Improved 3D Reconstruction Algorithm for Ultrasound B-scan Image with Freehand Tracker

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ABSTRACT

EM algorithm for the reconstruction of freehand B-Scan ultrasound image was developed by Joao M. Sanches et al. The reconstruction has a parameter κ which can be adjusted so that the results can be smoother or sharper depending to the value of κ. In order to make the image smoother inside the organs but sharper in their boundaries simultaneously, we introduced a improved EM algorithm: EM algorithm with a diffusion filer or is referred as EMD algorithm. There was a cubic average filter inside the loop of the iteration of the EM algorithm. This average filter is replaced by a diffusion filter in the EMD algorithm. The diffusion filter offers an additional parameter K_d which can be used to adjust the reconstructed image with better optimization in both smoothness insider the human organ and sharpness in its boundary.

Two above mentioned reconstruction algorithms for the freehand B-scan ultrasound image are compared through the simulation and the phantom measurements. In the simulation, strong noises are added to the ultrasound frame data. The parameters of two algorithms are optimized to get smallest errors. The errors are compared between two algorithms with optimized parameters.

For the measurement with phantom, the Eigen's tracker system is used to continuously measure the coordinates of the ultrasound probe. The ultrasound B-scan frame is synchronously recorded with the probe coordinates. Zonare ultrasound machine is used to acquire the 2D frame images. The segmentation of the reconstruction results is done. The segmentation volumes of the prostate phantoms are compared.

The results shows that EMD algorithm is better at reducing the noises and keeping the image edge comparing to EM algorithm. Eigen's tracker can acquire freehand ultrasound data for a 3D image reconstruction with high quality.

Keywords: U ltrasound, Scan, Image, Reconstruction, Interpolation, Expectation Maximization, Diffusion, Filter, Tracker, Freehand

1. INTRODUCTION

The ultrasound B-Scans are irregularly distributed in the space. There are holes where are no any measured data. Normally the ultrasound data is noisy and the noises are related to the signal. 3D ultrasound image is required to be interpolated or reconstructed from the measurements on a set of planes. Available techniques for this are Voxel-Based methods (VBM), Pixel-Based Methods (PBM) and Function-Based Methods (FBM). The PBM scans all pixels on 2D frames. One or a few surrounded voxels of that pixel will fit suitable values from that pixel. Then the holes where have no data in the space are patched. An example of PBM was given by TR Nelson et al.[2] The VBM scans all voxels in the space. A voxel get its value from the measured values of a few nearest pixels. An example of VBM was given by Qing-Hua Huang et al.[1] Huang's interpolation algorithm calculates according to the square distance. The sharpness of the organ edge and smoothness of the image are adapted through value of σ/μ, here σ and μ are deviation and mean value of the input data in the local region. There are two major contributors for the function based method (FBMs), one is the radial function based interpolation using Splines developed by Rohling et at;[3] another is expectation maximization (EM) method developed by Sanches et al.[4,5]

In Sanches' algorithm, the reconstructed image voxel is produced from an interpolation between the input data and surrounding voxels of prior iteration. A parameter κ is used to balance the sharpness and smoothness. Sanches' algorithm is enhanced by adding diffusion technique by GUO Qing et al.[6] in a rotational scanning

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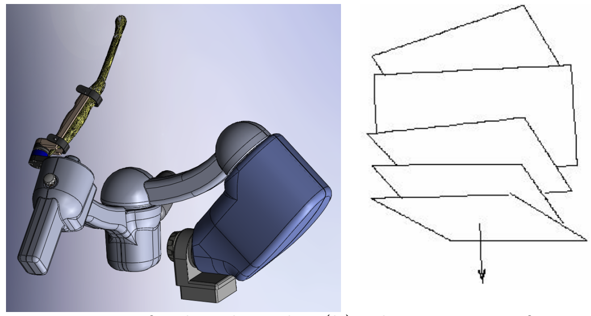

Figure 1. (a)Eugen's tracker system in freehand mode. (b) The 2D image frames of the free-hand scanning.

ultrasound system. The parameter in Sanches' method is adapted through diffusion-like technique in GUO's algorithm. In this article the EM method is combined with diffusion technique in a freehand scanning mode.

The Sanches' method is iterative. Inside a loop of the iteration the output image is obtained through a weighted addition of the ultrasound input image and the output of filtered image from last iteration. The filter normally is a cubic filter with 27 neighbor points. The EM algorithm is improved by replacing the cubic filter by an anisotropic diffusion filter in this article. To distinguish with the original EM algorithm, the improved EM algorithm is referred as EMD algorithm (EM with diffusion). The diffusion filter can be seen in reference.[7] The parameters used in two algorithms (EM and EMD) are optimized using the smallest errors. Here the error means the differences between the reconstructed image and the original object to create simulated ultrasound 2D frame data.

EM method is reviewed in section 2. The EMD algorithm is introduced in section 3. Section 4 introduces the noise mode of the ultrasound system. Section 5 discusses optimization of the parameters. In section 6 the reconstructed images from the different algorithms are compared using simulated data. In section 7 the coordinates of tracker system and the transform matrix is discussed. In section 8 the tracker system is simulated, projections data are produced and reconstruction is with this data. In section 9 a set of measured freehand 2D frame images from the prostate phantom are used as input, the images are reconstructed and compared. Here Eigen's tracker (see Figure 1) is used to acquire 2D frame images,.

2. THE EM ALGORITHM

The EM algorithm[4] is a iterative a algorithm. The updated image $u_p^{(n+1)}$is a combination of the two parts. The first is the image u_p^{ML} created from freehand ultrasound input data using a simplest interpolation. Here the simplest interpolation means the holes in the data are not patched. The second is a neighborhood average image $\bar{u}_p^{(n)}$ which is obtained from the last iteration,

$$u_p^{(n+1)} = (1 - k_p)u_p^{ML} + k_p \bar{u}_p^{(n)} \tag{1}$$

where $p = (i, j, k)$ is a set of 3D grid indexes. The input and the initialization of the iteration are defined as following,

$$u_p^{ML} = \left\{ \begin{array}{cc} \sum_l y_l \eta_p(x_l) / \sum_l \eta_p(x_l) & \exists x_l \in \Omega \\ 0 & \forall x_l \notin \Omega \end{array} \right\}, \;\; u_p^{(0)} = 0$$

where $\Omega = [i - \Delta, i + \Delta][j - \Delta, j + \Delta][k - \Delta, k + \Delta]$, x_l is the l-th pixel coordinates of a set of 2D frames in the ultrasound measurement. y_l is l-th measured value corresponding x_l. l is pixel index. Δ is the space between two voxels. We take Δ=1 for simulation. Δ will take different values for the reconstruction with measured data. $\eta_p(x) = h(x - p)$, and

$$h(x) = \left\{ \begin{array}{cc} \prod_{k=1}^{3} (1 - |x^k|/\Delta) & x \in \delta \\ 0 & otherwise \end{array} \right\} \tag{2}$$

where $x = (x^1, x^2, x^3)$. $x^k = x^1$, x^2or x^3. $\delta = [-\Delta, \Delta]^3$. k_p in (1) is defined as following,

$$k_p = \frac{1}{1 + A_p}; \quad A_p = \left\{ \begin{array}{cc} \kappa \sum_l \eta_p(x_l) & \exists x_l \in \Omega \\ 0 & \forall x_l \notin \Omega \end{array} \right\} \tag{3}$$

where κ is a constant parameter. The average in 27 neighborhood points is defined as

$$\bar{u}_p = \frac{1}{27}[\sum_{a=-1,b=-1,c=-1}^{1,1,1} u_{(i-a,j-b,k-c)}] \tag{4}$$

3. THE EMD ALGORITHM

The square filter $\bar{u}_p$ is replaced by a diffusion filter Du_p, hence we have

$$u_p^{(n+1)} = (1 - k_p)u_p^{ML} + k_p D u_p^{(n)} \tag{5}$$

The anisotropic diffusion filter[7] is defined as the following,

$$Du_p = u_p + tc\, L_\vartriangle u_p \tag{6}$$

where t is constant, $L_\vartriangle$ is Laplace operator which is defined as,

$$\begin{aligned} L_\vartriangle u_p \equiv (u_{(i-1,j,k)} - u_{(i,j,k)}) + (u_{(i,j-1,k)} - u_{(i,j,k)}) + \\ (u_{i,j,k-1)} - u_{(i,j,k)}) + u_{(i+1,j,k)} - u_{(i,j,k)}) + \\ (u_{(i,j+1,k)} - u_{(i,j,k)}) + (u_{i,j,k+1)} - u_{(i,j,k)}) \end{aligned} \tag{7}$$

c is defined as $c = g(||\frac{\nabla u_p}{m}||$. ∇ is grad operator. where m is a normalization factor. $m = \max(f(x))$ is the max value of the image. $g(x)$ has two forms as the following:[7]

$$g_1(\mathbf{x}) = \exp(-(x/K_d)^2); \quad g_2(x) = \frac{1}{1 + (x/K_d)^2} \tag{8}$$

In the following sections, only g_1 function is used. K_d is the diffusion threshold constant. In the limit situation, if we take $t = \frac{1}{7}$, $K_d \to \infty$, $g(x) \to 1$, and then,

$$Du_p \to \frac{1}{7} L_\Delta u_p + u_p = \bar{u}_p \tag{9}$$

Here the above $\bar{u}_p$ is not the 27-point neighborhood average as in 4. Instead it is a 7- point neighborhood average:

$$\begin{aligned} \bar{u}_p = u_{(i,j,k)} = \frac{1}{7}(u_{(i-1,j,k)} + u_{(i,j-1,k)} + \\ u_{(i,j,k-1)} + u_{(i,j,k)} + u_{(i+1,j,k)} + u_{(i,j+1,k)} + u_{i,j,k+1}) \end{aligned} \tag{10}$$

In this case, the effect of diffusion filter is eliminated. The EMD algorithm $\to$ EM algorithm.

4. NOISE MODE AND SIMULATED DATA

Simulated 2D frame images for B-scan ultrasound are calculated according to the following noise mode:

$$y_l = f(x_l) + f(x_l)\xi$$

where x_l is the 3D coordinates of the grid of 2D frame images. Here the plane of frames are random distributed in the space. The plane can be described as through a point $P = (P_1, P_2, P_3)$. Here the distance of the plane to the origin is $d = ||P||$. Here the normal vector is $\hat{n} = \frac{P}{d}$. We can assume P obey 3D Gauss distribution. l is the pixel index for a set of the 2D frames. $l \in L$. L is the set of grid points in all frames. $f(x_l)$ is the values of original object which is used to generate the simulated data. where is ξ a pseudo random variable satisfies uniform distribution in the region [-0.5,05]. y_l is the generated 2D ultrasound image values corresponding to grid point x_l. Figure 2 shows the simulated data.

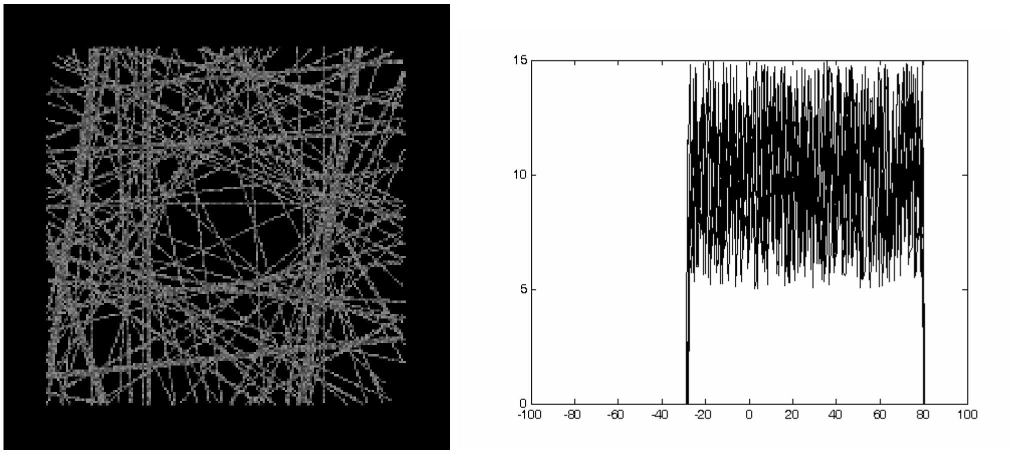

Figure 2. Left is the simplest interpolated image u_p^{ML} using the data (x_l, y_l). Here we only show the middle plane of $k = 100$. Here the 3D image size is $200 \times 200 \times 200$. Right is the profile of the left. It shows the noised data, the horizontal axis is the x_l^1, the vertical axis shows y_l which is the pixel values.

5. OPTIMIZATION OF THE PARAMETERS

The parameters used in two algorithms are optimized according to the absolute errors and square errors between the reconstructed image and the original object. The error functions are defined as following,

$$Err_m = \frac{1}{N^3} \sum_{i=0,j=0,k=0}^{N-1,N-1,N-1} |u_r(i,j,k) - u_o(i,j,k)|^m \tag{11}$$

where $m = 1, 2$ corresponding to absolute error and square error respectively. The above errors Err_1 and Err_2 are the function relate to the parameters of the algorithms. For the EM algorithm the parameter is κ. For the EMD algorithm the parameters are κ and K_d. These parameters are optimized according to

$$[\kappa] = \arg\min(Err) \tag{12}$$

$$[\kappa, K_d] = \arg\min(Err) \tag{13}$$

where we have chosen $Err = \tau Err_1 + \sigma Err_2$ as a integrated error which considers the square error and absolute error together. Here τ and σ are weight of the integration. For the purpose of simplification, we take $\tau = \sigma = 1$.
.

5.1 The optimization of the EM algorithm

The parameter are found from above equation. Assume the size of 3D the image is 200^3. The original object function used to produce the simulated ultrasound data is a cube. The number of scanning 2D frames used to cut the 3D image is 100. The number of iteration in the reconstruction algorithm is taken as 1000. the optimization in Table 1 shows $\kappa = 3.6$ for the EM algorithm. It is remarkable that we have found the above optimized parameter is not sensitive to the different object function.

No	1	2	3	4	5	6	7	8
κ	0.3	0.45	0.9	1.8	2.7	3.6*	5.4	7.2
Err_1	1.12037	0.982879	0.816238	0.727079	0.703761	0.69733*	0.69941	0.706903
Err_2	2.80993	2.39905	1.90952	1.64456	1.57741	1.56387*	1.58623	1.62678
Err	3.9303	3.381929	2.725758	2.371639	2.281171	2.2612*	2.28564	2.333683

Table 1. Optimization of the parameter of κ=3.6 is found. Here Err_1 is absolute error, Err_2 is square error. Err is integrated error. The smallest value is found and marked with "*".

5.2 The optimization of the EMD algorithm

For the EMD method there are two parameters: κ, K_d. Assume the size of 3D the image is 200^3. The original object function used to produce the simulated ultrasound data is a cube. The number of scanning 2D frames used to cut the 3D image is 100. Iterative searching is used to found the smallest error Err. Table 2 shows if κ is fixed value 0.4, K_d=0.25, 0.275, 0.288 is found for smallest Err_1, Err_2 and Err. We chose all of them for the further optimization.

No	1	2	3	4	5	6	7
κ	0.4	0.4	0.4	0.4	0.4	0.4	0.4
K_d	0.2	0.225	0.25	0.275	0.288	0.3	0.325
Err_1	0.386619	0.337024	0.335985*	0.360026	0.37945	0.403598	0.473295
Err_2	1.40846	1.10868	0.973846	0.918293	0.904993*	0.912398	1.0089
Err	1.795079	1.445704	1.309831	1.278319*	1.284443	1.315996	1.482195

Table 2. κ is fixed at value 0.4, K_d is changed. The parameter K_d=0.25, 0.275, 0.288 is found at smallest error.

Table 3 shows if K_d is fixed at value 0.25, the parameter $\kappa = 0.25$ is found with smallest error Err. We have also calculated the situation in which K_d is fixed at 0.275 and 0.288, no smaller error Err was obtained. We found that the further iterative searching of parameters doesn't reduce the error Err too much. Hence the final results of iterative searching results are: $K_d = 0.25$, $\kappa = 0.25$.

No	1	2	3	4	5	6
κ	0.8	0.6	0.4	0.3	0.25*	0.2
K_d	0.25	0.25	0.25	0.25	0.25*	0.25
Err_1	0.382777	0.35265	0.335985*	0.346762	0.361572	0.410167
Err_2	1.33925	1.17057	0.973846	0.876938	0.82995*	0.867875
Err	1.722027	1.52322	1.309831	1.2237	1.191522*	1.2780

Table 3. K_d is fixed at value 0.25, κ is changed. The parameter $\kappa = 0.25$ is found at smallest error Err

The quality of the reconstructed image is controlled through the sharpness of the organ edge and the smoothness of the image. The sharpness and smoothness can be seen directly from the reconstructed image. In the following section the reconstructed images are compared.

6. THE RESULTS OF THE SIMULATION

The above optimized parameters in section 5 and simulation data in section 4 are utilized in the image reconstruction. Figure 3 shows the reconstructed images. Figure 3 (a) is the reconstruction using the EM algorithm (with cubic filter) (b) is the reconstruction using the EMD algorithm. The Number of iteration is $N = 1000$. The size of image is 200^3. We can see that image (b) has sharper edge at the boundary of cube and has smoother image inside the cube compared to the image (a). Figure 3 (c) and (d) show the corresponding profiles to Figure 3 (a) and (b).

We also try another situation, the original object is a cube however it is tilted by 3 Euler angles $\Phi = \frac{\pi}{6}$, $\Theta = \frac{\pi}{6}$, $\Psi = \frac{\pi}{6}$. The Number of iteration is $N = 1000$. The size of image is 200^3. Figure 4 (c) and (d) show the corresponding profiles to Figure 4 (a) and (b).

In the Figure 3 only a sample of simulated data is used in the comparison. It is not fare that only one of sample of simulated data is used. In next paragraph 72 different tries of the simulated data are used to test the errors defined in (11).

In different simulation samples, the planes of 2D frame images are created through a group points in the space. The points P(defined in section 4) obey the Gauss distribution. Points P are generated using pseudo random variable. Here different seeds are used in these random variables to create different samples. Figure 5 left shows the errors Err_1 corresponding to the two algorithms. Figure 5 right shows the errors Err_2 corresponding to the two algorithms.

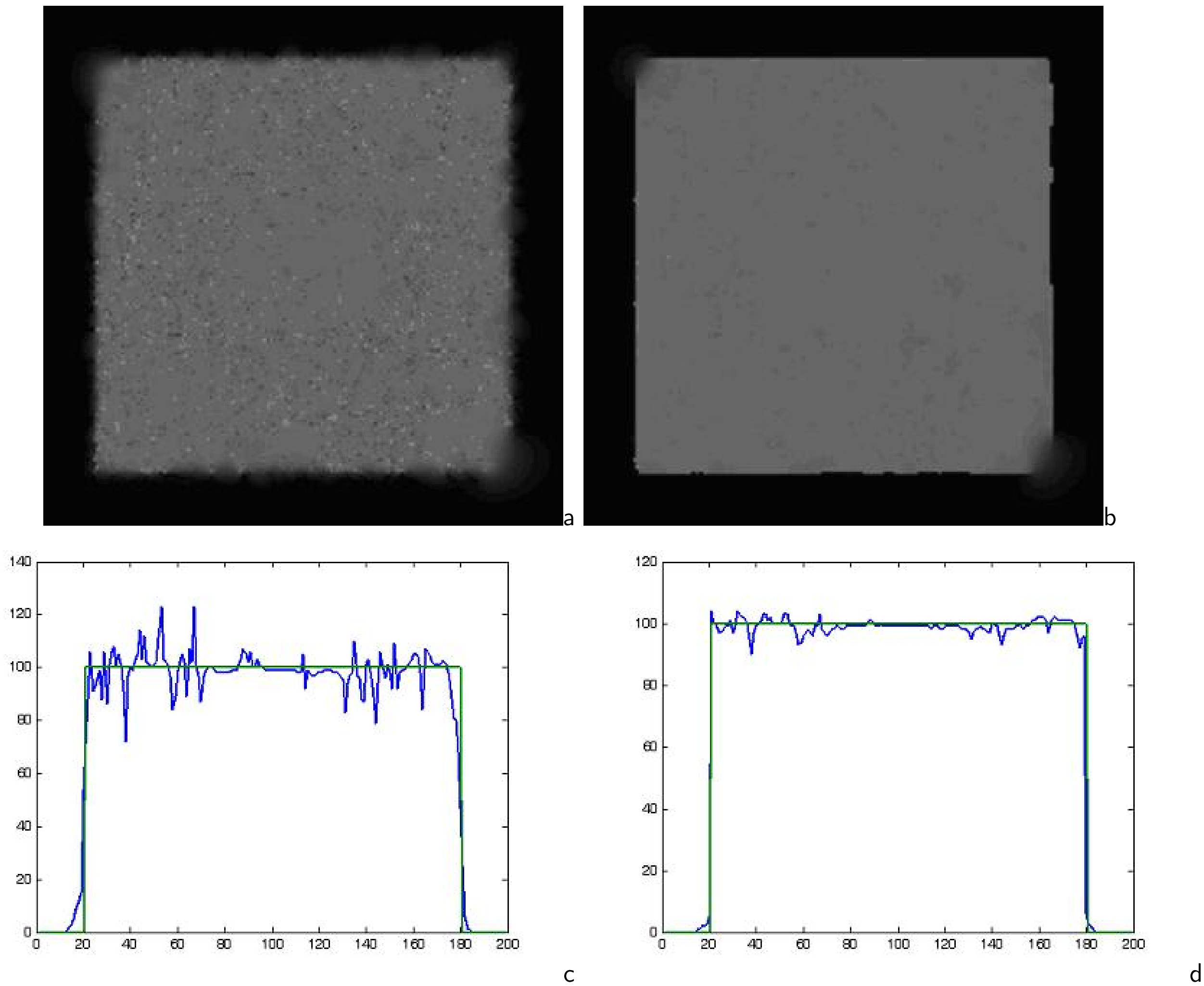

Figure 3. (a) The reconstructed image using the EM algorithm κ=3.6. (b) The reconstructed image using the EMD algorithm with K_d=0.25, $\kappa = 0.25$. Here the reconstructed 3D image is shown at the plane $k = 100$. The number of 2D frame image generated in the simulation is 100. (c) and (b) are the corresponding profiles of (a) and (b) respectively. The profile is cut at $i = 40$.

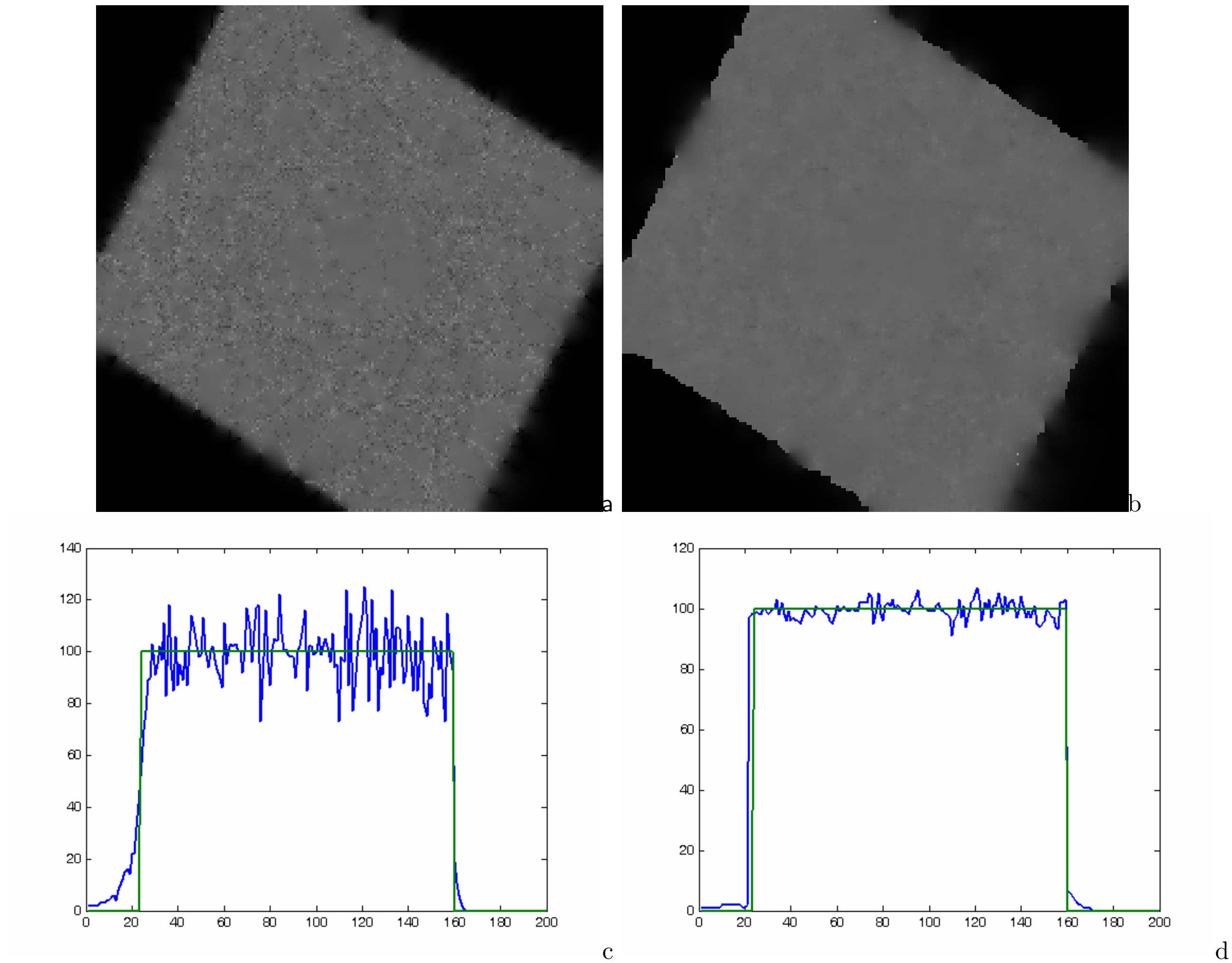

Figure 4. (a) The reconstructed image using the EM algorithm with κ=3.6. (b) The reconstructed image using the EMD algorithm with K_d=0.25, $\kappa = 0.25$. Here the reconstructed 3D image is show at the plane $k = 100$. The number of 2D frame image generated in the simulation is 100. (c) and (d) are the corresponding profiles of (a) and (b) respectively. The profile is cut at $i = 40$.

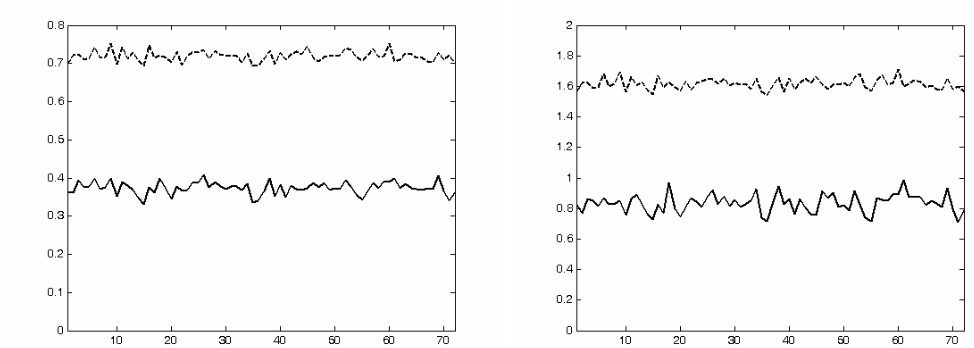

Figure 5. The errors of the reconstructed images. Left is the absolute errors Err_1. Right is the square errors Err_2. The virtual line is corresponding to the EM algorithm. The solid line is corresponding to the EMD algorithm. The number of plane of 2D frames in the simulated data is $M = 100$. The number of tries or samples is 72.

7. THE COORDINATES OF TRACKER SYSTEM AND THE TRANSFORM MATRIX

Free hand B-scan data are acquired using Eigen's tracker system, see Figure 1. This tracker has two arms. The ultrasound probe is fixed on the end of the second arm. The probe can be rotated around its axis. This system has 4 freedoms. The rotation of first arm corresponding to angle α. The rotation of second arm corresponding the angle γ. The self-rotation of the probe corresponding to ρ. The displacement of the probe corresponding to δ. The 2D Frame image are acquired in continuous mode. The rate of frames is 33 per millisecond. The coordinates x_l^{video} of grid on the frame and the ultrasound pixel value y_l are measured. There are different coordinates. The frame coordinates which are the positions of grid in the acquired 2D frames. The probe coordinates which has the origin at the top of probe and one of their axis is alone the direction of ultrasound probe axis. The tracker coordinates which describes tracker and fixed on the earth. The home coordinates which is at the place to display the 3D image. The volume coordinates X^{volume} which is scaled coordinates to suit the reconstruction voxels.

$$X^{volume} = T(\alpha, \gamma, \theta, \delta)\, X_l^{video} \tag{14}$$

Here $T(\alpha, \gamma, \theta, \delta)$ is the product of all the above mentioned 4D coordinate transforms include the rotations and the displacement. After data acquisition a set of data $\{(X_l^{video}, y_l),\, l \in [0, L)\}$ is obtained. L is the number of pixels of all acquired frames. This data is used in the reconstruction algorithms in section 2 and section 3. In the following the prostate phantom is used in a few experiments to compare the results of image reconstruction from the two algorithms. The 3D reconstructed images and the volumes after segmentation are compared. The segmentation is a half automatic algorithm, which is done with the software "3DQuatify", version 3.7.1 developed by Robarts Research Institute.

The 2D image is measured on frame coordinates. The reconstruction is done on the coordinates of first frame. The following define the coordinates:

video coordinates: The coordinates of a image pixel at the video screen. We assume $z = 0$ in this coordinates. The origin is at the low and left corner. image pixel position is at $r_{video=}(x_{video}, y_{video}, 0)$. Video coordinates can move with the tracker angle changing.

NGP (Needle geometric position) coordinates: . The origin of this coordinates is at top of needle. This coordinates is also parallel to the video coordinates. The only difference is the translational movement of the two origins. video coordinates and NGP coordinates move with change of the tracker angles.

Tracker coordinates. This coordinates is fixed to earth. The position of video frame is decided by four parameters of tracker. The rotation angle of arm A:α, the rotation angle of arm B: γ, the self-rotation angle of the needle:ρ and the displacement of the probe: δ.

Home coordinates. In the start time of the scanning $(t = 0)$ fixed the NGP coordinates to the earth. Here t is scan time. This coordinates is fixed to earth. It can not move with tracker angle changing. This lead that

${}^{NGP}T_{video}$ the 4-D matrix transform from video coordinates to NGP coordinates. ${}^{tracker}T_{NGP}$, ${}^{home}T_{tracker}$, ${}^{volume}T_{home}$ have the similar definition.

It is remarkable that,${}^{NGP}T_{video} = {}^{NGP}T_{video}(\delta)$ and ${}^{tracker}T_{NGP} = {}^{tracker}T_{NGP}(\alpha, \gamma, \rho)$

Here α, β, ρ is the rotation angles of the first arm, the second arm, and the angle of the needle self-rotation. According to the definition of the home coordinates there is:

$${}^{home}T_{tracker} \cdot {}^{tracker(t=0)}T_{NGP} = I \tag{15}$$

or

$${}^{home}T_{tracker} = [{}^{tracker(t=0)}T_{NGP}]^{-1} \tag{16}$$

The volume coordinates is the coordinates used for 3D volume reconstruction. This coordinates parallel to the home coordinates, however its origin is translational moved to the center of the frame in the starting time of the scanning $(t = 0)$.

The coordinates of reconstructed volume can be calculated from:

$$T(\alpha, \gamma, \rho, \delta) = {}^{volume}T_{home} \cdot {}^{home}T_{tracker} \cdot {}^{tracker}T_{NGP} \cdot {}^{NGP}T_{video} \tag{17}$$

In the above discussion the home coordinates is corresponding to the first frame. Actually the home position can be chosen for any frames. For our prior product there only rotational scanning, the home is chosen at first frame. In the case of freehand scan, if the scanning is almost-parallel, the home is better chosen corresponding to the middle of all the frames. Otherwise you have to scan to and fro in order to make the reconstructed image fill the most reconstructed volume.

8. RECONSTRUCTION FROM SIMULATED TRACKER SYSTEMS

In the section 5 the 2D image reconstruction from simulated ultrasound data is discussed. In that section the needle tracker coordinate system is not involved. The frame is assumed as a set of random planes. In this section the frames are simulated through the needle tracker coordinate system which has been discussed in section 7. The purpose of this section is to test the coordinate calculation of our tracker system. The scanning of our prior product Artemis system is rotational that means there is only one rotational axis. We know that if the calculation of tracker system correct for free hand scanning, it should also correct for a simplified case in which there are two different rotational axes of scanning. The rotation with two different axes are chosen in this section. In this simplified case if our tracker calculation is wrong the reconstructed image will have double images. Then If the reconstructed image do not contains double image, the calculation of our tracker is perhaps correct.

The number projections of the scanning is 50, the first 25 projections is rotated with a axis. The last 25 projections are rotated with a different axis. Two axes have different α, β values which are the rotation angles of the tracker arms. The object of the phantom is a ball inside a cube. The ball does not stay at the center of the cube. We did not add any noises to the 2D image frames. The reconstruction is done only with the EM algorithm. EMD method is not discussed here since the purpose of this method is to reduce the noise, but our simulated data doesn't have any noises.

The projection data is produced with Matlab. The reconstruction is calculated with C++. Figure 6 shows the reconstructed image. The reconstruction parameter (in section 5) is κ=4.0.

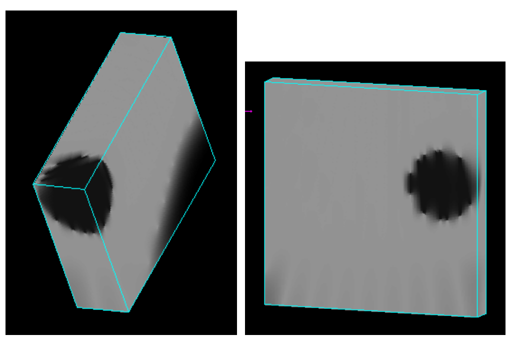

Figure 6. The reconstructed image from simulated two-axis scanning with 500 iterations, EM method is used, κ=4.0.

The correct reconstruction results in 6 shows there are no any double images, hence our tracker system has been simulated correctly in rotational scanning with two *axes*. In this section we do not directly use freehand scanning is that if tracker measurement is wrong there is multiple image instead of double image. It is difficult to correct the system with multiple image.

9. THE RECONSTRUCTION RESULTS FOR MEASURED DATA

In the strict sense, freehand scanning should includes any scanning compounds, parallel movement and self-rotation movement. Actually freehand scanning is a kind of scaning "almost parallel" see Figure 1 b). In this case, only one angle α or γ changes very fast, other parameters change slow. We call it as almost-parallel freehand scanning. In case self-rotation angle ρ change fast, other parameter change slow, we call it as almost-rotational freehand scanning.

In the section 6 we discussed how to reduce the noises of reconstructed image. In case to reconstruct a image from real measured ultrasound 2D frame, there are other important issues. Which are list as following:

(1) Accuracy of the tracker measurement $(\alpha, \gamma, \rho, \delta)$. The mechanical deformation of the tracker system during the measurement have big influence to the accuracy of the tracker.

(2) Synchronization between tracker and frame measurements. We have known that the frame image time lag (τ) is at one hundred more milliseconds. The tracker time lag is at a few nanoseconds it can be seen as 0 second. When the image is grabbed, and then measure the tracker position is too later. Tracker should measured at τ time before.

(3) Frame image calibration, before using the tracker ultrasound system the origin of the frame and the needle tracker need to be calibrated.

(4) Mechanic system calibration. The needle self-rotational axis should consist with all frames, i.e. the needle self-rotational axis should go through all frames.

If the above requirements (1) and (2) doesn't meet, severe double image artefacts are produced. If the above requirements (3) and (4) doesn't meet, severe misalignment artefacts are produced for the case of almost-rotational scanning.

9.1 Comparison of Reconstructed Images using Freehand Scanning

In this experiment we freely move the first arm of Eigen's tracker forward and backward and then move the second arm forward and backward. This means that in the scanning α is continuously changed and then γ is continuously changed. In the scanning process the other angles and displacement are not fixed. They can also have a change in a small amount. Here the 2D ultrasound frame images are acquired with Zonare ultrasound machine with depth setting 7cm. 30 frames is taken per second. The total scanning time is around 20 seconds. One slice of reconstructed images with EM algorithm and EMD algorithm are show in Figure 7. The optimized parameters in section 5 are used in both algorithms.

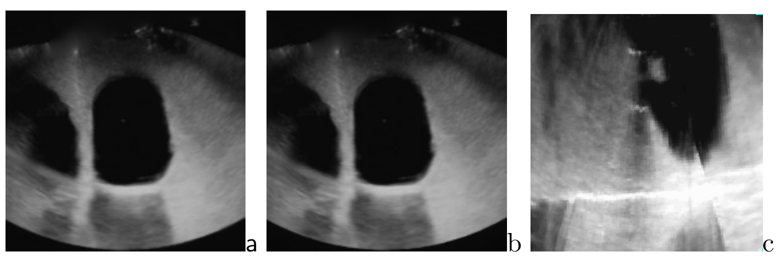

Figure 7. The reconstructed ultrasound images in freehand mode. The first arm of the tracker is moved forward and backward. The second arm of the tracker is moved forward and backward. So the scanning is almost-parallel. (a) is the reconstruction using EM method. (b) is the reconstruction using EMD algorithm. The two image is take from the lice of z=120 in [0, 199]. (c)Needle is rotated. The scanning is almost-rotational freehand scanning. The reconstruction method is EM method.

Figure 7 (a) and (b) show that EM and EMD algorithms both can get very good reconstruction results if phantom is scanned. EMD method is generalized method of EM method. if $K_d \to \infty$, EMD method $\to$EM method. Since the 2D frame image of phantom has very small noises and these noises have been further filtered by the Zonare ultrasound machine, for this data we don't expected there is clear difference between two algorithms.

Figure 7 fig (c) shows the reconstruction with almost-rotational scanning. There is recognizable misalignment artefacts. This kind artefacts happened at the connect place of start and end scanning. Misalignment artefacts are caused by the conditions (3) and (4) of last sub-section. The misalignment artefacts is also happened in our prior products Artemis System which has purely rotational scanning if the condition (3) and (4) are not met.

Since it is not very sencitve to almost-parallel scanning to the condition (3) and (4), we can more easily get better reconstruct images with almost-parallel scanning than that with almost-rotational scanning.

In most situations freehand scanning actually means almost-parallel scanning. 7 shows Engen's tracker is capable to get accurate data for freehand ultrasound scanning and 3D image reconstruction.

9.2 More reconstructed results

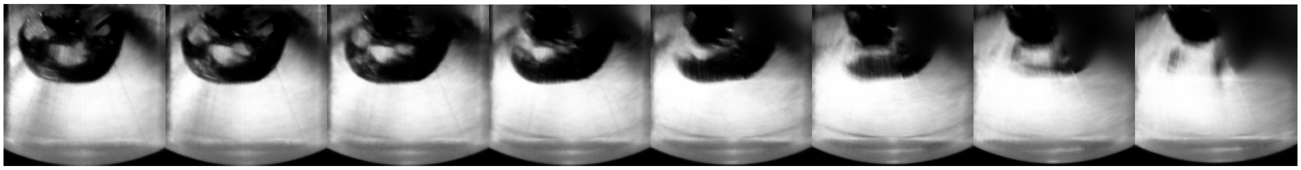

Figure 8. Reconstructed image at z =110,115,120,125,130,135,140,145. The scanning mode is almost-rotational. Reconstruction is with EM method. The parameter is $\kappa = 4.0$

In Figure 8 the scanning is almost-rotational. The scanning object is a prostate phantom with its volume as 36 cm^3. Reconstruction method is EM method with parameter $\kappa = 4.0$. The number (110), (115), (120) are slice number along z axis.

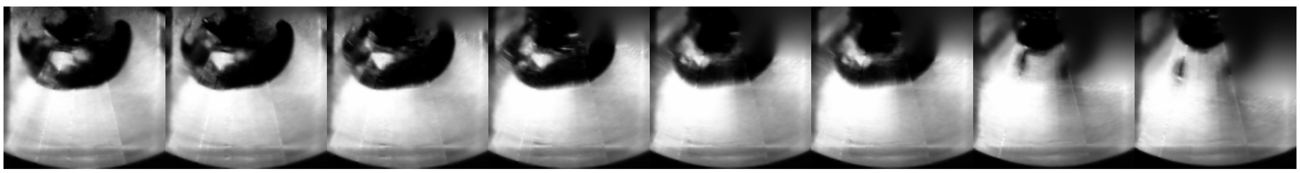

Figure 9. Reconstructed image at z =110,115,120,125,130,135,140,145. The scanning mode is almost-parallel. Reconstruction is with EM method. The parameter is $\kappa = 4.0$

In Figure 9, the scanning is almost-parallel. The phantom, reconstruction method and reconstruction parameter used are the same as above. In this time we have carefully calibrated the system, so that the misalignment artifacts does not severe.

Figure 8 and 9 show that almost-parallel and almost-rotational freehand scanning both can get very good image quality.

9.3 Calculated Volumes

Table 4 shows the comparison of volumes obtained from different algorithms and different modes of scanning. A prostate phantom is scanned 6 times in purely rotational and almost-parellel freehand modes. The first line shows the index number of samples. The results in the second line are obtained through rotationally scanning the same phantom as above. The scanning has only one freedom which is the angle of ρ. The frame image is taken in every 0.9 degree in a 180 degree rotation. The scanning time is around 15 seconds. The 3D image and its volume is reconstructed and calculated through Eigen's software of Artemis. For the third line and fourth line of the Table 4, EM and EMD algorithms are used to calculated the reconstructed image. The segmentation and the volume calculation are done with "3D Quantify" software. The last column is the average of the volume in different lines.

Samples	1	2	3	4	5	6	Av
Artemis	31.08	30.43	31.73	30.9	29.8	32.52	31.07
EM	30.56	32.56	33.65	29.41	29.56	28.93	30.56
EMD	30.12	31.72	32.37	29.42	29.05	28.95	30.27

Table 4. Comparison of volumes between two algorithms, the unit of volume is cm^3

This table shows that the calculated volumes using EM and EMD algorithms with freehand scanning are very close to the volume calculated and reconstructed by using Eigen's Artemis system with rotationally scanning.

10. CONCLUSIONS

The EM algorithm is reviewed. EM algorithm has combined the interpolation and filter together. The filter of EM method is a cubic average filter. The EMD algorithm is introduced in this article. In EMD method the cubic average filter is replaced by a diffusion filter. The two algorithms EM and EMD are compared. 3D Simulated freehand B-scan data is produced. The simulation result with strong noises shows EMD algorithm can achieve better reconstruction according to the absolute errors and square errors comparing to the EM algorithm. As we understand that EMD algorithm has an additional parameter K_d which offers a better optimization compared to EM algorithm, it achieves better balance between the smoothness of the image and the sharpness in the boundary of the human organ.

The results with measured data from prostate phantom shows both EM and EMD algorithms offer good reconstruction results. Since the phantom has very small noises, the difference of two algorithms are not distinct. We expected that EMD can achieves better reconstruction results if noisier data like human organs are scanned instead of phantoms. This kind of comparison will be done in the future. We have used GPU CUDA technique to speed the iterative diffusion calculation of the EMD algorithm.

In case with measured data there are other important issues, synchronization between the tracker and 2D ultrasound image frame, accuracy of measurement, calibration of the system and mechanical deformation of tracker in scanning process. The results show Eigen's tracker system can overcome all this kind of difficulties and produces good freehand scanning ultrasound data which can be further used to reconstruct high-quality 3D image.

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