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Fast Ultrasound to Ultrasound Auto-Registration for Interventional Cardiology

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***Abstract*—Many interventional procedures rely on 3D trans-esophageal echocardiography (TEE) imaging to guide the interventionist during instrument/device manipulation inside the heart. Manual registration of ultrasound to ultrasound or other modalities is commercially available. However, auto-registration is highly desirable, to avoid manual re-labelling and to estimate probe motion for motion compensation during surgery. There are a number of published auto-registration methods, but they are not suitable for interventional setups as they are not fast enough.In this work we present a close to real-time 3D ultrasound volume to 3D ultrasound volume auto-registration method that is implemented on the GPU. The auto-registration method performed on the clinical data was successful and good visual alignment was observed in 7 out of 8 cases. Decomposing a volume into its polynomial coefficients took 73ms and was performed once for each pyramid level. One iteration took on an average of 53ms. The proposed registration method is fast enough to allow real-time usage on the ultrasound scanner. When the method converges to a good result it is close to the expert's manual registration with discrepancies of *1.14 ± 1.27 mm* for translation in the X direction, *2.3 ± 1.79 mm* for translation in the Y direction, *2.08 ± 2.17 mm* for translation in the Z direction, *$0.7 \pm 0.7^\circ$* for rotation about the Z axis, *$1.85 \pm 2.55^\circ$* for rotation about the Y axis, and *$2.63 \pm 1.82^\circ$* for rotation about the X axis.**

***Index Terms*—Interventional ultrasound, ultrasound registration, 4D ultrasound volume, TEE, realtime registration**

I. Introduction

Heart valve intervention is the second most frequent cardiac intervention after angioplasty. Traditionally, open-heart surgery has been the standard to treat insufficient hearth valves by repair or replacement and correct other heart defects. However, open-heart surgery is associated with complications and mortality both in the short and long term. With the development of new technologies, minimally invasive heart surgery and catheter-based percutaneous intervention has proved to be safer compared to open-heart surgery for many procedures. These procedures are less painful for the patients and require shorter recovery time, with less need for hospitalization and rehabilitation. With the success of minimally invasive heart operations, the trend is nowadays moving from open-heart surgery to minimally invasive surgery. Hence new heart devices and new interventional ultrasound systems were developed for the minimally invasive surgery. The interventional ultrasound system is used for guiding the operations. It becomes an essential tool in the interventional cardiologists daily toolbox. There are many interventional procedures that rely on 3D trans-esophageal echocardiography (TEE) imaging to guide the interventionist during instrument/device manipulation inside the heart. In particular, percutaneous mitral valve procedures, who are performed in a rapidly growing number of patients, rely on 3D TEE. TEE provides higher quality images than the trans-thoracic echocardiography due to probe location being in the esophagus that is directly behind the heart. Using novel interventional ultrasound system such as GE's ultrasound system, it is possible to get real-time 3D/4D high quality ultrasound volumes for the TEE.

It would be very beneficial to have successful real-time registration of the 3D ultrasound volume to 3D ultrasound volume or ultrasound to other modalities. There are many different medical applications for it such as motion estimation, motion compensation, or left ventricle motion tracking during surgery. Manual registration of ultrasound to ultrasound or to other modalities is commercially available. However, auto-registration is highly desirable, to avoid manual re-labelling and to estimate probe motion during surgery. There are several published auto-registration methods, but they are not suitable for interventional setups as they are not fast enough [1]–[3], [5], [6]. In this work we present a close to real-time 3D ultrasound volume to 3D ultrasound volume auto-registration method that is implemented on the graphics processing unit (GPU).

II. Materials and Methods

A. Patient data

For the evaluation of the proposed method, 3D TEE B-mode images were obtained by cardiologists with echocardiographic expertise from 8 patients using GE Vivid E95 and E9 systems with a 6VT-D probe (GE Vingmed Ultrasound, Horten, Norway). All patients were examined in the clinic for diagnostic purposes. For each patient at least one complete cardiac cycle was captured. The frame rate of the recordings was in the range of 6 to 27 frames per second. No selection of the patients was performed, and the field of view was not specially adjusted, therefore the image quality is representative of that encountered in the clinic.

For qualitative comparison purposes an expert echocardiographer was asked to align the volumes manually. Additionally, the location of 5 landmark points indicated by the expert as well as the final registration matrix for the manual alignment were saved.

B. Data preprocessing

All samples were anonymized before analysis, and to facilitate processing, the images were converted from the proprietary DICOM format to 3D volumes by applying a polar-Cartesian transform on the raw B-mode lines. All datasets were resampled to isotropic volumes with a pixel size of 0.7 *mm*. The isotropy is required in order to improve the computational speed of the Farneback decomposition as detailed below.

To reduce the amount of speckle present in the volumes a *5x5x5* median filter was applied, followed by a standard Gaussian filter with an extent of 3 in each dimension and a standard deviation of 1.5. Finally, a custom 1D transfer function was applied in order to attenuate the gray values inside the heart cavity.

C. Farneback decomposition

To estimate the displacement between a reference image and a moving image, both images needed can be represented by polynomials as follows:

$$f(x) = x^T A x + b^T x + c \tag{1}$$

with A is a 3x3 matrix containing second order coefficients, b is a 3x1 vector for the linear components, and c is a constant value [1], [2]. Our approach is similar to the work of Danudibroto et al. [3], however we also tested local affine deformation constraints and weighting of the polynomial coefficients in order to improve the alignment results.

For computational efficiency the Farneback decomposition was implemented on the GPU. In order to compute the polynomial coefficients a convolution with kernels of a predefined extent needs to be implemented. Assuming isotropic voxel size the convolution can be simplified and implemented as one dimensional separable filters. Three passes are required in order to compute all the coefficients of the decomposition. The voxel wise convolutions for the *x, y* and *z* directions are computed in parallel on the GPU, as the value of a voxel are independent of the neighboring ones. However, a temporal synchronization when moving to the next direction needs to be enforced in order to ensure consistent results.

D. Similarity metric and transformation estimation

At each voxel in the volume the displacement $d(x,y,z)$ between the reference and the moving image can be estimated by enforcing:

$$f_r(x) = f_m(x - d) \tag{2}$$

As detailed by Farneback [1], [2], finding d in equation (2) can be done by:

$$d = \frac{1}{2} A_f^{-1}(x)(b_f(x) - b_m(x)) \tag{3}$$

However, when considering two distinct volumes their decompositions cannot be expected to be identical. Additionally, we would like to introduce an *a*-priori local deformation, to allow for an iterative approach. Therefore, the following values for $A(x)$ and $\Delta B(x)$ are used instead:

$$A(x) = \frac{A_r(x) + A_m(\tilde{x})}{2} \tag{4}$$

$$\Delta b(x) = \frac{1}{2}(b_r(x) - b_m(\tilde{x})) + A(x)\tilde{d}(x) \tag{5}$$

$$\tilde{x} = x + \tilde{d}(x) \tag{6}$$

Adding a locally affine constraint on the deformation is also possible by imposing the following constraints of the deformation:

$$d_x(x, y, z) = a_1 x + a_2 y + a_3 z + a_{10} \tag{7}$$

$$d_y(x, y, z) = a_4 x + a_5 y + a_6 z + a_{11} \tag{8}$$

$$d_z(x, y, z) = a_7 x + a_8 y + a_9 z + a_{12} \tag{9}$$

Which in matrix form becomes:

$$d(x) = S(x)p \tag{10}$$

with

$$S(x) = \begin{pmatrix} x & y & z & 0 & 0 & 0 & 0 & 0 & 0 & 1 & 0 & 0 \\ 0 & 0 & 0 & x & y & z & 0 & 0 & 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & x & y & z & 0 & 0 & 1 \end{pmatrix} \tag{11}$$

and

$$p = (a_1\ a_2\ ...\ a_{12})^T \tag{12}$$

Furthermore, when considering a given neighborhood, the contribution of voxels with similar polynomial decompositions can be enhanced by adding a Frobenius norm of the form:

$$\gamma(x) = e^{-\frac{\alpha||A_f(x)-A_m(\tilde{x})||_F^2}{||A_f(x)||_F^2+||A_m(\tilde{x})||_F^2}} \tag{13}$$

This is quite useful for ultrasound images as dropouts and other artifacts can appear between two acquisitions.

Finally, it was observed that for ultrasound images an approximation based only of second order terms gave better results, therefore original equations proposed by Farneback were modified as follows:

$$d = \left(\sum A(x)^T A(x)\right)^{-1} \sum A(x)^T \Delta b(x) \tag{14}$$

for the non-constrained version, whereas the locally affine deformation is computed as:

$$G = \sum \gamma(x) S(x)^T A(x)^T A(x) S(x) \tag{15}$$

$$h = \sum \gamma(x) S(x) A(x)^T \Delta b(x) \tag{16}$$

$$p = G^{-1} h \tag{17}$$

Once p is estimated, the local deformation vector d can be then computed using equation (10)

All the sums required by (14) and (17) are implemented on the GPU. Solving them over the entire image will give the global translation or affine deformation between the two volumes. However, since in our case (intra-patient alignment with a small temporal delay) a rigid deformation between the two volumes is desirable, we have chosen to estimate d as given by (14) or (17) locally over a given neighborhood and then use Procrustes analysis to find a rigid deformation between the two point sets in the fixed and moving images. The singular value decomposition method presented in Eggert et al. [4] was preferred as it can handle measurement noise robustly.

To capture larger deformations, a pyramid approach was adopted. The original reference and moving images are stored as textures on the GPU, as a result they can be easily subsampled. By placing the sampling points equidistantly to existing vertex locations in the high-resolution volume, an additional smoothing can be also achieved with no computation overhead.

E. Implementation

The algorithm was implemented on a DELL Precision 5510 laptop with Intel(R) Core(TM) i7-6820HQ CPU, 16 GB RAM, Intel(R) HD Graphics 530 and Nvidia Quadro M1000M graphic cards. The Nvidia Quadro M1000M graphic card included 512 CUDA cores, 32 TMUs and 16 ROPs.

III. Results

For the figures mentioned in the following sections, the moving 3D ultrasound volume is presented in copper color and the reference 3D ultrasound volume is presented in silver color. The moving ultrasound images were placed on top of the reference images. There are four sub-figures in each of the figures. The top left, top right, and bottom left sub-figures were 3 different slice views of the same volume. The top left sub-figure is the short axis view of the heart. The top right and bottom left sub-figures are the long axis view of the heart. The length of the long axis view is 15 *cm*. The open angle of the axis view is 60°. The bottom right sub-figure is the EGC recording with the first and second red points corresponding to the time stamps when the reference and the moving 3D ultrasound volumes were obtained, respectively. Assume that the X-axis is in the left to right direction, Y-axis is in the top to bottom direction, and the Z-axis is in the direction towards the image. The sub-figure in the top right corner of each figure is in the YZ plane, the sub-figure in the bottom left corner is in the YX plane and the top left sub-figure is in the ZX plane.

The manual registration was done in two steps: alignment based on placing the landmarks on the reference and the moving 3D ultrasound volumes; manual alignment to fine tune the results. The manual registration results were used as ground truth for the auto registration method in this work. The auto-registration method performed on the clinical data was successful and good visual alignment was observed in 7 out of 8 cases. Using the DELL laptop mentioned in the section above, decomposing a volume into its polynomial coefficients took 73 *ms* and was performed once for each pyramid level. One iteration took on an average of 53 *ms*.

We obtained transformation matrices for all the data using manual as well as auto registration methods. Euler angles were also extracted for each transformation matrix. The discrepancies for the translation in the X, Y, and Z directions were calculated based on the differences between the 4$^{\text{th}}$ columns of the transformation matrices of the manual and the auto registration. The discrepancies for the rotation about the Z, Y, and X axes were calculated based on the differences among the Euler angles of the results from the manual and the auto registration. The discrepancies are $1.14 \pm 1.27mm$ for translation in the X direction, $2.3 \pm 1.79mm$ for translation in the Y direction, $2.08 \pm 2.17mm$ for translation in the Z direction, $0.7 \pm 0.7°$ for rotation about the Z axis, $1.85 \pm 2.55°$ for rotation about the Y axis, and $2.63 \pm 1.82°$ for rotation about the X axis.

IV. Discussion

One sample case was chosen to demonstrate the results of the method. Fig.1 presents the data from patient 8 without any alignment, Fig.2 presents the result of the manual registration from the expert, and Fig.3 presents the result of the auto registration method. When we look at the sub-figure on the top right in Fig.2 and Fig.3, there is a discrepancy of about 1 mm for translation in the Z direction, also 1 mm in the Y direction between the manual alignment and the auto registration method. When we look at the sub-figure on the bottom left in the figures, the discrepancy is still 1 mm in the Y-direction and is also about 1 mm in the X direction between the manual alignment and the auto registration method. Finally, when we look at the top left figures, there is a discrepancy of about 7 degree for the rotation about the Y direction between the manual alignment and the auto registration.

The overall result of the auto registration is good, and it is close to the expert's manual registration. The estimation of the algorithm is good for translation and there is room to improve for both the translation and rotation estimations to get better registration results. The implementation is fast and can be used for close to real-time registration on an ultrasound scanner. If a more powerful and economical GPU is used, the processing speed will be faster and it also improves the accuracy of the auto-registration. It is also noted that the data need to have pre-processing in order to get good results from the auto registration method.

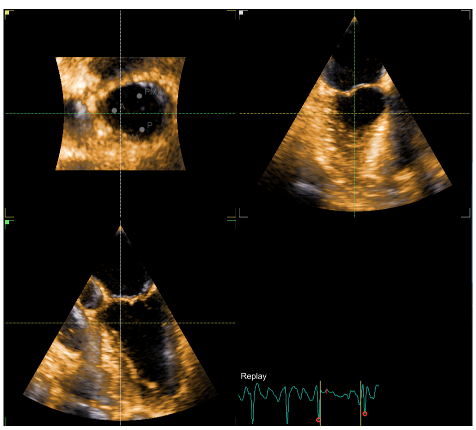

Fig. 1. Data of patient number 8 without alignment

V. Conclusion

In this work, a new auto-registration method was performed on the clinical data. The registration was successful and good visual alignment was observed. The proposed registration method is fast enough to allow real-time usage on an ultrasound scanner.

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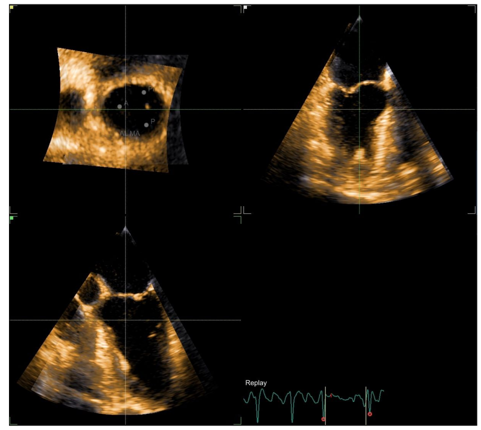

Fig. 2. Manual registration result of patient number 8

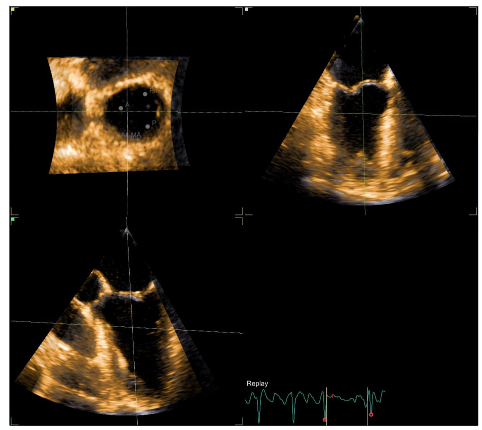

Fig. 3. Auto registration result of patient number 8

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