Preprocessing and Efficient Volume Rendering of 3-D Ultrasound Image

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SUMMARY Visualization of 3-D ultrasound images is a challenging task due to the noisy and fuzzy nature of ultrasound imaging. This paper presents an efficient volume rendering technique for 3-D ultrasound image. A preprocessing technique of 2-D truncated-median filtering is proposed to reduce speckle noise of the ultrasound image. This paper also introduces an adaptive boundary detection method to reduce the computation time for volume rendering of ultrasound image. The proposed technique is compared to the conventional volume rendering methods with respect to the computation time and the subjective image quality. According to the comparison study, the proposed volume rendering method shows good performance for visualization of 3-D ultrasound image.

key words: volume rendering, 3-D ultrasound imaging, truncated-median filter, ray casting, depth cueing

1. Introduction

Two-dimensional (2-D) real-time ultrasound imaging (also called B-scan imaging) is commonly used in a variety of medical fields including obstetrics, gynecology, cardiology, fetal neurology, surgery and needle-guidance. A drawback of conventional 2-D ultrasound imaging is the limitation of 3-D visualization of anatomy. The 3-D ultrasound imaging involves the acquisition of a set of 2-D data, image reconstruction, volume rendering, and volumetric data analysis [1].

Volume rendering techniques have used opacity, color, scattering and emissivity to visualize the 3-D data. Udupa and Odhner had proposed a technique for rendering semi-transparent surfaces, which partly overcomes the high computational complexity and storage requirements of ray casting methods [2]. They defined data structure called the shell, where a surface is identified as a subset of voxels. The shell is effectively a fuzzy set of boundary points that can be projected directly onto the image plane. Splatting is another object-oriented method [3] where each voxel is splatted onto the image plane with an appropriate spread function along the view direction.

One of the major reasons for the limited acceptance of the 3-D ultrasound is the lack of an appropriate visualization technique to display clear surfaces out of the acquired data. The well-known techniques for surface extraction, which are used for magnetic resonance imaging (MRI) and computed tomography (CT), are not appropriate to ultrasound imaging because of the following features of ultrasound data [4]:

a. Significant amount of noise and speckle.
b. Low dynamic range in comparison with CT and MRI.
c. High variations in the intensity of neighboring voxels, even within homogeneous tissue areas.
d. Region boundaries are not clear in ultrasound image.

Since ultrasound image has low dynamic range, it is difficult to recognize image signal from speckle noise. Therefore, most of the ultrasound images require a preprocessing operation to improve the image quality. In order to overcome these features of ultrasound imaging for volume rendering, an efficient preprocessing technique is proposed in this paper. Also the rendering time is reduced by adaptive boundary detection for ray casting.

This paper presents an efficient preprocessing method and an adaptive boundary detection method for 3-D volume rendering of the ultrasound data. Section 2 describes the proposed techniques for visualization of 3-D ultrasound data including the preprocessing, boundary detection, and ray casting. Some simulation results are shown in Sect. 3 to verify the performance of the proposed technique. Finally, Sect. 4 gives conclusions.

2. Efficient Volume Rendering for Ultrasound Data

2.1 Characteristics of Ultrasound Image

The quality of ultrasound image is relatively poor compared with X-ray computed tomography and magnetic resonance images. The object boundaries in ultrasound images carry structural information that is vital for diagnostic purposes. However, boundary detection is a difficult processing in ultrasound imaging because of speckle, acoustic reverberations and several other limitations [5].

The sound waves can be divided into specular
reflection and diffuse scattering (volume scattering). Specular reflections appear between tissue layers with different acoustic impedances. The specular reflections are represented by bright boundaries in ultrasound image. The scattering of ultrasonic waves is caused by spatial variations of the acoustic impedance in the human body [6]. Low intensity echoes from the volume scattering are scattered over a wide angular range in the beam plane due to inhomogeneities of the medium whose size is smaller than the ultrasound wavelength. Because wide-angle back-scattered echoes are nearly independent from incidence angle, they form the basis of pulse-echo imaging. Every different tissue has a different volume scattering so that it has its own gray-level in ultrasound image. This makes it possible to detect the boundaries between tissues by computing local gradient estimates. However, the constructive and destructive interferences between neighboring scatterers give rise to a multiplicative noise that is modulated with image signal. The multiplicative noise, called speckle, generates multiple unclear edges when gradient operation is applied. It is, therefore, essential that the gradient estimates should be as insensitive to speckle as possible.

2.2 Preprocessing of Ultrasound Images

A noise reduction filtering is necessary for visualization of 3-D ultrasound images. However, it also involves a trade-off of information loss. One of the goals of this paper is to maximize the visualization quality while minimizing the loss of information from the filtering process. The most popular filter to reduce the speckle noise is the median filter [7]. The median filtering, typically with a square window in 2-D or 3-D image, achieves less blurring and better edge preservation than simple averaging and linear filtering techniques. Median filters are not separable and require large amount of computation. Sakas had introduced a median filter of \(3 \times 3 \times 3\) or \(5 \times 5 \times 5\) followed by a Gaussian filter of \(3 \times 3 \times 3\) for preprocessing of the 3-D ultrasound images [7].

The characteristics of granular or mottled appearance in ultrasound images, commonly referred to speckle, are the result of a complex superposition of many factors including interference effects, attenuation and electrical processing. The granular pattern does not correspond to the actual tissue microstructure [8]. Since speckle intensity in ultrasound images often exceeds specular echo intensity, it is difficult to apply traditional boundary extraction and segmentation algorithms efficiently. Consequently, speckle-reduction filtering is an important aspect for visualization and segmentation of ultrasound images. Speckle is modelled in a multiplicative Rayleigh noise that degrades ultrasound images by concealing fine structures and reducing the signal to noise ratio (SNR) [9]. Kotropoulos and Pitas [9] derived a maximum likelihood estimator for ultrasound signal corrupted by noise. The maximum of the Rayleigh distribution provides a maximum likelihood estimator [9]. The maximum of the probability density function (PDF) is given by derivative of the PDF with respect to \(z\) as follows,

\[
\frac{\partial p(z)}{\partial z} = 0 = \exp\left(\frac{-z^2}{2\sigma^2}\right) \left(1 - \frac{z^2}{\sigma^2}\right)
\]

where \(\sigma^2\) is the variance of the Rayleigh distribution. The PDF has a maximum value at \(z = \sigma\). Given a signal corrupted by Rayleigh noise, the mode of a set of observations then corresponds to the maximum likelihood estimator of the original signal.

The truncated-median filter was first proposed as an alternative of a mode filter by Davies [10]. The truncated-median filter was devised on the basis of modifying the median filter so that it approximates the mode filter. Three means appear in the normal order of mean-median-mode for many of the distributions as shown in Fig. 1. To approximate the mode with median, the truncated-median filtering is to truncate the original distribution so that the maximum distance from median to left-end value can be the same as that to right-end value. A new median value is computed again from the truncated signal that is closer to the mode. According to the iteration of these processes, the truncated-median output moves further towards the mode. Two iterations of the truncated-median filtering have been found to be sufficient in practical applications [11].

Truncated-median filtering technique is used as an alternative of mode filter for preprocessing of the ultrasound image. The median and truncated-median filters operate at a similar speed in case of 2-D image window. Thus the preprocessing of truncated-median filter is faster than that of 3-D median filter [11]. Therefore, we propose the truncated 2-D median filtering as a pre-filtering of ultrasound image rather than 3-D median filtering. After truncated-median filtering, 3-D Gaussian filtering is applied so as to compensate the misalignment of the successive 2-D images. Since 3-D Gaussian filter is separable, the computation complexity of this processing is not critical in comparison with the median filtering.

![Fig. 1](image-url)  
*Fig. 1* Normal order of mean, median and mode in a unimodal distribution and a method of truncation.
2.3 Adaptive Boundary Detection

Levoy had proposed two different methods for the opacity computation of 3-D volume rendering which are region boundary and isovalue-contour surface methods[12]. Levoy’s opacity formulas are based on a fixed threshold of the gray-level and its gradient. The fixed threshold can lead to surface gaps and holes because the gray level along an object boundary depends on the surface curvature, distance, and orientation to the sound source.

In the proposed volume rendering, the boundary surface of 3-D ultrasound images is defined as a region with height and width greater than some user-defined thresholds along the view direction. If a voxel has a larger gray level than the predefined noise threshold and the voxel has neighbor voxels whose gray levels are also larger than the noise threshold and the width of the neighbor voxels is wider than the predefined threshold, then the voxel can be included in boundary voxels. The boundary voxels have uniform opacity value, $\alpha_v$, in the proposed volume rendering. The noise threshold is determined by the mean value of non-zero voxels. The location of the starting point of boundary along the ray direction is utilized for the ray casting and depth cueing. Figure 2 illustrates examples of the adaptive boundary detection. In Fig.2, $z_{buff}$ indicates the starting point of boundary voxels along the ray direction, so that the ray casting can be performed from the voxel at $z_{buff}$. The proposed method can make an improvement in both classification and ray casting by searching for the boundary surface effectively and reasonably.

Also, the number of voxels participating for the ray casting is determined by the accumulated transparency, $T_a$, which is defined as follows,

$$T_a(i,j)_K = \prod_{m=z_{buff}}^{K} (1 - \alpha(i,j,m))$$ (2)

where $\alpha(i,j,m)$ is the opacity coefficient at $(i,j,m)$, and $z_{buff}$ is the starting point of boundary surface along the view axis, $m$. If the accumulated transparency at $m = K$ is smaller than a given threshold, the voxels of $m > K$ can not contribute to the volume-rendered image. The lower the threshold of the accumulated transparency is, the larger the number of voxels participating in ray casting is.

2.4 Ray Casting

The ray casting integrates the colors and opacities of voxels along the view direction as a ray penetrates the object. There have been several techniques to speed up the ray casting, such as adaptive subsampling[13], progressive refinement and hierarchical data structures[14]. To speed up the ray casting in this paper, a boundary surface is detected by the adaptive boundary detection method in Sect.2.3. Since the voxels which locate in front of the object boundary $z_{buff}$ along the view direction do not contribute to the volume rendering, skipping the voxels in front of the object boundary could provide significant speed up without affecting image quality. Early termination of ray casting is also used for computational efficiency. The threshold of accumulated transparency for early termination is 0.02 in this paper. The ray casting is formulated with the adaptive start and termination as follows,

$$I(i,j) = \sum_{k=z_{buff}}^{z_{buff}+\beta} \left[ c(i,j,k)\alpha_v \prod_{m=z_{buff}}^{k} (1 - \alpha_v) \right]$$ (3)

where $\alpha_v$ is the uniform opacity for boundary voxels, $I(i,j)$ is the volume-rendered image with view drec-
tion of \( k \), and \( c(i,j,k) \) is the color value at \((i,j,k)\). In Eq. (3), \( \beta \) indicates the termination point that is determined by thresholding the accumulated transparency.

Equation (3) shows that the greater the threshold of the accumulated transparency is, the smaller the number of voxels participating in ray casting is. Finally, the volume-rendered image is modified by a depth cueing with \( z_{buff} \). Depth-cued image is visually more realistic, and is defined by the following equation,

\[
I_d(i,j) = I(i,j) \left\{ s_0 + (s_1 - s_0) \frac{D_{\max} - Z(i,j)}{D_{\max} - D_{\min}} \right\}
\]

where \( I_d(i,j) \) is the depth-cued image, \( I(i,j) \) is the original volume-rendered image, and \( D_{\min} \) and \( D_{\max} \) denote the distances from viewpoint to foreground and background, respectively. In Eq. (4), \( Z(i,j) \) is the value of \( z_{buff} \), and \( s_0 \) and \( s_1 \) are the minimum and maximum gray levels, respectively for control of depth cueing effect. It is obvious that \( I_d(i,j) \) decreases when the depth value \( Z(i,j) \) increases; thus, the closer to the background the pixels are, the dimmer they appear.

3. Simulation Results

The dataset used in this simulation is a 148\( \times \)194\( \times \)112 voxel image (0.425 \times 0.425 \times 0.425 millimeters) of fetus acquired from ‘Voluson 530’ ultrasound system by Kretz in Austria. In these experiments, two different preprocessing methods were tested and compared, which are 3-D median filtering followed by 3-D Gaussian filtering and 2-D truncated-median filtering followed by 3-D Gaussian filtering. Also \( z_{buff} \) value is determined by thresholding with noise threshold. Next, the two opacity formulas of Levoy, such as the isovalue contour surface and the region boundary surface, are compared. The most time-consuming operation was the preprocessing. In these experiments, the noise threshold and the width used in the boundary estimation were 40 in 8-bit gray level image and 3, respectively.

Figure 3 shows the geometry of ultrasound data and the successive image slices. In Fig. 4, the volume-rendered images were obtained by two different preprocessing methods such as 3-D median filtering followed by 3-D Gaussian filtering and 2-D truncated-median filtering followed by 3-D Gaussian filtering to reduce noises in ultrasound images. For this rendering,
constant opacity values were assigned to the boundary voxels. Table 1 presents the computation times for preprocessing of 148 × 194 × 112 voxels dataset. Figure 5 illustrates the volume-rendered images from the fixed thresholding method in determining of the object boundary. In Fig. 5, the fixed threshold values are arbitrarily selected. For acceptable image quality, many iterations are needed to select a suitable fixed threshold value and the selection is not robust in other 3-D ultrasound images. From the criteria on the image quality and preprocessing time, 2-D truncated-median filtering followed by 3-D Gaussian filtering with adaptive boundary detection is more reasonable than other combinations of prior methods.

Table 1 Comparison of computation times for preprocessing of 148 × 194 × 112 voxel image by using Pentium 120MHz with 64 Mbytes memory.

<table>
<thead>
<tr>
<th>Filter Type</th>
<th>Preprocessing Time</th>
</tr>
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<tbody>
<tr>
<td>3 × 3 × 3 median</td>
<td>39.65 sec</td>
</tr>
<tr>
<td>3 × 3 truncated-median</td>
<td>10.88 sec</td>
</tr>
</tbody>
</table>

Fig. 5 Volume-rendered images from fixed thresholding for region boundaries. Preprocessing was performed by 2-D truncated-median + 3-D Gaussian filtering, and region of ray casting was \([z_{buffer} \sim (z_{buff} + \beta)]\) and a constant opacity \(\alpha_v = 0.6\). The fixed thresholds are (a) \(f_\omega = 50\), (b) \(f_\omega = 70\), (c) \(f_\omega = 90\), and (d) \(f_\omega = 110\).

4. Conclusions

Two major reasons for the limited acceptance of 3-D ultrasound are the low signal-to-noise ratio and the fuzzy nature of the object boundaries. This paper presented two improvements for 3-D volume rendering of ultrasound images: One is to apply the 2-D noise-reduction filtering with fast computing time. The other is to extract the boundary surface adaptively so that the result is utilized in the classification and the ray casting to get the volume-rendered image with fast computation time. Given a signal corrupted by Rayleigh noise like speckle, the mode is equivalent to a maximum likelihood estimator and is implemented by using the truncated-median filter to remove speckle noise. The truncated-median filter in 2-D ultrasound image offers an acceptable performance and processing time. Most volume rendering methods use a fixed thresholding in classification of the medical data. However, ultrasound image is not suitable for the conventional classification methods. In order to adapt the fuzzy nature of boundary surface of ultrasound image, an adaptive boundary detection was proposed. As a result, the proposed scheme makes an improvement in processing time with an acceptable 3-D image quality.

References


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