses only the data that is relevant for the simulation process. The process is shown in Fig. 4.



**Fig. 4.** Processing pipeline - segmentation of CT scan series, preparation of aggregate mask, ROI volume extraction

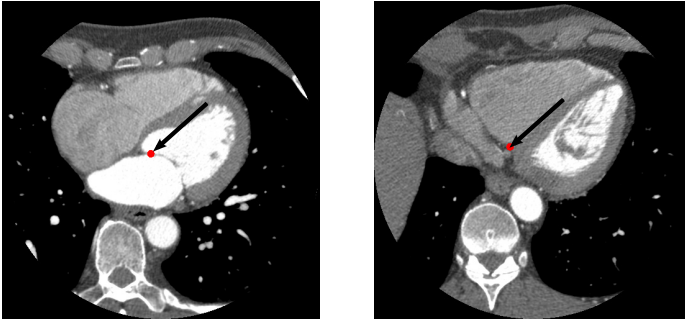
The concept for heart image segmentation is presented in Fig. 5. The threshold condition for ray propagation is set such that the ray should stop at the pericardium.



**Fig. 5.** Heart muscle outline using ray casting

A simple heuristic is used for selection of the origin point based on locating the center of mass of the image. Presence of contrast agent results in areas filled by blood exhibiting higher pixel values in CT images. Heart structures - ventricles,

atria, muscle tissue are thus well visible. Given that, the center of mass will be found inside the area enclosed by heart. The resulting position for 2 samples is shown in Fig. 6.



**Fig. 6.** Ray origin positioning

The threshold value for attenuation was empirically found to be  $20HU$  and roughly corresponds to the values expected for area occupied muscle tissue. The value was confirmed by examining a number of data sets, however all images were obtained using the same CT device. The authors strongly suggest to perform additional verification and examine data collected using a particular CT scanner. The Hounsfield value is converted to pixel brightness based on metadata stored in DICOM [17, 18]. Lower values indicate lower measured attenuation, most likely the ray has entered pericardial cavity or lungs.

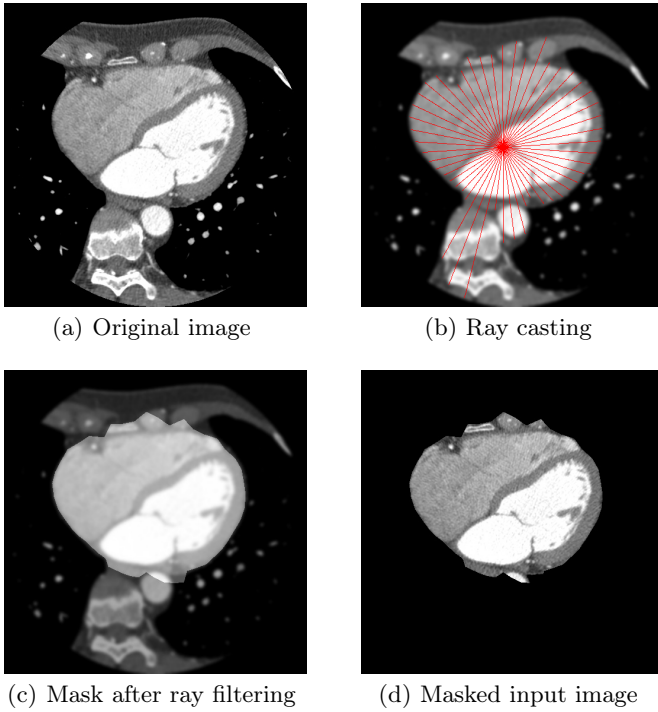
The result of ray casting is shown in Figure 7(b). One can clearly observe that at the sides, most of the rays stopped at pericardium, mainly due to the large difference in pixel intensity, and thus attenuation, between lungs and heart. However in anterior and posterior regions, the rays propagated farther than desired.

The boundary condition may not be met as the intensity level at the position where the ray propagation should cease is higher than expected. This may be caused by too large ray increase step, amplified by deficiency of the imaging method or existence tissue with similar properties in direct neighborhood. This problem of *mask leakage* can be limited by post processing ray casting points.

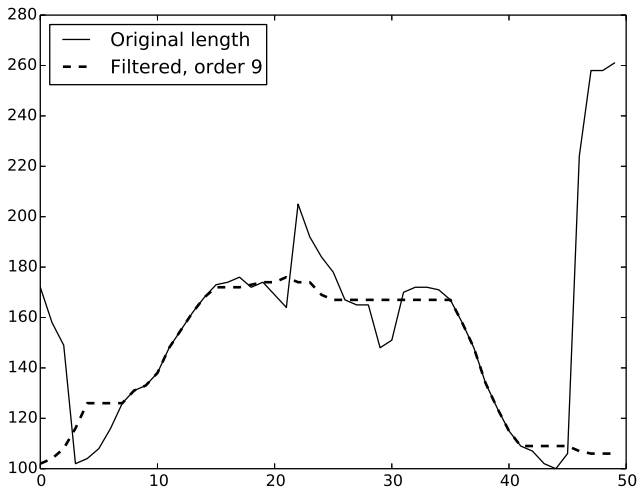
The heart can be considered to have a smooth surface, hence any abrupt changes in ray length are improbable and indicate mask leakage. The problem is displayed in Fig. 8. Significant change in ray length is visible both at the center and at the edges of the graph correspond to rays propagating towards ribs at the top and towards aorta at the bottom in Fig. 7(b). Application of a median filter with empirically selected window length alleviates the problem. The resulting ray lengths are shown in Fig. 8 using dashed line. The final shape of the mask obtained in segmentation process is displayed in Fig. 7(c).

Evaluation has been performed by verifying if the resulting masks completely enclose outlines of most significant heart regions (ventricles, atria, pericardium).



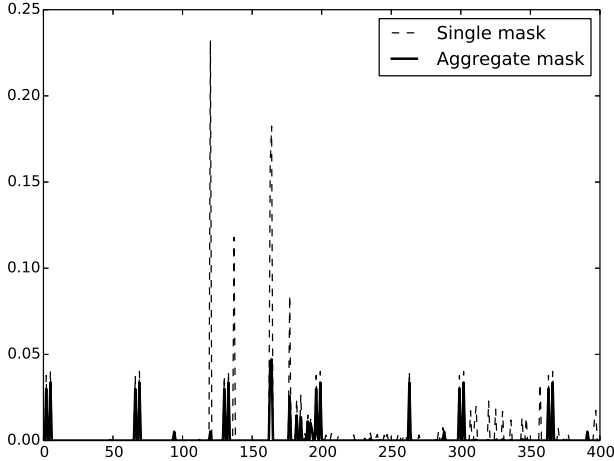


**Fig. 7.** Segmentation using ray casting



**Fig. 8.** Consecutive ray lengths from Fig. 7(b) (starting at bottom center, counter-clockwise), before and after filtering

For this purpose a test data with an outline of heart muscle was prepared. The obtained segmentation masks, were then verified to completely contain the reference regions. The graph showing comparison of the obtained mask to the reference area is shown in Fig. 9.



**Fig. 9.** Measured reference area not included in the mask, relative to mask size

There is a clear difference between the results for individual mask and an aggregate mask. Clearly, single masks have disadvantage in case the ray casting process failed to perform segmentation correctly. However, as assumed previously, the effect is compensated by use of aggregate mask. For all of reference samples, the missing area is below 5% of the mask size. Further analysis revealed that the missing area is at the edge of pericardium and does not impede the final outcome of the segmentation process. It may be expected that inclusion of a dilation step will yield an improvement and cover the missing regions.

After compression the original data set size has been reduced by factor of 18. Further steps that employed ray casting for image segmentation and extraction of the region of interest resulted in another size reduction by a factor of 2.5. Thus a single slice stored in 16-bit PNG format has been downsized to 50kB of data. At the same time the region that needs to be processed during the simulation pass has been marked and reduced significantly, as the relative area occupied by the heart is at most, less than 40% of a single horizontal slice.

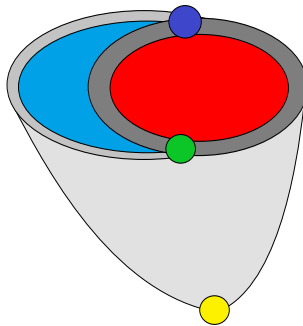
### 3 Automated Identification of Left Ventricle Characteristic Points

Fusion of medical data is a widely regarded practice that aims at improving the diagnostic value individual imaging methods by providing a combined interpretation, thereby revealing indirectly visible aspects of the medical data.

Fusion of morphological data obtained in CT and functional information from SPECT (perfusion at rest and stress) allows for comprehensive evaluation of location and severity of ischemia, thereby increasing the diagnostic and prognostic value of noninvasive imaging techniques [19, 20].

Another application of ray casting plays a significant step in a fusion of ECHO AFI (Automated Function Imaging) [21] output with a 3D model of LV[22]. Combination of a bull's-eye representation of LV [23] stretched on a 3D mesh extracted from a series of CTA (CT Angiography) scans is vital from diagnostic point of view. The method may lower the need for performing classical, invasive angiography or can augment the analysis and interpretation of unclear CTA results. The result of the fusion process is presented in a single view, where the measured LV function is correlated spatially with a 3D model of LV [22].

Successful fusion requires knowledge of characteristic points on the left ventricle that allow for proper positioning of the bull's-eye diagram as a texture. Given the spatial reference of bull's-eye diagram, at least three distinct points need to be found on a 3D mesh. Apex and two additional points at the base of LV, near interventricular septum, were identified as sufficient, see Figure 10.



**Fig. 10.** Characteristic points of left ventricle: apex, two points at the base, near interventricular septum

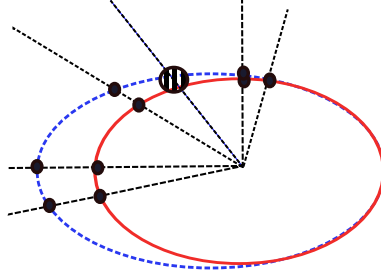
A 3D mesh, automatically extracted by CT software does not directly contain the required information. The authors propose a ray-casting based method for location of two characteristic points at the base of LV, both points marked in Figure 10 with blue and green markers.

Each of the points is found by evaluating the conditions that refer to the locality of certain features in the 3D mesh.

The blue marker is found by iteratively casting rays, while rotating towards the back of the heart, and measuring distance between pericardium and LV wall. With each iteration, the distance is compared with previous value. Conducted analysis showed that at least 25% increase in distance is a sufficient estimate. Once, the condition has been met, the characteristic point is known to be located at the previous ray position.

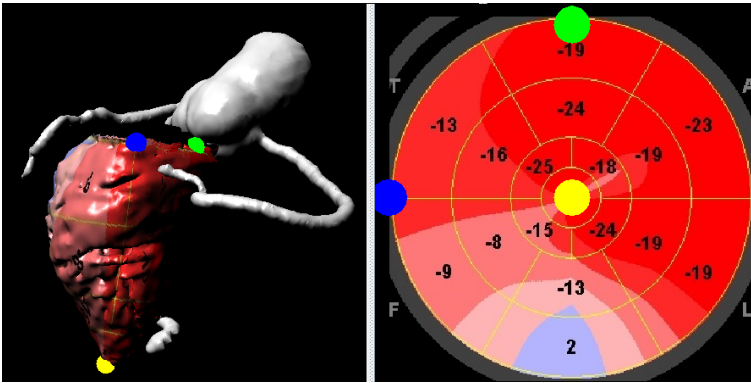
The green marker is found using a similar approach. However, the distance is measured between the wall of LV and aortic sinus. The point of interest is located on the ray for which distance measurement was the smallest.

The concept is visualized in Figure 11.



**Fig. 11.** Locating points of interest (marked with dashed circle) using ray casting

Presented implementations of modified ray-casting method for tracking LV anatomical points of interest provide the opportunity to define a constellation of markers essential for the images fusion performance, presented on Fig. 12.

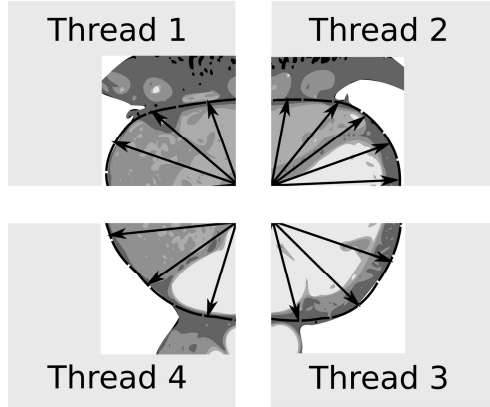


**Fig. 12.** Fusion of 3D model of LV with ECHO

## 4 Improvements

While ray casting as algorithm do not provide room for improvement, the respective implementations can be enhanced for greater performance. The algorithm may be easily extended to support a multi-threaded implementation. An inherent data level parallelism allows for work partitioning at the input image level. Lack of dependency between respective rays, allow an approach like in Fig. 13.

The profiling analysis revealed that a large portion of execution time is spent in the process of ray casting, with only little time allocated to ray length filtering, which is a sequential process. Thus, according to Amdahl's law [24, 25], splitting of ray casting task between a number of threads should yield a considerable speedup.



**Fig. 13.** Parallel ray casting implementation

Another improvement that is possible to apply is to exploit the aliasing effect of the ray propagation near the origin. The input image, as indicated in the introductory section, has a limited resolution. If a large number of rays is used, each ray will, at least for a certain length, starting from the origin, trace the same path as its neighbors. The authors propose a use of a look-up table, addressed by coordinates of each pixel. The table would hold a binary flag indicating if a pixel was already claimed by a ray. Since all rays are checked against the same boundary condition, the flag allows for pixels that were already tested to be skipped.

## 5 Conclusions

The authors have presented two successful applications of ray casting for processing of medical imaging data. Although the core method is well known, its application and enhancements proposed by authors bring a deal of novelty.

Application of ray casting in the segmentation process, although already providing good results, still needs some improvement. It has been observed that for certain samples, not all of the reference area is included within the segmentation mask. Especially at points where transition between pericardium to lungs appears, the segmentation process may partially fail, with ray not propagating far enough. It is expected that the problem can be easily addressed with a dilation step, however this would result in unnecessary increase of the whole mask.

Ray casting implementation for locating characteristic points of LV is a part of software package for supporting CAD (coronary artery disease) diagnosis. Purpose of presented algorithm is to prepare mapping for final texturing step in a fully automated manner. Should the algorithm fail or yield inadequate results, the operator has a possibility to move the markers freely, using the found positions as starting points. Tests in clinical environment using non-pathologic samples are in progress and show promising results. Still the authors have not yet verified the performance of the algorithm in case the input data set contains anomalies such as hypertrophy of the left ventricular wall.

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# A New Aortic Aneurysm CT Series Registration Algorithm

Joanna Czajkowska<sup>1,2</sup>, Christian Feinen<sup>2</sup>, Marcin Grzegorzek<sup>2</sup>,  
Matthias Raspe<sup>3</sup>, and Ralph Wickenhöfer<sup>4</sup>

<sup>1</sup> Silesian University of Technology, Faculty of Biomedical Engineering,  
Department of Informatics and Medical Equipment, Zabrze, Poland

<sup>2</sup> University of Siegen, Institute for Vision and Graphics,  
Research Group for Pattern Recognition, Siegen, Germany

<sup>3</sup> SOVAmed GmbH, Koblenz, Germany

<sup>4</sup> Herz-Jesu-Krankenhaus, Dernbach, Germany

joanna.czajkowska@polsl.pl

**Abstract.** Nowadays, vascular diseases are the most challenging health problems in developed countries. Despite the fast development of modern contrast-enhanced Computed Tomography (CT), providing complex 3D datasets, the tremendous amount of problems still remain unsolved. The vascular segmentation as well as registration techniques are the topics of past and on-going research activities. In this work we focus on an abdominal aortic aneurysm registration technique. The developed approach makes it possible to match all voxels belonging to the aorta from pre- and post-operative CT data. The presented technique is based on aorta lumen segmentation and graph matching method. To segment the lumen area a hybrid level-set active contour approach is used. The matching step is performed based on a path similarity skeleton graph matching procedure. The registration results have been tested on the database of 8 patients, for which two different contrast-enhanced CT series were acquired. All registration results achieved with our system and verified by an expert prove the efficiency of the approach and encourage to further develop this method.

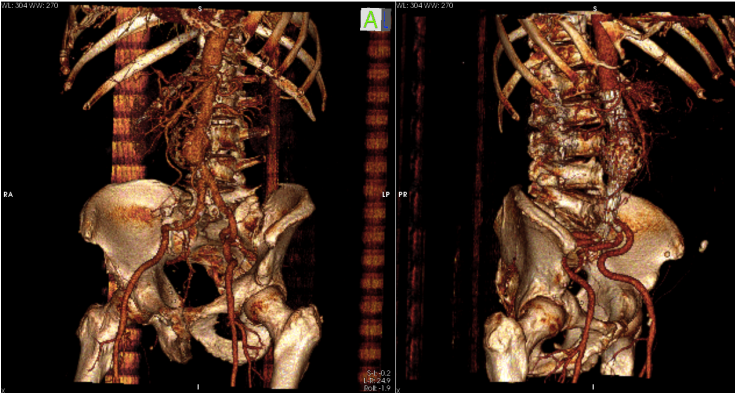
**Keywords:** Hybrid Level-Set Active Contour, Graph Matching, Image Registration, Skeletonization.

## 1 Introduction

Nowadays, vascular diseases belong to the most challenging health problems in developed countries. An abdominal aortic aneurysm (AAA), addressed in our approach, is a dilated and weakened segment of the abdominal aorta. It is an abnormal ballooning of the abdominal portion of the aorta, that occurs as a consequence of aortic medial degeneration and can break open causing death. An AAA can develop in anyone, however it is mostly seen in males over 60, having one or more risk factors. During last 30 years, the occurrence of AAA has increased threefold. To prevent from rupturing, interventional radiologists offer



minimally invasive treatment for abdominal aortic aneurysm, which is specially important when it reaches 5 centimeters in diameter. Currently, there are different AAA treatment options. The open surgical repair by a vascular surgeon is the most commonly used for a large, unruptured aneurysm. The less invasive and relatively new technique, eliminating the need for a large abdominal incision, is placing a graft within the aneurysm. It redirects blood flow and stops direct pressure from being exerted on the weak aortic wall [2, 1]. An exemplary 3D volume rendering of CTA series of pre- and post-operative study created using Osirix software is shown in Fig. 1.

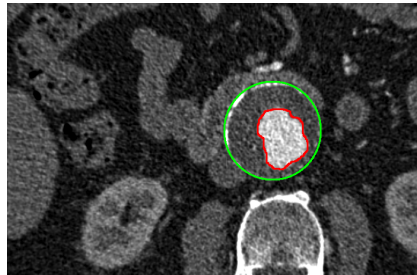


**Fig. 1.** 3D volume rendering of CTA series of pre- (left) and post-operative (right) study of aortic aneurysm provided by Osirix software

The AAA is mostly diagnosed by a physical examination as a soft mass in the abdomen. For more accurate and efficient diagnosis the development of imaging techniques provides numerous tools used to examine vessels and display their details. A contrast-enhanced CT angiography (CTA), which replaced a conventional angiogram, is an imaging technique commonly used in vascular diagnosis. Despite the fast development of modern contrast-enhanced Computed Tomography (CT), providing complex 3D datasets, the tremendous amount of problems still remain unsolved. The vascular segmentation [3] and registration techniques are the topics of past as well as on-going research activities.

From the medical background of AAA, two region of interests: aorta lumen and thrombus, can be defined. An exemplary manual segmentation results of both of them in a CT scan are shown in Fig. 2. The newest approaches in AAA segmentation [3, 5, 6] address either both of the problems or only a chosen one. An automated method for the segmentation of thrombus in abdominal aortic aneurysms from CTA data is presented in [5]. The Active Shape Model (ASM) fitting is performed in sequential slices. As the starting point for the analysis the results obtained for the adjacent slice are used. The full 3D segmentation technique in CTA is reported in [6]. The system analyses both global features,

incorporating a priori knowledge of the intensity, volume, and shape of the aorta and other structures, and the local information like voxel location, intensity, and texture information. All of them are used for training and driving a support vector machine classifier. As reported in [3] the current state of the art in AAA segmentation is modelling, feature analysis or their combination, and in all these areas different efficient techniques can be found. However, the authors also claim, that there are some problems, which still remain unsolved. There is no standardly accepted databases and validation criteria for most vascular segmentation applications and the direct performance comparisons of the segmentation results have not been performed yet.



**Fig. 2.** An example of an contrast-enhanced CT scan of an abdominal aortic aneurysm with segmented thrombus (outer green contour) and lumen (inner red contour)

Despite the fact that the segmentation of vascular structures is valuable for diagnosis assistance, treatment and surgery planning the currently developed computer aided diagnosis (CAD) software target in efficient image registration. It does not only allow measurements of lumen or thrombus volume, but combining different image information is also useful for treatment planning and monitoring. Thanks to it, the comparative analysis of consecutive (pre- and post-operative) CTA studies as well as matching of different image modalities is possible.

Depending on the application, various registration techniques have been reported [7–11]. The registration methods, which address the problem of simultaneous analysis of different image modalities are given in [7–10], whereas the pre- and post-operative CTA sequence matching algorithm is presented in [11]. The authors of [9] propose a registration technique based on the overlaying the preoperative 3D model of the aorta onto the intraoperative 2D X-ray images. The presented technique utilizes two X-ray images showing the abdominal aorta from different angles in an integrated way. They developed a hierarchical registration scheme deployed by a sensible partition of the registration parameter space based on the image acquisition protocol and the patients motion constraint.

The 2D/3D registration technique is also addressed in [7]. The non-rigid method enables information from the CT to be overlaid onto the fluoroscopy images during the implantation procedure. The authors have investigated the

use of manually picked landmarks and the thin plate spline algorithm to deform the CT surface so it more accurately represents the interventional scene. The automatic movement compensation in 2D/3D registration of fluoroscopy and preoperative volumetric data is presented in [10]. The paper proposes a pelvis boundary detection method that enables real time monitoring of patient movement and an automatic 2D/3D re-registration algorithm that compensates for it.

The idea of a graph-based approach in this context is presented in [8]. The introduced 2D/3D registration method is there formulated on a 3D graph and applied for AAA interventions. As an input, the algorithm takes the 3D graph generated from a segmentation of the CT volume and the 2D distance map calculated from the 2D X-ray image. For computing the graph similarity, different measures are then used in a length preservation and a smoothness regularization term.

In this work we focus on 3D abdominal aortic aneurysm registration technique. The developed approach makes it possible to match the aorta segmented in pre- and post-operative CTA data. The presented technique is based on an aorta lumen segmentation and graph matching technique. In the segmentation step a hybrid level-set active contour approach is employed. The applied hybrid medical image segmentation method in the level-set framework [12] uses the object's boundary as well as region information. In this approach a boundary gives the information concerning object location, whereas the region features help to avoid the boundary leakage. The matching step is performed based on a path similarity skeleton graph matching procedure introduced in [21].

In the following section, a short introduction to the hybrid active contour approach [12] applied for AAA lumen segmentation is given. In Section 3, the 3D skeletonization algorithm for graph extraction is described. Section 4 introduces the graph matching technique and Section 5 presents the experiments and the obtained results. Then, the last section (Section 6) concludes the work and outlines plans for the future.

## 2 Abdominal Aortic Aneurysm Segmentation Method

The active contour model for image segmentation was originally developed by Kass et al. [13] and the energy minimization techniques in image segmentation have attracted researches in the last two decades. The basic idea of the snake method [13] is to iteratively evolve the initial contour towards the regions described by some certain features. The movement of the energy minimizing-spline is guided by the geometry of the evolving curve (internal force) and influenced by image features (external force). The image information pull the contour into the lines, edges or terminations. Local minima of the contour energy correspond to desired image properties. Since the classical implementation of the snake method [13] was introduced, different modifications and improvements dictated by its new applications were incorporated. Active contour model expanded by gradient vector flow is presented in [14]. This efficient algorithm remains limited

to the segmentation of structures with well defined contours. Geodesic active contour model [15] and Chan-Vese approach [16] are two most important techniques improving this segmentation technique standing out for boundary-based and region-based methods.

In all the cases mentioned above, the image segmentation methods are based on minimizing a predefined energy functionals. To solve the curve evolution partial differential equations (PDEs) different numerical methods are applied. In a classical approach the finite set of contour points approximates the parametric contour  $\Psi(s) = (x(s), y(s))$ ,  $s \in [0, 1]$  and the contour changes in time  $\Psi(s, t) = (x(s, t), y(s, t))$ ,  $s \in [0, 1], t \in \mathbf{R}_+ \cup \{0\}$  - the contour evolves with the points movement. Geodesic [15], Chan-Vese [16] and the hybrid active contour [12] used in this paper belong to the models developed in a level-set framework. In the implementation [15] the solution of the particular energy snakes model is given by a geodesic curve in a Riemannian space, being induced from the image. The Chan-Vese active contour model [16] is based on the mean curvature motion. The curve  $C$  is implicitly represented via a Lipschitz function  $\phi$  and by  $mathit{C} = \{(x, y) | \phi(x, y) = 0\}$ . The initial contour is defined by the set  $\{(x, y) | \phi_0(x, y) = 0\}$  and the evolution of the curve is given by the zero-level curve at time  $t$  of the function  $\phi(t, x, y)$ . To solve the PDE the curve  $C$  evolves in normal directions with the speed  $F$  [16]

$$\frac{\partial \phi}{\partial t} = |\nabla \phi| F, \phi(0, x, y) = \phi_0(x, y). \quad (1)$$

## 2.1 Hybrid Level-Set Method

The same as in the active contour given in [16] in the hybrid technique [12], employed in our work, the active contour  $C$  is represented by the zero set of embedding function  $\phi$ , such that  $C = \{x | \phi(x) = 0\}$ . The points inside and outside the contour have positive and negative  $\phi$  values, respectively. The minimized functional in image  $I$  domain  $\Omega$  is defined as

$$E(\phi) = -\alpha \int_{\Omega} (I - \mu) H(\phi) d\Omega + \beta \int_{\Omega} g |\nabla H(\phi)| d\Omega, \quad (2)$$

where  $g = g(|\nabla I|)$  is a boundary feature map related to the image gradient. The parameters  $\alpha$  and  $\beta$  balance the two terms of (2), and  $\mu$  indicates the lower bound of the gray-level of the target object. Thanks to it, the curve evolves to enclose the regions greater than  $\mu$ . The PDE of the functional (2) is derived from the Gateaux derivative gradient flow [12]

$$\phi_t = |\nabla \phi| \left[ \alpha(I - \mu) + \beta \operatorname{div} \left( g \frac{\nabla \phi}{|\nabla \phi|} \right) \right], \quad (3)$$

and the explicit curve evolution PDE is represented by [12]

$$C_t = \alpha(I - \mu) \vec{N} - \beta \langle \nabla g \cdot \vec{N} \rangle \vec{N} + \beta g \kappa \vec{N}. \quad (4)$$

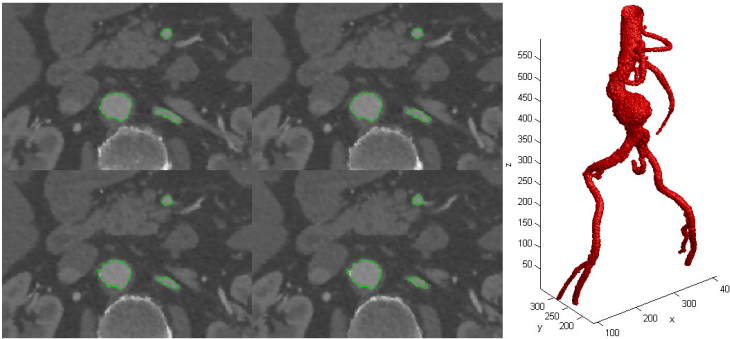
The direction of the curve normal  $\vec{N}$  is defined to point outward the curve and  $\vec{N} = -\frac{\nabla\phi}{|\nabla\phi|}$ . The curvature  $\kappa$  is given by  $\kappa = \text{div}\left(\frac{\nabla\phi}{|\nabla\phi|}\right)$ . The used iterative curve evolution algorithm, based on additive operator splitting (AOS) approach is in detail described in [12]. Some preprocessing steps and parameter set up are described in Sections 2.2 and 5, respectively.

## 2.2 Preprocessing and Initial Surface Selection

The analysed CTA series are affected by artefacts and noise, which can influence the final segmentation results. Therefore, as a preprocessing step a 3D adaptive filtering procedure was employed. The selected filtering technique based on anisotropic diffusion [18] increases the Signal-to-Noise Ratio (SNR) and preserves the edges.

In the hybrid level-set implementation applied to volumetric CT data, the authors of [12] used a sphere as an initial surface. The results presented by them show that it successfully converges to the target object. However, the performed experiments proved, that the time to converge the hybrid level-set algorithm [12] strongly depends on this surface. In our work, the size of the analysed AAA CTA data ( $512 \times 512 \times n$ , where  $n \in [220, 680]$ ) determined the clustering-based initial surface selection procedure. For this, we used a weighted fuzzy *c*-means clustering procedure introduced in [17]. The initial surface was then created by all the voxels belonging to the cluster with the highest mean gray intensity value.

Thanks to this two preprocessing techniques the hybrid level-set algorithm enables fast and robust AAA segmentation. An exemplary final 3D segmentation results of abdominal aorta lumen in CT series are shown in Fig. 3.



**Fig. 3.** An example of 3D abdominal aorta lumen segmentation results in an image view (left) and volume rendering (right)

### 3 Skeletonization

The CTA volume matching procedure being an overall goal of our work is based on graph matching step described in the next section. For this a 3D skeletonization step is incorporated. The 3D skeleton is obtained using the method described in [19]. This automatic algorithm computes subvoxel precise skeleton of volumetric data based on subvoxel precise distance field. The advantage of the subvoxel approach over a voxel precise skeleton is, that it computes an accurate, more precise and centered skeleton also for objects that are less than a single voxel thick. The authors of [19] have proven, that it is a proper solution for the accurate measurements of the object, such as vessel cross section or volume.

The input for the skeletonization method described in [19] is a subvoxel precise distance field. To obtain this field the authors suggest a two steps preprocessing technique. First, a level set time-crossing map calculation followed by a distance field computation is performed. Then, a sampled level set time-crossing map with the embedded zero-crossing isosurface is estimated. In our approach, the isosurface required for this, which yields the object's true boundary is created based on the previously obtained segmentation results. Having a implicit representation of the boundary, we estimate the subvoxel precise Euclidean distance transform for  $n$ -dimensional data [20].

The Euclidean distance field is then used to find the point with the largest distance from the boundary and to determine a speed image used as an input for the fast matching propagation step. The speed image being a function of the distance field ( $d$  - distance value and  $D$  - maximum distance value)

$$v = \left( \frac{d}{D} \right)^2 \quad (5)$$

is used to determine the curve evolution velocity for each pixel. The point at the global maximum distance from the objects boundary is calculated in a single pass through the distance field. The first one encountered in scanline order is used in the situation, if no unique maximum point exists. It is a start point for a fast marching propagation algorithm, in which the obtained speed image is used. The fast marching propagation is augmented to calculate the geodesic distance (Manhattan distance) inside the object starting at the global maximum point of the distance field. Based on the obtained results the branch points are then estimated. The furthest point of the model from the global maximum distance point is used as the start point of the branch. The remaining points of the branch are determined by performing a gradient descent, back-tracking procedure on the fast marching time-crossing map. This process is repeated for each branch of the created skeleton [20]. The exemplary results of application of this method to AAA data are shown in Fig. 4.

### 4 Graph Matching

The previously obtained aorta skeletons are now matched to properly register the two analysed 3D CTA series for each examination. The registration procedure