Medical Informatics Europe '97
C. Pappas et al. (Eds.)
IOS Press, 1997

A 3D ultrasound scanner : real time filtering and rendering algorithms

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Abstract
The work described here has been carried out within a collaborative project between DIST and ESAOTE BIOMEDICA aiming to set up a new ultrasonic scanner performing 3D reconstruction . A system is being set up to process and display 3D ultrasonic data in a fast , economical and user friendly way to help the physician during diagnosis. A comparison is presented among several algorithms for digital filtering , data segmentation and rendering for real time , PC based , three-dimensional reconstruction from B-mode ultrasonic biomedical images .Several algorithms for digital filtering have been compared as relates to processing time and to final image quality . Three-dimensional data segmentation techniques and rendering has been carried out with special reference to user friendly features for foreseeable applications and reconstruction speed .

1. Introduction

Nowadays there is a great interest toward ultrasonic imaging due to its versatility .
Ultrasonic scanners are very easy and fast to use and they are cheap compared to other tomographic devices . Moreover , ultrasounds is the only imaging technique available today which is non invasive without restrictions .Three dimensional ultrasonic imaging is at present one of the most interesting fields for industries . A collaborative project between DIST and ESAOTE BIOMEDICA is being carried out , aiming to realize a new high-performance 3D ultrasonic scanner . The present work focuses on the algorithmic aspect of the system . Whereas three-dimensional reconstruction from tomographic data is a very common and widely studied technique for several imaging methods like X-CT , MRI and PET , this is not the case for B-mode ultrasonic data which is regarded as a difficult and challenging field , due to strong limitations in the imaging system compared with other tomographic techniques. In ultrasonic imaging there are several noise sources and aberrations which cause great difficulties in the image processing and data segmentation steps , before rendering . Signal to noise ratio is very low compared to other imaging methods , ultrasonic biomedical images are corrupted by speckle and thermal noise , there is a general lack of contrast and several parts of the field of view are often out of focus .
In image processing of ultrasound data linear filters are generally useless , so we have considered different types of median filters such as 3D median and adaptive median filtering [1] which represent a good trade off between computational complexity , noise cleaning and edge preserving ability. Regarding data segmentation we have developed an almost totally automatic algorithm based on pseudo-3D *region growing* to simplify the reconstruction process for the physician and to maintain three-dimensional coherence between the different structures in the data set . For the last step in the reconstruction process we have used two different classes of rendering algorithms : solid voxel based photorealistic techniques like *Back to Front* [2] and *Front to Back* , together with *depth shading* and *gradient shading* [3,4] to enhance final image quality , and non photorealistic techniques based on direct features mapping [5] like *mean value projection* , *maximum value projection* and

transparent gradient shading .The reason why we have followed both approaches is that solid techniques are useful for morphology analysis on reconstructed structures , whereas direct feature mapping rendering gives a look-thru projection in some way similar to X-ray because all the information coming from the 3D data set is condensed using transparency on a two-dimensional view plane .
This semi-transparent representation is appreciated by physicians because it is closer to their cultural background than the photorealistc solid reconstruction .

2. Materials and methods

To acquire the data set in the beginning we used a conventional bidimensional ultrasonic probe operating in free hand mode . With this setup it was only possible to perform parallel acquisitions because there was no information about the spatial relationship between the different slices acquired. Later we have used a 6 degree of freedom position sensor , Polhemus Flock of Bird , directly attached on the ultrasonic probe .This makes it possible to acquire the sections of the data set with any position and orientation for the probe , then with a three-dimensional interpolation algorithm it is possible to give the data set homogeneous density filling the missing voxels .
For processing and rendering steps we have used a conventional , hi-end personal computer based either on a Pentium or on a Pentium Pro CPU with Windows 95 operating system .

3. Filtering

Working with ultrasonic images , linear filters are almost useless , the main result is a big information loss due to an excessive blurring which modifies the edge content of the processed image . A good filter for ultrasonic data must have a strong noise suppression ability without affecting edges , wich in ultrasound tomography are often not very visible . Therefore non linear filters have to be considered . We have examined several kinds of *median filters* and scale space filtering based on *anisotropic diffusion* [6] . The next step will be *wavelet* filtering , which will be compared with median filtering and anisotropic diffusion . Computational complexity for anisotropic diffusion is too heavy for a real time application based on commercial hardware , like in our case . Since this kind of filtering is an iterative process , the number of iterations required to get good results is too high for our case , although it's a powerful method well suited for ultrasonic data , and for applications without strong time constraints . We have focused on median filters , which represent a good trade off between CPU time , noise suppression and edge preserving ability. It's possible to obtain pretty good results even using the bidimensional median filter with a 5 by 5 or a 7 by 7 mask , but as we are working in a three-dimensional environment it is possible to get better results using a 3D median filter with a 3 by 3 by 3 or a 5 by 5 by 5 mask . The advantage in using the 3D median instead of the 2D one is due to the fact that spatial correlation for noise between different sections of the data set is much lower compared to real signal correlation , thus working with a 3D filter we gain noise suppression ability .

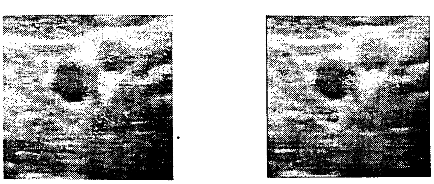

Figure 1 Left , original ultrasonic image of a carotid . Right , same image filtered with a 3 by 3 by 3 median.

Normal median filters are able to preserve ideal edges while modifying noise corrupted ones , so even with 3D median filter there is edge blurring , which can be a problem for the segmentation and rendering steps. There is a possible extension of the median filter [1] which is able to preserve image detail also increasing noise suppression .This filter is an adaptive median which can modify its mask size depending on the local statistical image content . In large and homogeneous areas the largest possible window is used , obtaining maximun noise suppression , in detail areas the window size is reduced so to retain important structures present in the image .The parameter used to fix the window size is $\frac{\sigma_x^2}{\bar{x}}$ where variance and mean are calculated on the maximum window , which is 2k+1 by 2k+1. For each pixel the size of the window for the median used is given by $n = \left[k\left(1 - c\frac{\sigma_x^2}{\bar{x}}\right)\right]$, where c is a scaling factor. The parameter c regulates the filter bandwidth , consequently noise suppression and edge preserving capability , and also processing time . For values of c close to 0 the filter behaviour is similar to the one of a pure median , so the maximum window size is always used , obtaining a high noise suppression but the processing time is long and the edges are not preserved to a good extent , for higher values of c noise suppression is lower , but edge preserving ability is higher . As a good compromise between speed and final image quality , we have used c=0.30 and maximum size 9 for the mask.

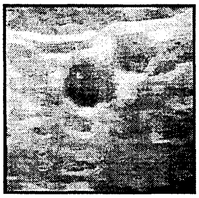

Figure 2 Same image filtered with adaptive median c=0.30 , mask size 9.

For a good 3D reconstruction the whole data set must have at least 64 sections .On a Pentium 120 PC with the 3D median filtering is performed in about 1 minute , with the variable window median it takes about three minutes , which are both reasonable times , since filtering is needed only once per acquisition , whereas segmentation and rendering can be repeated and modified without need of preprocessing . We will also consider the use of *wavelets* as low computational complexity non linear filtering , to gain speed in the noise cleaning step for 3D reconstruction .

4. Segmentation

Segmentation is important to choose the structure to be reconstructed from the whole data set . We have developed an almost totally automatic segmentation technique based on *region growing* , which is able to classify the 3D data set into regions assigning the same label to corresponding parts in different sections . The only parameter needed for segmentation is the number of regions that we want to use to segment the data set .

Region seeds are chosen in the same way for the whole data set , to give three-dimensional coherence to the segmented data . To increase noise immunity seeds are not chosen from real voxels in the data set , but the gray level range , 0 ... 255 , is partitioned depending on the number of regions chosen for segmentation . Regions are then created on the basis of minimum gray level distance from one of the available seeds . Practically using different data sets we have found that for ultrasonic images 4 or 6 regions are enough to obtain good reconstructions , anyway for a general data set the right number of regions for segmentation can be found with few trials . Regions found with this technique are generally not connected and often contain both interesting and unwanted structures ; great improvements can be obtained first using *spatial clipping* and then filtering among the remaining structures those whose area is below a threshold.

Figure 3 figure 1 image segmented with 4 regions

This algorithm is simple and requires low processing times , so it's well suited for a fast reconstruction application . The whole data set is processed in a few seconds and with very good results for the final rendering .

5. Rendering

Due to cheap commercial hardware and high speed constraints we have focused on voxel based rendering algorithms because their computational complexity is much lower than surface based or ray tracing algorithms . We have implemented both photorealistic solid methods and non photorealistic transparent algorithms . The first technique we have used is *Back to front* [2] . Voxels are projected on the screen and the data set is transversed with decreasing distance from the observer , so voxels far from the observer are covered by closer ones . The first depth cue used is depth shading : the intensity of each voxel projected on the screen is inversely proportional to view plane distance , voxels close to the observer are given a high intensity while a low intensity is chosen for far voxels . To gain more details depth shaded images have been processed with two different *gradient shading* algorithms based on the *Z-buffer* [3,4], using Phong illumination model [3] . The depth shaded image is completed in 2 seconds , the first gradient shading algorithm requires less than 1 second , while the second , which is more complex but gives better results , requires between 4 and 6 seconds depending on the parameters used .

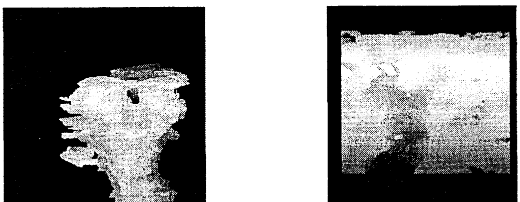

Figure 4 Left reconstruction of a carotid with gradient shading , right reconstruction of a carotid section with depth shading, both using back to front.

The second method used is *Front to Back* , which gives the same kind of reconstruction of Back to front but it's faster is some cases . The data set is transversed with increasing distance from the observer using rays which are normal to the view plane , the data set is often not completely analyzed resulting in a faster processing time for reconstruction .In addition depth shading and gradient shading techniques are used to get more structural details . With front to back it's possible to save about 30 percent of processing time. Solid rendering algorithms are useful to analyze organ structure looking for deformations or to check organs for correct dimensions . Moreover , physicians are particularly interested in transparent methods which make look-thru projection possible .Using this kind of imaging technique , which is much more similar to X-ray , it's possible to compare data coming from different imaging methods . For instance , in digital mammography interesting results are sometimes obtained by comparison with ultrasounds .The first transparent rendering technique we have used is *mean value projection* [5] , in which the data set is transversed with rays which are normal to the view plane like in the front to back case and the mean value of all the voxels along each ray is calculated and then projected onto the screen .

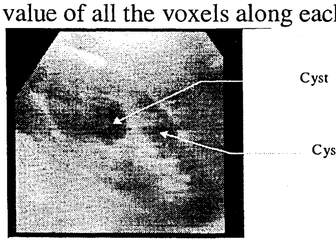

Figure 5 Mean value projection of a kidney with two Cysts .

This technique can also be modified to provide the user with a better depth feeling : the mean value is computed weighting each voxel on the basis of distance from the view plane , in this way the voxels nearer to the observer contribute more than the others to the final reconstructed image.Mean value projection images are generally not very detailed . Animation is a good technique to resolve visual ambiguities and to achieve a better trasparency feeling. From the

computational point of view mean value projection is heavier than former methods requiring about 20 to 30 seconds for a reconstruction . *Maximum value projection* has the same structure of the previous algorithm but the maximum value along each ray is projected instead of the mean value . It's useful to underline strong acoustic impedance discontinuities in the ultrasonic data set , like bone structures , but it's a very noise sensitive method .
The last technique we have tried is *transparent gradient shading* in which there is a transparent projection of gradients magnitude along each ray rather than a projection of the data set voxels. The aim is to make a look-thru representation of the different surfaces between different tissues in the data set . Since when dealing with ultrasound to extract surfaces from the data set is very difficult , gradient calculation is used to enhance discontinuities due to surfaces .

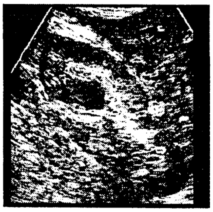

Figure 6 Previous figure kidney rendered with transparent gradient shading .

We have used a threshold to suppress small gradients along rays because these are often only noisy patterns which can cause great disturbance for final results after being integrated all along the ray. A better view of spatial relationship between tissues surfaces is achieved trough animation . Regarding processing time , this method is the heaviest among the algorithms we have tried , due to gradient calculation for each voxel , reconstruction time is about 30 to 40 seconds on a Pentium 120 .

6. Future work

For the future we will focus our attention first of all on the preprocessing step which is the main bottleneck of the system . We will work on fast and edge preserving filtering techniques based on wavelets . We are also planning to speed up all rendering algorithms . Moreover , we will focus on transparent rendering not only from the speed point of view , but also from the feature significance point of view . This will be performed in close connection with physicians , considering the high resemblance between transparent rendering and X-ray image representation .

7. References

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