SEGMENTATION OF RENAL CALCULI FROM ULTRASOUND IMAGES

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ABSTRACT

In the past four decades, computerized image segmentation has played an increasingly important role in medical imaging. Segmented images are now used routinely in a multitude of different applications, such as the quantification of tissue volumes, diagnosis, localization of pathology, study of anatomical structure, treatment planning, partial volume correction of functional imaging data, and computer-integrated surgery.

Image segmentation remains a difficult task, however, due to both the tremendous variability of object shapes and the variation in image quality. In particular, medical images are often corrupted by noise and sampling artifacts, which can cause considerable difficulties when applying classical segmentation techniques such as edge detection and thresholding. As a result, these techniques either fail completely or require some kind of post processing step to remove invalid object boundaries in the segmentation results.

To address these difficulties, deformable models have been extensively studied and widely used in medical image segmentation, with promising results for detecting renal calculi.

Keywords:Kidney Ultrasound – Edge detection – Improved Gabor filter – Segmentation - Deformable models.

1. INTRODUCTION

Ultrasound (US) imaging is non-ionizing, fast, portable, inexpensive and capable of real time imaging, but, unfortunately, US images typically contain significant speckle and other artifacts which complicate image interpretation and automatic processing [3]. If anatomical structures of interest could be visualized and localized with sufficient accuracy and clarity, US may in fact become a strong practical alternative imaging modality for selected applications in Kidney disease diagnosis, particularly for computer-assisted applications where the image can be processed to provide quantitative information on the kidney structures.

The segmentation technique basically divides the spatial domain, on which the images is defined, into 'meaningful" parts or regions. Trained radiologists normally do image segmentation manually. But the manual analysis of the image is a time consuming process and is susceptible to human errors. So there is a need for computer-assisted approaches to analyze the medical images. The recent developments in image processing research and the exponential growth of computational power has made computer-assisted medical image analysis viable.[8]

Deformable models are curves or surfaces defined within an image domain that can move under the influence of internal forces, which are defined within the curve or surface itself, and external forces, which are computed from the image data. The internal forces are designed to keep the model smooth during deformation. The external forces are defined to move the model toward an object boundary or other desired features within an image. By constraining extracted boundaries to be smooth and incorporating other prior information about the object shape, deformable models offer robustness to both image noise and boundary gaps and allow integrating boundary elements into a coherent and consistent mathematical description. Such a boundary description can then be readily used by subsequent applications. Moreover, since deformable models are implemented on the continuum, the resulting boundary representation can achieve subpixel accuracy, a highly desirable property for medical imaging applications.

There are basically two types of deformable models

a) Parametric deformable models and
b) Geometric deformable models.

Parametric deformable models represent curves and surfaces explicitly in their parametric forms during deformation. This representation allows direct interaction with the model and can lead to a compact representation for fast real-time implementation. Adaptation of the model topology, however, such as splitting or merging parts during the deformation, can be difficult using parametric models.

2. RELATED WORKS

Manual identification of kidney surfaces in 2D US is time consuming and operator dependent, and thus limits clinical practicality. While some studies have shown some promise in automatically identifying the kidney surface based on intensity and gradient information (or a combination of both) these techniques were limited to 2D US, and remained sensitive to typical image variability and choice of processing parameters [5].

Daanen *et al* [6] proposed a method where prior knowledge of object structure appearance was incorporated. However, diseased surfaces do not have a continuous smooth surface and prior knowledge of fragment shape is mostly unavailable. Other approaches combined intensity and gradient-based techniques with multimodal registration of US to preoperative CT [7]. However, preoperative CT requires additional time and expense and is not always considered necessary for diagnosis or treatment, so it is only available in selected cases.

3. SYSTEM MODEL

In US images, kidney surfaces typically appear blurry with non-uniform intensity and substantial shadowing beneath the surface. The thickness of the response at the leading edge ranges from 2-4 mm for a typical transducer. The actual kidney surface lies between the highest gradient and the highest intensity points of this thick response. The proposed system uses edge detector, to identify the texture features of the kidney surface and useful in segmenting the inner layers for disease diagnosis.

The purpose of edge detection is to capture the major axis of symmetry of a feature at some specified spatial scale. Signals that have even symmetry about the origin will have real (and even) Fourier transforms, while signals that have odd symmetry will have imaginary (and odd) Fourier transforms. Signals that are neither perfectly odd nor perfectly even will have complex Fourier transforms (i.e. have both real and imaginary parts) where the resultant phase values reflects their degree of symmetry.

4. EXPERIMENTAL EVALUATION

The experimentation of the speckle noise removal from ultrasound kidney images is conducted to quantitatively evaluate the performance of the proposed 2D local phase and edge texture extraction for future kidney segmentation.

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Parametric deformable models

There are two different types of formulations for parametric deformable models for segmentation. i.e., an energy minimizing formulation and a dynamic force formulation. Although these two formulations lead to similar results, the first formulation has the advantage that its solution satisfies a minimum principle whereas the second formulation has the flexibility of allowing the use of more general types of external forces.

Energy minimizing formulation

The basic premise of the energy minimizing formulation of deformable contours is to find a parameterized curve that minimizes the weighted sum of internal energy and potential energy. The internal energy specifies the tension or the smoothness of the contour. The potential energy is defined over the image domain and typically possesses local minima at the image intensity edges occurring at object boundaries. Minimizing the total energy yields internal forces and potential forces. Internal forces hold the curve together (elasticity forces) and keep it from bending too much (bending forces). External forces attract the curve toward the desired object boundaries. To find the object boundary, parametric curves are initialized within the image domain, and are forced to move toward the potential energy minima under the influence of both these forces.

Dynamic force formulation

The deformable model was modeled as a static problem, and an artificial variable was introduced to minimize the energy. It is sometimes more convenient, however, to formulate the deformable model directly from a dynamic problem using a force formulation. Such a formulation permits the use of more general types of external forces that are not potential forces, i.e., forces that cannot be written as the negative gradient of potential energy functions.

External forces

External forces are applicable to both deformable contours and deformable surfaces.

Multi-scale Gaussian potential force

When using the Gaussian potential force, must be selected to have a small value in order for the deformable model to follow the boundary accurately. As a result, the Gaussian potential force can only attract the model toward the boundary when it is initialized nearby. To remedy this problem, it was proposed using Gaussian potential forces at different scales to broaden its attraction range while maintaining the model's boundary localization accuracy. The basic idea is to first use a large value to create a potential energy function with a broad valley around the boundary. The coarse-scale Gaussian potential force attracts the deformable contour or surface toward the desired boundaries from a long range. When the contour or surface reaches equilibrium, the value is then reduced to allow tracking of the boundary at a finer scale. This scheme effectively extends the attraction range of the Gaussian potential force. A weakness of this approach, however, is that there is no established theorem for how to schedule changes. The ad hoc scheduling schemes that are available may therefore lead to unreliable results.

Pressure force

Cohen proposed to increase the attraction range by using a pressure force together with the Gaussian potential force. The pressure force can either inflate or deflate the model. Hence, it removes the requirement to initialize the model near the desired object boundaries. Deformable models that use pressure forces are also known as balloons

It must be carefully selected so that the pressure force is slightly smaller than the Gaussian potential force at significant edges, but large enough to pass through weak or spurious edges. When the model deforms, the pressure force keeps inflating or deflating the model until it is stopped by the Gaussian potential force. A disadvantage in using pressure forces is that they may cause the deformable model to cross itself and form loops

Distance potential force

Another approach for extending attraction range is to define the potential energy function using a distance map. The value of the distance map at each pixel is obtained by calculating the distance between the pixel and the closest boundary point, based either on Euclidean distance or Chamfer distance. By defining the potential energy function based on the distance map, one can obtain a potential force field that has a large attraction range.

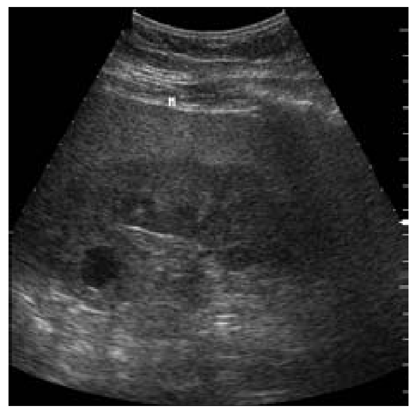
Gradient vector flow

The distance potential force is based on the principle that the model point should be attracted to the nearest edge points. This principle, however, can cause difficulties when deforming a contour or surface into boundary concavities. Notice that at the concavity, distance potential forces point horizontally in opposite directions, thus preventing the contour from converging into the boundary concavity. To address this problem, employ a vector diffusion equation that diffuses the gradient of an edge map in regions distant from the boundary, yielding a different force field called the gradient vector flow (GVF) field. The amount of diffusion adapts according to the strength of edges to avoid distorting object boundaries.

The proposed algorithm was implemented in MATLAB and run on an Intel Pentium 4 PC (3.64 GHz, 2GB of RAM). In addition to the quantitative speckle noise removal, the proposed model also present qualitative results for the texture extraction of the kidney image edges. The deformable model is used for segmentation.

Table 1: Average distance found using deformable models

Patients	Training Sample 1 Age (0 to 15)	Training Sample 2 Age (16 to 35)	Training Sample 3 Age (36 to 60)
Bad initialized images [%]	0.0	18.0	4.1
Average distance [pix]	6.07	15.06	12.18



a) Renal Calculi before segmentation

b) Renal calculi after segmentation

5. CONCLUSION

The proposed approach for accurate and fully automatic extraction of kidney surfaces directly in 2D ultrasound volumes based on local phase symmetry image features that employ improved filters. Kidney surface localization accuracy assessed using normal kidney image standards and showed a maximum mean error of 0.12 mm and a low standard deviation across the sampled points of only 0.05 mm. These errors were relatively independent of the depth of the edges of the kidney internal segment interface and of the inclination of the probe relative to the outer kidney surface.

Geometric deformable contours behave similarly to the parametric deformable contours but have the advantage of being able to change their topology automatically. The relationship between parametric deformable contours and geometric deformable contours can be formulated more precisely.

Hence, computer assisted approaches for analyzing the images of renal calculi is preferred over the manual interpretation of the images as manual interpretation is time consuming process and is susceptible to human error. The wider implications of this result are that the framework can be adapted to the other medical applications, where ultrasound images are used. Using the basic ultrasound property of shadows, many potential problems like mass and obstructions can be detected. Also, the algorithmic detection of calculi is useful for constructing expert systems, which are useful to give accurate expertise related treatment.

6. FURTHER ENDEAVORS

Numerous extensions have been proposed to the deformable models, particularly to extend the parametric deformable models. These extensions address two major areas for improving standard deformable models. The first area is the incorporation of additional prior knowledge into the models. Use of prior knowledge in a deformable model can lead to more robust and accurate results. This is especially true in applications where a particular structure that requires delineation has similar shape across a large number of subjects.

Traditional parametric and geometric deformable models are local models contours or surfaces are assumed to be locally smooth. Global properties such as orientation and size are not explicitly modeled. Modeling of global properties can provide greater robustness to initialization. Furthermore, global properties are important in object recognition and image interpretation applications because they can be characterized using only a few parameters. Note that although prior knowledge and global shape properties are distinct concepts, they are often used in conjunction with one another. Global properties tend to be much more stable than local properties. Therefore, if information about the global properties is known a priori, it can be used to greatly improve the performance of the deformable model.

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Biography

Ms.P.R.Tamilselvi received the Master of Science in Computer Science in 1996 from Bharathiar University. She completed her Master of philosophy in Computer Science in the year 2001. She has guided 25 M.Phil research scholars from various universities. She has presented and published many papers in national and international conferences and journals. Currently she is a Senior Lecturer and research scholar at Kongu Engineering College, Anna University. Her area of research interest is Medical Imaging.

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