A Progressive Fast Volume Rendering for Medical Images

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ABSTRACT

There are various 3D visualization methods such as volume rendering and surface rendering. The volume rendering (VR) is a useful tool to visualize 3D medical images. However, a requirement of large computation amount makes it difficult for the VR to be used in real-time medical applications. In order to overcome the large computation amount of the VR, we have developed a progressive VR (PVR) method that can perform the low-resolution VR for fast and intuitive processing and use the depth information from the low-resolution VR to generate the full-resolution VR image with a reduced computation time. The developed algorithm can be applicable to the real-time applications of the VR, i.e., the low-resolution VR is performed interactively according to change of view direction, and the full-resolution VR is performed once we fix the view direction. In this paper, its computation complexity and image quality are analyzed. Also an extension of its progressive refinement is introduced.

Keyword: Visualization, volume rendering, progressive method, 3D medical images, spatial correlation

1. INTRODUCTION

Scientific visualization plays an important role in the study of multidimensional data. Scientific visualization uses computer graphics techniques to give a scientist insight. Insight is usually achieved by extraction of scientifically meaningful information from numerical descriptions of complex phenomena through imaging systems. Volume visualization is the sub-field of scientific visualization, which is a process of projecting a multidimensional data set onto a 2D image plane for visualizing the volumetric data¹. In general, the 3D data set is composed of a scalar value or vector value at every grid point on the 3D lattice. Volume visualization techniques must offer understandable data representations as well as fast data manipulation.

Volume visualization method has been studied for medical applications and well established as a clinical tool. Several methods have been developed for volume visualization of 3D medical data. Volume rendering (VR) method is characterized by mapping elements directly into image plane without geometric information². It is especially appropriate for creating semi-transparent images from fuzzy data sets like medical volumetric data, which are computer tomography (CT), magnetic resonance imaging (MRI), and ultrasound images. Surface rendering method³ is typically faster than VR method since surface-rendering method only traverses the volume once to extract surface primitives, but the VR method must traverse the entire data set whenever an image is rendered. In spite of this shortcoming, VR has become one of the most interesting and fast growing areas in medical imaging because of its superior image quality and interior projection. The quality of surface rendered images is highly dependent on the performance of a segmentation process that is difficult and highly affected by noise, whereas the VR does not require the segmentation process.

To overcome the large computation amount and improve interactivity in VR, many optimization methods as well as several special-purpose VR systems have been developed. Among them, a progressive approach⁴ is appropriate for a real-time application, which makes the resolution and quality of a VR image to be improved successively. In order for the VR techniques to be widely used, users should be able to interactively change parameters such as view direction and to see the rendered image instantly for real-time applications.

In this paper, we present and evaluate a new progressive approach to accelerate the VR process. A progressive volume rendering (PVR) method has been developed to visualize volumetric data, which is frequently used in medical applications. This paper is organized as follows: Section 2 describes the degree of voxel contribution to VR, and section 3 describes the spatial correlation of medical images. The proposed PVR is described in section 4. In section 5, experiment results are

presented. Finally, section 6 gives the conclusions.

2. DEGREE OF VOXEL CONTRIBUTION TO VR

Since 3D medical images usually have many voxels, if all the voxels participate in the generation of each image, rendering time increases linearly with the size of the dataset. However, all the voxels in 3D medical images are usually not necessary for VR. Until a ray encounters the first voxel that is classified into an object, the ray progresses along the view direction without any rendering process. As soon as a ray encounters the first object voxel, VR process is performed on the voxels along the ray until ray termination² as shown in Figure 1, where $Depth_1$ and $Depth_2$ are the positions of start of rendering and ray termination, respectively.

Voxels, whose opacities^{2,5} are larger than a low-opacity threshold and before the ray is terminated, participate in VR, where the low-opacity threshold and the accumulated-opacity threshold for ray termination are user-defined parameters characterizing the VR image. The other voxels are redundant for the VR since such voxels make negligible contribution to the rendered image. In VR processing, discriminating the redundant voxels from overall voxels consumes much computation amount. If there is pre-information to skip the redundant voxels, the VR processing time can reduce so much. We measured the rates of effective voxels that contribute to VR, and Figure 2 describes that the rates range from 3% to 15% of total voxels with the low-opacity threshold of 0.05 for starting of VR process and the accumulated-opacity threshold of 0.95 for ray termination, which are used in this paper.

Figure 1. Region of interest for VR

Figure 2. The rates of effective voxels for VR

3. SPATIAL CORRELATION OF MEDICAL IMAGES

Neighbor pixels in most images are similar and highly correlated with each other, so gray-level difference between neighbor pixels is mostly small except for object boundaries in the image. In particular, medical images such as X-ray CT, MRI, and ultrasound images have very high spatial correlation. This inter-pixel correlation in 2D image can be extended to inter-voxel correlation in 3D image.

If the spatial correlation of an image is high, the high-frequency components of the image are very small. Thus, the sub-sampling process of an image with high spatial correlation does not lose much information⁶. The high spatial correlation property of an image is a fundamental feature for image compression, and for multi-resolution visualization⁷. This paper also utilizes the high spatial correlation property to develop a progressive VR method.

In order to investigate the correlation of neighbor voxels, we choose a measure, which is a correlation coefficient, defined as,

$$\rho = \frac{E[(f(x, y, z) - \bar{f})(f(x+1, y+1, z+1) - \bar{f})]}{E(f(x, y, z)^2) - \bar{f}^2} \quad (1)$$

where $f(x, y, z)$ is a voxel value at 3D coordinates of (x, y, z) , $\bar{f}$ is the mean value of $f(x, y, z)$, and $E[\cdot]$ is the expectation function. The correlation coefficient ρ shows a correlation between just diagonal neighbor voxels. Medical images usually have the correlation coefficient larger than 0.97 as shown in Figure 3. This means the correlation of medical image is usually high compared to the Lena and airfield images that are generally used in the image processing area.

Figure 3. Spatial correlation coefficients of various 3D and 2D images

4. PROPOSED PROGRESSIVE VOLUME RENDERING

4.1. OVERALL BLOCK DIAGRAM

Figure 4 shows the overall block diagram of the proposed PVR, which consists of three parts such as preprocessing, the first step, and the second step. The first step is to generate a low-resolution VR image from the sub-sampled voxel data for taking an outlook, and the second step is to obtain a full-resolution VR image by using the low-resolution depth information, $(Depth_1, Depth_2)$, which is obtained from the first step. Detail procedures of the PVR are described in this section.

Figure 4. Block diagram of the proposed VR algorithm

4.2. PREPROCESSING

The proposed algorithm requires a preprocessing step to reduce computation time effectively, where 3D image data are sub-sampled by two-fold decimation in x, y, and z directions with an average value of a $2 \times 2 \times 2$ block⁸ as follows,

$$f_2(x, y, z) = \frac{1}{8} \sum_{i=0}^1 \sum_{j=0}^1 \sum_{k=0}^1 f(2x+i, 2y+j, 2z+k) \quad (2)$$

where f and f_2 are the original full-resolution and the sub-sampled 3D images, respectively.

The average value of the $2 \times 2 \times 2$ block is correlated with the average values of the adjacent blocks as well as the original voxel values in the block. Since spatial correlation of an image is high, sub-sampling process of the image does not lose much information⁶.

Data classification is the most difficult function that a VR user has to perform. The essential task of classification is to calculate opacity for each voxel. Among various equations to present opacity in VR, the opacity equation using only gray level such as Eq. (3)⁵ and the opacity equation using gray level and gradient such as Eq. (4)² are widely used. Since these opacity functions classify the object fuzzily, not discretely, the image of semi-transparent surfaces can be generated.

$$\alpha_1(\mathbf{x}_i) = \alpha_v \begin{cases} \frac{f(\mathbf{x}_i) - f_{\min}}{f_{\max} - f_{\min}}, & \text{if } f_{\min} \leq f(\mathbf{x}_i) \leq f_{\max} \\ 0, & \text{otherwise} \end{cases} \quad (3)$$

$$\alpha_2(\mathbf{x}_i) = \alpha_v \begin{cases} 1, & \text{if } |\nabla f(\mathbf{x}_i)| = 0 \text{ and } f(\mathbf{x}_i) = f_v \\ 1 - \frac{1}{r} \frac{|f(\mathbf{x}_i) - f_v|}{|\nabla f(\mathbf{x}_i)|}, & \text{if } |\nabla f(\mathbf{x}_i)| > 0 \text{ and} \\ 0, & f(\mathbf{x}_i) - r|\nabla f(\mathbf{x}_i)| \leq f_v \leq f(\mathbf{x}_i) + r|\nabla f(\mathbf{x}_i)| \\ & \text{, otherwise} \end{cases} \quad (4)$$

where α_v is the maximum value of an opacity, $f_{\min}$ and $f_{\max}$ are the minimum and the maximum values of interesting gray-level range, respectively, f_v is a selected gray value to be visualized, and r is the desired thickness in voxels of the transition region². In Eqs. (3) and (4), $f(\mathbf{x}_i)$ and $|\nabla f(\mathbf{x}_i)|$ are the gray value and the gradient value at position vector $\mathbf{x}_i$, respectively. In this paper, the opacity equation of Eq. (3) is adopted for experiments. The opacity is calculated and stored into memory as a lookup table at preprocessing. If a user changes parameters during observation, the opacity table can be modified in very short time.

4.3. FIRST STEP

A disadvantage of using VR method is that the entire dataset must be traversed whenever an image is rendered. A low-resolution pass or random sampling of the data¹ is sometimes used to quickly create low-resolution images, so that

parameter values and view direction are interactively analyzed.

As shown in section 3, a voxel value is usually correlated with neighbor voxels. Therefore, the opacity of each voxel has correlation with the opacities of the neighbor voxels. Using spatial correlation of 3D medical images, the neighbor voxels have similar opacities, which means that opacity of the sub-sampled voxels is similar to that of full-resolution voxels. Therefore, the opacity of the sub-sampled voxels is multiplied by two in order to simulate the full-resolution VR image with the sub-sampled voxels.

A low-resolution VR image is obtained from the sub-sampled 3D data having an eighth of the original 3D data in data amount, and it is effective for fast processing and adaptive to taking an outlook over the volume data for searching an interest view or region of interest (ROI).

Each voxel's contribution to the pixel of an image is calculated and composited by using an opacity table. Ray progress used in this paper is a front-to-back traverse of the voxels along view direction in the volumetric dataset.

In the first step of sub-sampled VR, the depth information consists of the position of start depth, $Depth_1$, where a ray encounters the voxel whose opacity is larger than a low-opacity threshold, α_{min} , and the position of end depth, $Depth_2$, where the ray is terminated as shown in Figure 5, where $\alpha(x_i)$ is the opacity of a sub-sampled voxel, α_{acc} is an accumulated opacity, and α_{max} is an accumulated-opacity threshold.

Once an opacity of a ray in the ray-casting volume rendering becomes larger than a given low-opacity threshold that is 0.05 in this paper, the rendering starts and the start depth information of the voxel is stored as Eq. (5) and once an accumulated opacity⁴ of a ray in the ray-casting volume rendering becomes larger than a given accumulated-opacity threshold that is 0.95 in this paper, the ray-casting is terminated to save the computation time and the terminated depth information of the voxel is stored as Eq. (6).

$$Depth_1(u, v) = d(\mathbf{x}_M) \text{ for the smallest } M \text{ such that } \alpha(\mathbf{x}_M) \geq 0.05 \quad (5)$$

$$Depth_2(u, v) = d(\mathbf{x}_N) \text{ for the smallest } N \text{ such that } \alpha_{acc} = 1 - \prod_{i=1}^N (1 - \alpha(\mathbf{x}_i)) \geq 0.95 \quad (6)$$

where $Depth_1(u, v)$ and $Depth_2(u, v)$ are the depth values on the image space (u, v) , $d(\mathbf{x}_N)$ is the distance from a voxel at $\mathbf{x}_N$ to the image space (u, v) , and $\alpha(\mathbf{x}_i)$ is the opacity at $\mathbf{x}_i$. In eq. (1), the voxels of $\mathbf{x}_i$, $i=1, 2, \dots, N$, are on a ray projected to (u, v) .

Figure 5. The acquisition of the depth information for VR

Figure 6. Criteria for decision of full resolution voxels contributing to the full-resolution VR

4.4. SECOND STEP

To observe the interest view or ROI in full resolution, the image quality is improved by processing from the first step to the second step. In the second step, the depth information from the first step is compared with voxel depth of the full-resolution data. The full-resolution voxels corresponding to the sub-sampled voxels that are located between the start depth, $Depth_1$, and the end depth, $Depth_2$, with a given depth margin, d , are selected as shown in Figure 6, where the depth margin of d is a user-defined parameter to improve the VR image quality to the degree how much a user can be satisfied. However, the second step with the larger depth margin requires the longer computation time. The depth margin provides a quantitative tradeoff between the image quality and the computation amount. The selected full-resolution voxels contribute to the full-resolution VR. A full-resolution volume-rendered image is obtained from only the selected full-resolution voxels so that the computation amount can be reduced. As a result, the value of depth margin has an influence on the peak signal to noise ratio (PSNR) and the computation amount of the developed PVR.

According to view direction, the pixel coordinate, (u, v) , on a volume-rendered image has a relation with a 3D voxel coordinate, (x, y, z) , in a volume space as follows,

$$\begin{aligned} u &= x \cos \phi + y \sin \theta \sin \phi + z \cos \theta \sin \phi \\ v &= y \cos \theta - z \sin \theta \end{aligned} \quad (7)$$

where (θ, ϕ) is the view direction of the VR in the case that the first rotation is counterclockwise about the x axis with angle of θ and then the second rotation is counterclockwise about the y axis with angle of ϕ .

Shear-warp decomposition method¹⁰ is applied to VR process in this paper. After compositing the sheared volume data, an intermediate image that is geometrically distorted is produced. The intermediate image requires a 2D warping¹⁰ according to the view direction to reconstruct the correct VR image.

5. EXPERIMENT RESULTS

5.1. EXPERIMENTAL ENVIRONMENTS

There are four different 3D medical images to be used in the experiment, which include X-ray CT, MRI, and ultrasound images with various resolutions. Table 1 shows the image sizes such as width, height, and depth in order and the required computation amounts of compositing and warping of the shear-warp VR, where the warping procedure has much less computation in comparison with the compositing one. The view direction of all the data set is 30 degree on the downward direction, i.e. $(\theta, \phi) = (30^\circ, 0^\circ)$. The experiments were performed on a SUN UltraSPARC 1 workstation, and the computation amount was measured by using IPROF - Instruction Profiler¹¹. The IPROF calculates the total instruction usage statistics for every opcode including the arithmetic instructions, the jump/test/compare instructions, the load/store instructions, the floating-point instructions, and other instructions. The computation amount in this paper does not include preprocessing such as sub-sampling and opacity table generation.

Data	View angle (θ, ϕ)	Data Size	Computation amount	
			Compositing	Warping
Head CT	(30°, 0°)	256×225×256	650.84	5.34
Head MRA	(30°, 0°)	256×256×76	242.61	4.30
Body MRA	(30°, 0°)	256×256×96	324.81	4.47
Face US	(30°, 0°)	148×194×112	134.25	2.14

Table 1. Computation amount in million instructions of the shear-warp VR for 3D medical images

5.2. EXPERIMENTAL RESULTS

Figures 7 and 8 show the PSNR and the computation amount for compositing of the PVR, respectively, with respect to the depth margin. When the depth margin, d , becomes large enough, the PVR performs the second step with the overall full-resolution voxels. Therefore, the computation amount of the PVR with large depth margin may be larger than the conventional shear-warp VR, and the PVR image is just same as the conventional shear-warp VR image.

Figure 7. Comparison of PSNR of the proposed PVR with respect to the depth margin

Figure 8. Rate of computation amount of the proposed PVR compared with the conventional shear-warp method with respect to the depth margin

The PVR images of head CT and body MRA are shown in Figures 9 and 10, respectively. In the figures, relative computation amounts are described, which are the ratios compared to those of the conventional shear-warp VR. The computation amounts of the first-step PVR for head CT, head MRA, body MRA, and ultrasound images of fetus' face are

67.4, 26.95, 33.57, and 14.86 million instructions, respectively. The computation amounts of the first-step PVR are about 11% of those of the conventional shear-warp VR. Figures 9(a) and 10(a) show the results of the magnified first-step PVR image of sub-sampled data whose PSNR are 35.9 dB and 28.0 dB, and computation amounts are 10.3% and 10.4% of the conventional shear-warp, respectively, whose image quality and computation amount are acceptable for overview. Figure 9(b) and 10(b) are the results of the second-step PVR with depth margin of ∞ which are just same as the conventional shear-warp VR images. Figure 9(c) and 10(c) are the results of the second-step PVR with depth margin of 2 whose PSNR are 45.4 dB and 39.9 dB, and computation amounts are 34.8% and 45.0% of conventional shear-warp, respectively. The images are comparable to the conventional shear-warp images.

Figure 9. PVR images of head CT: (a) 1st-step low-resolution PVR image, (b) 2nd-step full resolution PVR image with depth margin of ∞ , and (c) 2nd-step full-resolution PVR image with depth margin of 2

Figure 10. PVR images of body MRA: (a) 1st-step low-resolution PVR image, (b) 2nd-step full-resolution PVR image with depth margin of ∞ , and (c) 2nd-step full-resolution PVR image with depth margin of 2

Figure 11. The depth information error caused by opacities of average voxels in sub-blocks

5.3. ERROR ANALYSIS AND SPATIAL CORRELATION

Figure 12. Comparison of PSNR at depth margin 2 according to the view direction

There are some drawbacks in the progressive VR method, which are caused by sub-sampling of voxels. Sub-sampling is to select one representative value from the neighbor eight voxels. Even the image has mostly low-frequency components, i.e., it is highly correlated in spatial domain, the sub-sampling process may lose high-frequency information. For example, sub-sampled values may have large difference from full-resolution values at object boundaries where the gray value changes abruptly as shown in Figure 11. This difference can make error in the PVR when the given depth margin does not overcome the difference. However, the image quality is not strongly dependent on the view direction since the spatial correlation between voxels is not much dependent on direction. Figure 12 shows the PSNR at depth margin 2 with respect to view direction. It shows the view direction does not affect strongly on the PSNR.

5.4. PROGRESSIVE STRUCTURE APPROACH

The process of successively increasing the resolution and quality of a VR image over time is called progressive refinement¹. Until now, the progressive refinement has been approached to using a pyramid volume representation^{7, 12}.

However, in order to improve the PVR image quality in this paper, we can perform refinement of the PVR image using depth margin increment. Block diagram of the refinement process of the PVR is presented in Figure 13. The second-step PVR is rendered from D_1 to D_2 without depth margin, the front-refinement of PVR is rendered from D_1-d to D_1 , and the rear-refinement of PVR is rendered from D_2 to D_2+d . Three rendered images are combined together to generate a final refined PVR image as follows,

$$I(x, y) = I_{fr}(x, y) + I_{pvr}(x, y)(1 - \alpha_{fr}(x, y)) + I_{rr}(x, y)(1 - \alpha_{fr}(x, y))(1 - \alpha_{pvr}(x, y)) \quad (8)$$

where $I(x,y)$, $I_f(x,y)$, $I_{PVR}(x,y)$, and $I_r(x,y)$ are the final refined, front-refinement, the second-step, and rear-refinement PVR images, respectively. In Eq. (8), $\alpha_f(x,y)$ and $\alpha_{PVR}(x,y)$ are the accumulated opacity values of front-refinement and second-step PVR, respectively. As a result, the high quality VR image is obtained from the refinement process.

Figure 13. Block diagram of refinement process of the PVR

6. CONCLUSIONS

In this paper, we proposed a real-time volume rendering method with a progressive structure using volume homogeneity. There are two steps in the developed PVR except for preprocessing: In the first step, fast and low-resolution VR image can be obtained in real time, which is obtained from sub-sampled voxels that are made from preprocessing. In the second step, a full-resolution VR image is obtained by using depth information obtained from the first-step VR. The proposed PVR method can save the computation time of the VR process, whereas the PVR image quality is compatible with the conventional shear-warp VR image, and there is a quantitative tradeoff between image quality and computation amount with respect to the depth margin. Due to the high correlation of medical images, the image quality is not much affected by the change of view direction. In addition, it is possible to improve the volume-rendered image quality progressively using effective refinement technique of the PVR.

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