KIDNEY SEGMENTATION IN ULTRASOUND IMAGES VIA ACTIVE CONTOURS

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Abstract

In this paper is presented an effective approach for kidney segmentation in 2D ultrasound (US) images.

In regard to different limitations of US images such as poor signal-to-noise ratio, signal drop-out, missing or misplaced boundaries of the objects, a preprocessing stage is proposed. It consists of contrast enhancement based on CLAHE and speckle noise reduction with modified homomorphic filter based on wavelet packet decomposition of the transformed US image. The method of segmentation is based on active contour without edges. It can detect objects whose boundaries are not necessarily defined by gradient.

Some experimental results are presented, obtained by computer simulation in the MATLAB environment. Implementation results are given to demonstrate the effectiveness of the proposed approach in clinical diagnostics.

1. INTRODUCTION

Kidney segmentation in US images is a particularly difficult task. The US images have a poor signal-to-noise ratio, signal drop-out, missing boundaries, misplaced boundaries and reconstruction errors [1]. The kidney is an elastic organ and can suffer large deformations. Furthermore, in US the interior of the organ exhibits heterogeneous structures with different intensities, and many of the boundaries are lost due to the density similarity to surrounding structures [2].

US segmentation methods have been previously classified according to the prior knowledge employed to improve the accuracy of results. These constraints include image-derived priors (intensity, intensity derivatives, local phase, texture), and application-derived priors (shape and motion) [1]. Some of the classified the methods in different categories based on the shape, region, if the segmentation is automatic or manual and finally combing some of the categories together [3]. In Fig. 1 is presented the classification of methods for kidney segmentation according to the above mentioned categories [4].

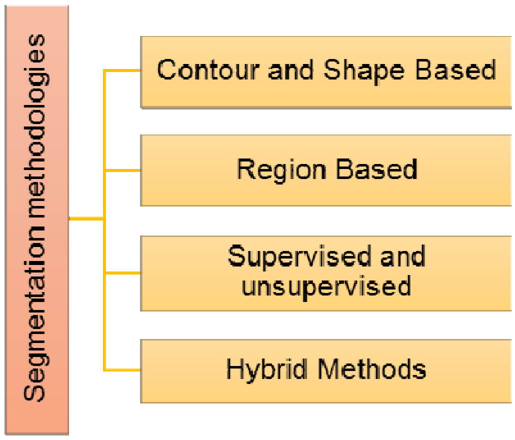

Figure 1. Classification of methods for kidney segmentation

The manually-selected elliptical approximations of the kidney region are usually provided in clinical practice [5]. However, the ellipse method is known to underestimate the kidney size up to a 25% error [6]. One of the best known and validated frameworks able to incorporate edges and shape priors is the active contour models []. The main idea behind this model is that the presence of an image feature depends not only on the value of the image at a given point in space or on its derivatives, but also on the spatial distribution of such features.

In this paper we propose an effective approach for kidney segmentation, which includes the preprocessing stage for enhancement of the US image, following by active contour segmentation based on active contour model of Chan and Vese [7]. Our goal was to implement this model and to investigate its advantage by application to US images in regard to their specific.

The paper is arranged as follows: In Section 2 is presented the main algorithm of processing; In Section 3 is presented the algorithm of kidney segmentation via active contour; in Section 4 are given some experimental results, obtained by computer simulation and their interpretation; in Section 5 - the Conclusion.

2. MAIN ALGORITHM OF PROCESSING

The flowchart of the main algorithm for US image processing is given in Fig. 2.

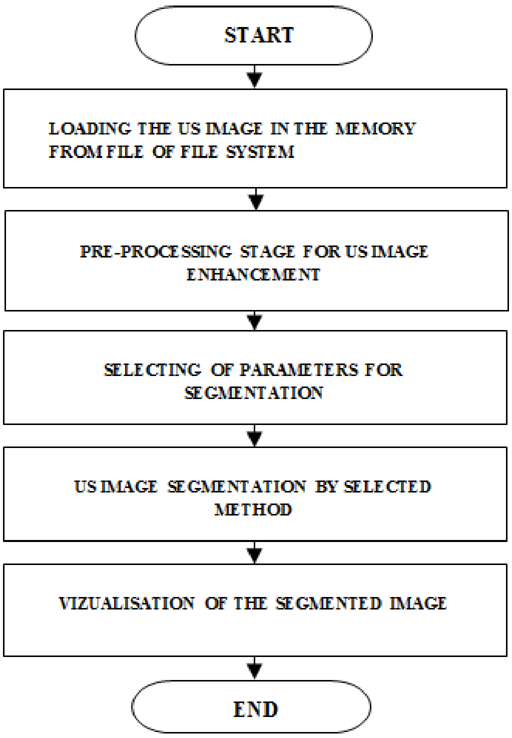

Figure 2. Flowchart of the maim algorithm

The pre-processing stage includes contrast enhancement based on CLAHE and speckle noise reduction. For modelling an image with speckle noise is used the generalized Gaussian distribution and generalized gamma distribution. For noise reduction is used modified homomorphic filter based on wavelet packet decomposition and adaptive threshold of the transformed US image [8].

The kidney segmentation is made by the implementation of the Chan and Vese active contour model. It is a special case of the Mumford–Shah problem [7]. The regularized Heaviside function H defines 2 different regions based on the level set function. The energy can be presented as a difference of the intensity function I from expected value c_1 and c_2, respectively. The Dirac function δ, which is the gradient of the Heaviside function, penalizes long boundaries between the regions. The minimization of the functional over region Ω optimizes this special case of the Mumford–Shah functional and is defined in (1):

$$E(\phi)=\int_{\Omega} a_1 H(\phi(x))[I(x)-c_1]+a_2[1-H(\phi(x))]\times [I(x)-c_2]+a_3\delta(\phi(x))dx \quad (1)$$

The weights a_1, a_2 and a_3 depend on the importance and/or reliability of the three constraints. If the functional is convex (in case if the two intensities c_1 and c_2 are fixed and known), optimization can be done based on histogram of data values $I(x)$. In the other case, the segmentation will be dependent on the initial level set function. We have used that the initial 0-level set can be defined by the doctor.

The parameters which can be selected for segmentation are following:

- *mask* — Initial contour at which the evolution of the segmentation begins;
- *n — Maximum number of iterations* to perform in evolution of the segmentation;
- *Rad –Radius of the location in pixels;*
- *Alpha* – 'Smooth Factor' – Degree of smoothness or regularity of the boundaries of the segmented regions.

After choosing all input information the procedure of segmentation begins. Then the final result from segmentation is visualized – US image with segmented object.

3. ALGORITHM OF SEGMENTATION VIA ACTIVE CONTOUR

The flowchart of the presented algorithm is presented in Fig. 3.

To get faster and more accurate results we propose an initial contour position, which is close to the desired object boundaries to be selected interactively by the doctor.

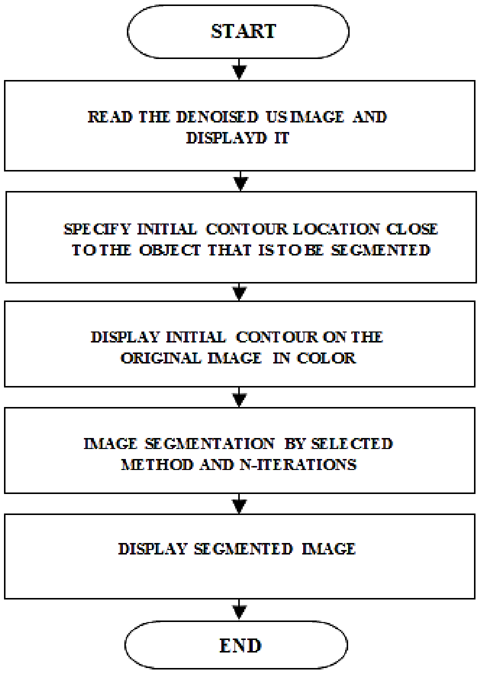

Figure 3. Flowchart of the algorithm for segmentation via active contour

If the initial contour position (specified by the region boundaries in mask) is far from the desired object boundaries, we propose to specify higher values of *n* in regard to achieve desired segmentation results. Higher values of *Alpha* produce smoother region boundaries but can also smooth out finer details. Lower values produce more irregularities (less smoothing) in the region boundaries but allow finer details to be captured.

4. EXPERIMENTAL RESULTS

The formulated stages of processing are realized by computer simulation in MATLAB 7.14 environment by using IMAGE PROCESSING and WAVELET TOOLBOXES [9]. In analysis are used 20 US images from kidney with size 640x480 pixels in jpg file format. For processing they are converted in bmp format.

Some results from simulation, which illustrate the working of proposed algorithm, are presented in the next figures below.

In Fig. 4 is shown the original US abdominal image of right kidney.

Fig. 5 illustrates the result from segmentation of the kidney. The segmentation is performing by *n=1000* iteration.

Figure 4. Original US image of right kidney

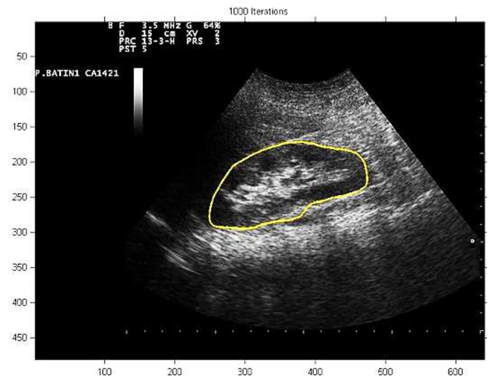

Figure 5. US image with segmented right kidney

For validation of the segmentation results, we compute the undirected partial Hausdorff distance [10] between the boundary of the computed segmentation and the boundary of the manually-segmented ground truth. The directed partial Hausdorff distance over two point sets *A* and *B* is defined in (2):

$$h_K = \underset{a \in A, b \in B}{K^{th}} \; min \| a - b \| \qquad (2)$$

where *K* is a quantile of the maximum distance. The undirected partial Hausdorff distance is defined in (3):

$$H_K(A,B) = max(h_K(A,B),\, h_K(B,A)) \qquad (3)$$

The obtained averaging results for the partial Hausdorff distance between automatic segmentation and the manually-segmented ground truth are given in Table 1.

Table1. Partial Hausdorff Distance

Method	K [%]
Manually-Segmentation	91
Active Contour Segmentation	95

The results shown in Table 1 indicate that virtually all the boundary points lie within some pixels of the manual segmentation, but the segmentation is better in the case of the proposed approach.

5. CONCLUSION

In this paper is presented a new and effective approach for kidney segmentation in US images. It consists of contrast enhancement based on CLAHE and speckle noise reduction with modified homomorphic filter based on wavelet packet decomposition. The method of segmentation is based on active contour without edges. The implemented study and obtained results have shown a high validation of the segmentation. We obtain that this method is very robust to initialization and gives better results, when there is a difference between the foreground and background. Our future work will be concentrated in hybrid methods for segmentation of other organs in US images in regard to their specific.

6. ACKNOWLEDGMENTS

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