Simultaneous evaluation of volumes and synergistic motion of the left ventricle from X-ray angiograms using a unified computational framework

Evaluación simultánea de volúmenes y movimiento sinérgico del ventrículo izquierdo a partir de angiogramas de rayos X utilizando un marco computacional unificado

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Abstract

X-rays angiograms of the left ventricle as the main cavity of the human heart are acquired at the catheterization rooms routinely in order to evaluate the cardiac dynamic function. The global measurement of volumes, ejection fraction and synergistic motion associated with the ventricular cavity are considered in the assessment of such function. A precise left ventricle silhouette on the angiograms is necessary to calculate the function descriptors. A segmentation method is required to obtain the projected cavity shape

Keywords: X-rays angiograms, human heart, left ventricle, ventricle volumes, synergistic motion, cardiac function.

Resumen

Los angiogramas de rayos X del ventrículo izquierdo como la cavidad principal del corazón humano se adquieren en las salas de cateterismo de forma rutinaria para evaluar la función dinámica cardíaca. La medición global de los volúmenes, la fracción de eyección y el movimiento sinérgico asociados con la cavidad ventricular se consideran en la evaluación de dicha función. Es necesaria una silueta precisa del ventrículo izquierdo en los angiogramas para calcular los descriptores de funciones. Se requiere un método de segmentación para obtener la forma de la cavidad proyectada.

Palabras clave: Angiogramas de rayos X, corazón humano, ventrículo izquierdo, volúmenes ventriculares, movimiento sinérgico, función cardíaca.

Introduction

The assessment of the left ventricular function has been performed for many years through the analysis of the ventriculography as x-ray angiography technique[1]. This quantitative analysis is based on evaluating the functional status considering different assumptions closely related to the left ventricle silhouette on the X-ray projections, time course of contraction, and prefixed reference points[2]. In the catheterization rooms before acquiring the cardiac images, a contrast medium is injected into the left ventricle in order to improve the information associated with X-ray projections of the cavity. This injection improves the LV contrast compared to other tissues.

In general, the application of segmentation methods to cardiac images is a necessary step for automatically reproducing the morphology of the left ventricle (LV) required for the analysis and diagnosis of cardiovascular dynamical function[3]. In this sense, several segmentation methods have been developed, however, as a research background of this work, some of the most recent methods that consider the level set as core of identification procedure will be presented below.

The left ventricle has been segmented from short-axis magnetic resonance images (MRI) using a two-layer level set. The segmentation approach considers a novelty term that allows fitting the contour to the endocardium through a circular shape constraint4. The set of data from short-axis MRI used in a cardiac segmentation challenge held by a MICCAI workshop in 2009 is considered for a quantitative validation which demonstrates the accuracy and robustness of the method.

Another reported technique allows detecting the endocardial and epicardial contours of the left ventricle from computerized tomography (CT) images using a two-step approach[5]. First, the salient information relating to cavity and neighboring cardiac tissues is determined by means the Hermite transform. Then, the active shape models and level sets, guided by the Hermite transform information about cardiac structures perform the final segmentation. The authors proposed the Dice coefficient, Hausdorff distance and ray feature error for evaluating their segmentation method. These metrics are calculated from LV cavities manually reproduced by cardiologists and from the left ventricle structures estimated with the proposed approach obtained from 28 CT volumes. From these results it is verified that the technique is accurate and efficient to identify the left ventricular cavity.

In this paper, a unified computational framework that combines a clustering technique and a level set approach is proposed in order to obtain the relevant information for evaluating of volumes and synergistic motion associated to left ventricle heart cavity from monoplane X-ray cineangiography images. The unified computational framework considers three of the fundamentals classes of algorithms for processing and analyzing of medical images, namely, image enhancement, segmentation, and image feature representation. From the image features obtained, a set of measures of performance are calculated in order to assess the ability of the method to automatically reproduce the left ventricle projected shape. Finally, the volumes.

Methods

The proposed computational solution requires as minimum specifications for its execution: an Intel or AMD x86 processor, 4 GB of memory RAM, 2.2 GB of HDD space for MATLAB only, 4–6 GB for a typical installation.

Image source

In order to test the performance of evaluation framework, 44 sequences of mono-plane ventriculographic images acquired with a General Electric Medical System (InnovaTM 4150) are considered. The acquisition protocol considers the right anterior oblique incidence (RAO 30º), a spatial resolution of 512 ´ 512 pixels, a pixel depth of 8 bits and a pixel size of 0.285 ´ 0.285 mm.

Stage 1: Image enhancement

The filters pipeline

The boundaries between the nearby structures of the left ventricle are nearly indistinguishable basically by the low contrast to noise ratio and because the blood and the contrast agent of the X-ray angiography images not homogeneously mixed. In this regard, an image enhancement stage is considered for improving the generally low contrast of these images. A pipeline that sequentially organizes the following set of filters is proposed as enhancement stage.

- Median filter: is an image nonlinear filter class[6]. The filter does not consider a convolution with a denoising kernel for noise attenuation. The values in a neighborhood (m×m) of an original image are sorted in an array, after, the median of this array is determined, and then this value is replaced in the current voxel[7]. The filtered image results from iteratively applying the procedure described above
- Gradient anisotropic diffusion filter: is formulated to attenuate unwanted information while preserving the specific characteristics of the images[8]. The method transforms the original image as a function of its derivatives in a higher dimensional space, this transformation represented the solution of the heat equation[9]. The conductance is obtained at each pixel according to a function to the gradient magnitude of the cardiac image.
- Erode filter: corresponds with a maximum filter and it is the morphological operation of erosion. This operator causes objects to shrink, it smoothes object boundaries and removes peninsulas, fingers, and small objects. The non–linear maximum operator requires also a structuring element[10].

In general, the major components of image noise are attenuated using the median filtering, while the remaining noise is attenuated by means the anisotropic diffusion filter which does not remove salient features of the images, meanwhile, the erode filter is applied for shrinking objects, smoothing the object boundaries and removing small objects.

Stage 2: Segmentation

First, the angiographic image is classified in two regions, namely, left ventricle and background. In this sense, a clustering algorithm whose core is a region growing technique is used. This grouping scheme considers the intensity of the current pixel and the average of the intensity of certain pixels at a neighborhood of current pixel as clustering criteria[11]. Then, the results of the region growing segmentation are improved by using a level set method[12].

Region growing technique

The dynamic linked–lists is used for computationally implementing the preliminary segmentation procedure. The clustering algorithm considers as inputs, two images of the same size, namely, the filtered image and an utilitarian image with all pixels set to zero (0). The list is organized using the First In First Out (FIFO) method and its utility is to store temporarily the location and the intensity of the filtered image pixels that fulfill the clustering criteria. The dynamically linked–lists implemented by the FIFO method result in a computational algorithm highly efficient with respect to memory overflows, especially in system with minimum memory requirements. The region growing technique requires a seed pixel located inside the left ventricle to perform the grouping. The information on this pixel, location and intensity, constitutes the first node allocated in the list.

The dynamically linked–lists based algorithm considers the following steps: 1) the information stores in the first node of the list is extracted, 2) the image intensity associated to current node is compared respect to the average intensity in an 8 pixel neighborhood. The information of those pixels in the neighborhood that fulfill the clustering criteria is queued at the end of the list, and the intensity of the utilitarian image is changed to 1 in the location of queued pixels. A pixel that does not fulfill the criteria is not inserted in the list, and 3) If there are nodes in the list the algorithm running. The algorithm output corresponds with the utilitarian image which is a binary image that contains the two classified regions. The MatLab script that corresponds with region growing technique is available on MathWorks[13].

Level set method

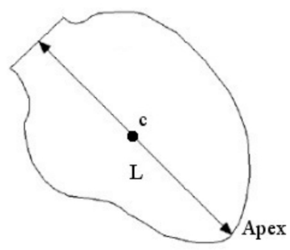
The level sets method allows the iterative deformation of certain geometric structures called snakes[14]. The method is based on finding a specific contour defined by the intersection between a surface and a plane. The surface is the three-dimensional representation of the image, which the image plane is the x-y plane, and the image intensity is the z coordinate. This formulation internally preserves the regularity of the level sets function of snake during the evolution. The expected location of contour is reached through the evolution of the zero level set which is driven by a global energy functional composed of a distance regularization term and an external energy term. A gradient flow approach is used for optimizing this energy functional. The Matlab code that implements the edge based geometric active contour model without reinitialization is available on MathWorks[15].

Stage 3: Image features representation

The binary image obtained in the segmentation stage is a matrix that contains information of the left ventricle and the background. A more compact representation of the binary image is required in order to depict the pixels that delimit the cardiac cavity. This representation is useful for performing analyzes or geometrically operations, for performing the extraction of the parameters that define the left ventricle, or for applying modification operations of the form thereof[16]. Data structures are considered in this research for storing and depicting the points that delimit the object as the coding technique of left ventricle contours.

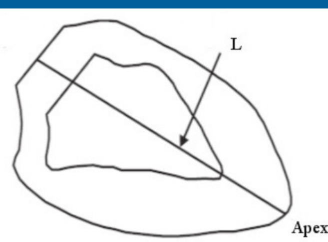
After cavity representation, the fundamental idea is to compute certain parameters that can give an adequate description of the left ventricle shape and from which the ventricular volumes and synergistic motion can be estimated. In this sense, the centroid of the left ventricle closed contour (c), the area of the cavity (A), the location of the apical point (Apex), and the anatomical axis (L) are required.

Figure 1. Left ventricle contour representation.



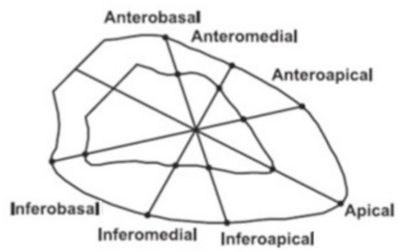
The mobility center required to analyze the synergistic motion is defined according to the following procedure: the LV contour at end-systole must be aligned with respect to the centroid of the end-diastole contour, and subsequently the anatomical axis (defined between the aortic valve and the apex) of both contours should be aligned, as shown in Figure 2.

Figure 2. Ventricular contours aligned.



The synergistic motion is evaluated using a technique that allows dividing the ventricle into a set of segments (see Figure 3), to then analyze the ventricular mobility of systole with respect to diastole in each one of them. The method is based on the measurement of the distances between the end-diastolic and final systole contours after being aligned with respect to a center of cardiac mobility[17].

Figure 3. Segments division of the LV applied to X-ray ventriculography.



The unified computational framework

The computational framework proposed considers the following steps:

Step 1: Identification of the diastole isovolumetric instant

Step 2: Application of the filter pipeline to the image of the isovolumetric instant in diastole

Step 3: Preliminary segmentation using region growing technique

Step 4: Optimization of preliminary segmentation using level set

Step 5: Representation of the image features

The median and erosion filters are implemented using Image Processing Toolbox of MatLab, meanwhile the gradient anisotropic diffusion filter is available on MathWorks[18].

Experimental setup

The value of parameters associated with the complete computational framework is established according to the heuristic evaluation method[19] applied to one image at end–diastole of the cardiac cycle. Table 1 shows the values of the parameters considered by procedures involved the proposed computational framework. From the Table 1, it is important to note that the absolute threshold level in the region growing MatLab implementation has three values, for images with low contrast, medium contrast and high contrast, respectively.

Table 1. Parameter values used in the proposed computational framework.

Procedure	Parameter	Value
Median filter	*m*–Neighborhood size	17
Anisotropic diffusion filter	Number of iteration	25
	Integration constant	1/7
	Gradient modulus threshold	5
	Conduction coefficient functions	2
Erode filter	Structure element size	5´5
Region growing	Absolute threshold level	[4, 7.5, 11.5]
	Maximum distance	180
Level set	Time step	5
	Weight of distance regularization term	0.04
	Number of iterations	23
	Weight of the weighted length term	5
	Weight of the weighted area term	-0.3
	Width of Dirac Delta function	1.95
	Scale parameter in Gaussian kernel	1.5
	Size of Gaussian kernel	15

Evaluation of ventricular volumes

The projected shape of the left ventricle obtained using the proposed framework and the left ventricle silhouette manually reproduced by two cardiologists (V.B. and AD-M) are compared in order to evaluate the accuracy with which the method automatically reproduces the projected shape of the cavity. In this sense, the idea is to calculate certain metrics that allow performing such comparison. The strategy proposed by Suzuki et al.[20] which considers two metrics that represent the contour error (E_C) and the area error (E_A) is initially used in the evaluation.

Additionally, the evaluation methodology proposed by Chalana and Kim[21] is also considered in order to reduce the variability introduced by the experts appointed to manually reproduce the projected shape of the ventricular cavity. This reduction is performed calculating an average contour from the contours reproduced by two physicians (Expert 1 and Expert 2), which is considered a mean ground truth contour. From this methodology, the incorporation of the mean absolute distance (MAD) as a comparison metric interpreted as the position error between the compared contours is also considered.

Furthermore, the Dice coefficient is used for comparison. The degree of overlap between two areas enclosed is measured using this coefficient, and corresponds to the ratio of twice the areas of intersection to the sum of the two areas[22].

Tables 2–4 show the values obtained when the contours automatically reproduced (M) are compared with the contours manually reproduced by Expert 1 (E_1) and Expert 2 (E_2), and the mean ground truth contour (E_a). These values are expressed using the mean, the maximum (max), the minimum (min) and the standard deviation (Std). Table 2 shows the comparison values obtained when all images included in all angiographics explorations are analyzed. Table 3 shows the values obtained only when the final diastole images of all the explorations are analyzed. And, Table 4 shows the values obtained for end-systole images.

Table 2. Contour and area errors obtained for all images processed.

	E_1 vs M [%]		E_2 vs M [%]		E_a vs M [%]	
	E_C	E_A	E_C	E_A	E_C	E_A
Min	2.32	0.75	2.25	0.05	3.21	0.18
Mean	5.88	2.47	6.53	3.08	6.27	2.57
Max	14.38	8.53	18.91	16.81	15.01	14.28
Std	2.31	0.58	3.03	0.71	2.64	0.82

The value of contour error obtained (mean ± standard deviation) when comparing the computational extracted contours and the contours reproduced by cardiologist 1 for all angiographies explorations is about 5.88% ± 2.31%. Meanwhile, the value of this metric when comparing the contour reproduced by the cardiologist 2 is 6.53% ± 3.03%. The value of area error obtained is

2.47% ± 0.58% with respect to Expert 1, and 3.08% ± 0.71% with respect to Expert 2. The values of contour and area errors with respect to the mean ground truth contour are 6.27% ± 2.64% and 2.57% ± 0.82% for all images in the sequences, respectively.

Table 3. Contour and area errors obtained for the end-diastole images.

	E_1 vs M [%]		E_2 vs M [%]		E_a vs M [%]	
	E_c	E_A	E_c	E_A	E_c	E_A
Min	3.76	3.35	2.96	1.89	3.24	1.49
Mean	4.98	7.18	5.57	3.87	5.36	3.68
Max	7.04	15.90	8.77	7.18	8.11	7.83
Std	1.43	4.12	2.14	2.37	1.87	1.93

Table 4. Contour and area errors obtained for the end-systole images.

	E_1 vs M [%]		E_2 vs M [%]		E_a vs M [%]	
	E_c	E_A	E_c	E_A	E_c	E_A
Min	5.71	4.13	6.52	3.97	6.62	3.49
Mean	9.62	9.29	9.83	9.31	9.48	7.23
Max	16.31	16.26	17.71	16.32	17.25	15.28
Std	3.97	5.34	3.77	3.96	3.73	3.62

It is of interest for the present research in determining the values of the comparison measures specifically at the end-diastole and end-systole instants for the 44 patients because the image features representation at these instants are used to assess the synergistic motion of the cavity. The contour error at end-diastole and end-systole is 4.98% ± 1.43% and 9.93% ± 3.97%, respectively, with respect to an Expert 1, and, the area error is 7.18% ± 4.12% and 9.29% ± 5.34%. Meanwhile, with respect to an Expert 2, the contour error is 5.57% ± 2.14% and 10.62% ± 3.77% at end-diastole and end-systole, respectively, and the area error is 3.87% ± 2.37% and 9.31% ± 3.96%. The values of contour and area errors with respect to the mean ground truth contour are 5.36% ± 1.87% and 3.68% ± 1.93% at end-diastole instant; meantime at end-systole instant, the errors are 10.48% ± 3.73% and 7.23% ± 3.62%.

Table 5 shows the values of the mean absolute distance computed between the contours achieved using the computational framework and the contours manually reproduced by an Expert 1, an Expert 2 and the mean ground truth contour.

Otherwise, the Dice coefficient calculated using the areas enclosed by the computational contour and by the average contour computed from the contours reproduced by an Expert 1 and an Expert 2, is 99.04% ± 0.70% for the 44 datasets of the images. The maximum value of this coefficient is 99.79% and the minimum value achieved is 97.20%. Therefore, the minimum value of the error reached is 0.21% and the maximum value of segmentation error is 2.80%.

Table 5. Mean absolute distance obtained for all images processed.

	MAD [mm]		
	E_1 vs M	E_2 vs M	E_a vs M
All images	0.68 ± 0.37	0.71 ± 0.41	0.70 ± 0.32
End-diastole images	0.61 ± 0.30	0.72 ± 0.31	0.52 ± 0.30
End-systole images	0.76 ± 0.39	0.86 ± 0.41	0.66 ± 0.34

In this work, the area-length method[23] is considered for calculating the ventricular volumes. This volume estimation method requires of the contour that represents the projected left ventricle shape. Once the volumes at end-diastole and end-systole are calculated it is possible to estimate the ejection fraction. From there, another way to evaluate the performance of the developed computational framework consists in computing the volumes at end-diastole and end-systole, and the ejection fraction using the contours reproduced by an Expert 1, an Expert 2, and the average contour, and then, correlates such clinical parameters with those calculated from computational contours. The correlation coefficients obtained by comparing the clinical parameters previously valued are shown in Table 6.

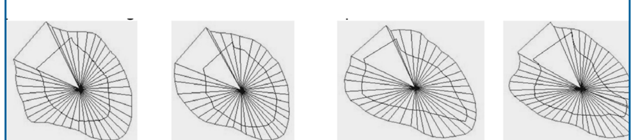
Table 6. Squared correlation coefficient for clinical parameters.

	ED Volume	ES Volume	EF
E_1 vs M	0.9841	0.9889	0.8452
E_2 vs M	0.9961	0.9892	0.8799
E_a vs M	0.9992	0.9899	0.8945

Evaluation of synergistic motion

Figure 4.c shows the synergistic motion analysis for two patients. Figure 4.a and 4.b show the manual contours and those obtained by the framework for a specific patient, while Figure 4.c and 4.d for another patient.

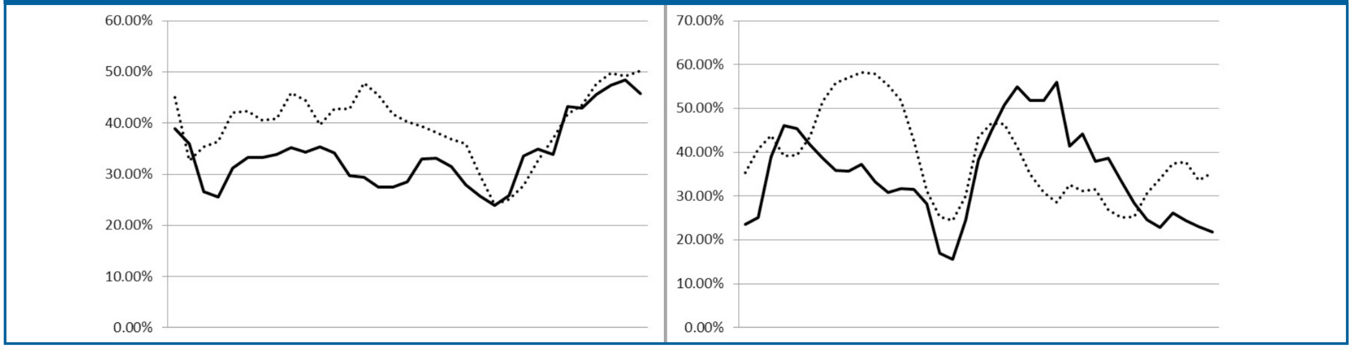
Figure 4. Left ventricles contours for evaluation of synergistic motion.



(a) (b) (c) (d)

Figure 5 shows the percentage curve of shortening that describes quantitatively the synergistic motion obtained for the contours of the two patients. The initial and final points of the graphs in Figure 5 correspond to the base plane of the heart (mitral valve), meanwhile the apex corresponds to the midpoints of the graphs. In each graph, the solid line represents the shortening calculated by analyzing the mean ground truth contours at diastole and systole, while the dotted line represents the shortening obtained from evaluating the contours generated by the proposed computational framework.

Figure 5. Graphics of synergistic motion.



Discussion

In order to evaluate the proposed unified computational framework two evaluation schemes are considered. First, the contours computationally reproduced are compared with the contours manually reproduced by two cardiologists. Secondly, from the computational and manually contours of the cavity, the ventricular volumes and ejection fraction are calculated and then correlated. The first evaluation scheme is supported by the strategies proposed by Suzuki et al.[20] and Chalana and Kim[21] which are used by Oost et al.[24].

The maximum value of the contour error at end-diastole and end-systole reported by Suzuki et al.[20] is 6.2% and 17.1%, respectively, however, this error calculated using the proposed computational framework is the 5.57% and 9.83%. The error achieved in this work is smaller in both cardiac instants. For the images at end-systole, the contour error is also smaller than the value reported by Vera et al.[25] (9.87%) for this cardiac instant.

The mean average distance or the position error obtained by the proposed computational framework when analyzing the contours at end-diastole and end-systole is smaller compared to those reported by Oost et al.[24]. The higher MAD at end-diastole and end-systole is obtained when comparing the computational contours with respect to the contours reproduced by an Expert 2, and the values are 0.72 ± 0.31 mm and 0.86 ± 0.41 mm, respectively. The position error calculated from the computational contours at end-diastole and end-systole varies between 0.22 and 1.03 mm and between 0.32 and 1.27 mm, respectively, meanwhile the error values reported by Oost et al.24 are in the intervals [0.31, 1.05] mm and [0.69, 2.21] mm, respectively.

The end-diastole and end-systole volumes, as well as the ejection fraction estimated from the computational left ventricle contours strongly correlate with the values estimated from the contours reproduced by cardiologists. The clinical parameters derived from a computational contour agree very well with those calculated from an average contour or mean ground truth contour (Table 5). The best correlation values for volumes (end-diastole and end-systole) and ejection fraction are 0.9992, 0.9899 and 0.8945, respectively, and they are obtained when the clinical parameters calculated from the computational contour are correlated with the clinical parameters estimated from the contours reproduced by an Expert 2.

Finally, it is observed that the analysis of the synergistic motion corresponds to the cardiac dynamics. In this sense, it is verified that during systole in adults, the plane of the mitral valve descends some centimeters towards the apex (the greater percentage of shortening), but the apex barely moves towards the base of the heart, that is, the shortening percentage is greater for the points associated with the base of the heart, in the graphs of Figure 5, than in the points that represent the apex. Also, the contraction in the anterior regions (basal, medial and apical) dominates and it is followed by the inferobasal and inferomedial shortening. This fact has justified their analysis for a clinical interpretation of the cardiac motion. When the cardiac muscle is normal the other three movements associated to cardiac dynamic, translation, rotation and three-dimensional twisting, are of less importance.

Conclusions

A unified computational framework to evaluate of volumes and synergistic motion of the left ventricle from x-ray angiograms has been developed. The proposed computational framework is used to assess the cardiac dynamic function from cineangiography sequences of real left ventricles. Three of the fundamentals classes of algorithms for processing and analyzing of medical images are combined to develop the proposed computational framework. An image filter pipeline is considered for enhancing the left ventricle information and removing/attenuating non-relevant information. A segmentation algorithm that initially discriminates the cardiac cavity considering a clustering technique, and secondly the approximation of the cavity is optimized using a method evolves the approximate contour, is used. The optimized contour is validated and then used to calculate certain clinical parameters that explain cardiac dynamics.

The values of the clinical parameters determined from the X-ray images of left ventricle are reliable, which is demonstrated since they are concise with the values compared in this paper and with those reported in the literature[2,26]. Added to this, in the evaluation of the computational framework performance a high correlation of the clinical parameters is achieved derived from the computational contours and manually reproduced contours. The unified computational framework also allows the monitoring, visualization and analysis of the synergistic motion and therefore the evaluation of cardiovascular function in a dynamic way.

Data availability

The X-ray angiograms dataset is available from: http://doi.org/10.5281/zenodo.2590511

Software availability

Region growing software available from: https://www.mathworks.com/matlabcentral/fileexchange/32532-region-growing-2d-3d-grayscale[13] .

Level set software available from: https://www.mathworks.com/matlabcentral/fileexchange/12711-level-set-for-image-segmentation[15] .

Anisotropic diffusion filter software available from: https://www.mathworks.com/matlabcentral/fileexchange/14995-anisotropic-diffusion-perona-malik[18].

Competing interests

No competing interests were disclosed.

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