Synthesis of Panoramic Views of Peripheral Vessels from Sequences of Echographic Images

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Abstract

Echographic imaging plays a decisive role in diagnostics. However, the narrow field of view which is obtainable with an ultrasound probe can be a serious drawback. Techniques are needed to extend the field of view by reconstructing panoramic images from image sequences. In this paper the results obtained with the algorithm developed by H. Bulthof et al. [1] are illustrated. The algorithm was used to estimate the motion of the probe between every pair of consecutive frames of echographic sequences. Subsequently, the gray level map of every single frame of the sequence was translated according to the estimated motion and added to the gray level maps of the previous frames with a mean operation. Three kinds of test sequences were used to evaluate the algorithm performances: simulated sequences, sequences of a calibration phantom and sequences of brachial arteries.

1. Introduction

An ultrasound video signal is a rich source of information which offers two main advantages with respect to still images; the opportunity of obtaining a continuously varying set of views of a scene and the opportunity of providing information on dynamic phenomena. In particular, the acquisition of a continuous set of views of a scene allows the reconstruction of panoramic images (mosaics). In this paper, a procedure which provides panoramic images of peripheral vessels - starting from a sequence of B-mode images - is shown, where the images are acquired by translating a linear array probe with a free-hand movement on the patient's body. According to Van den Elsen, Pol and Viergever [2], the methods for image matching can be divided into three main classes as follows: feature based methods where the inter-frame displacements are calculated starting from a small set of image features (landmarks) selected by the operator [3]; intensity based methods where the inter-frame displacements are computed by measuring the similarity of two images [4][5] and more complex systems where the inter-frame displacements are obtained with position sensors and mechanical arms [6].

Our paper is based on an intensity based method. We made this choice in order to avoid operator-dependence. Moreover, neither position sensors or articulated mechanical arms were used so as not to increase the complexity of the system, even though lighting changes and inadequate acoustic windows can affect the accuracy of the panoramic reconstruction. Since consecutive images within a video sequence usually have a large overlapping area, a spatial correlation algorithm was used to estimate the motion of the probe according to the following hypotheses: the movement is mostly unidirectional, two consecutive frames are strongly overlapped, the scene is basically static, the movement of the probe is slow and the panoramic image can be projected onto a main plane without introducing deformations.

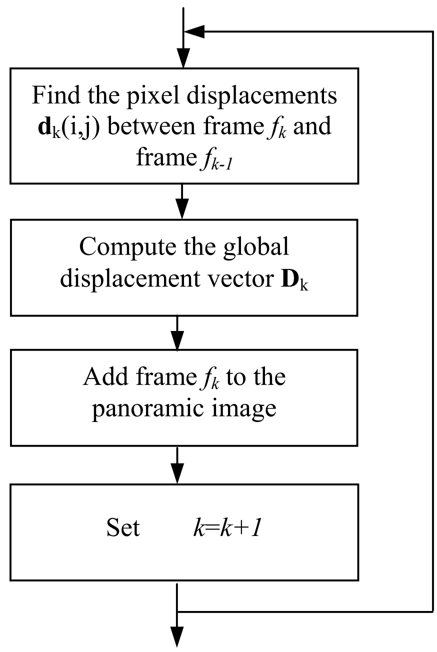

Figure 1: Procedure overview

Fig.1 shows how the algorithm can be split into the following three steps:

1. estimation of the displacements $\mathbf{d}_k(i,j)$ of every pixel of frame f_k of the sequence with respect to the corresponding pixels of frame f_{k-1},
2. estimation of the global motion $\mathbf{D}_k$ of frame f_k with respect to frame f_{k-1},
3. alignment of frame f_k with the previous frames.

These steps are repeated for each pair of frames of the sequence.

2. Motion estimation

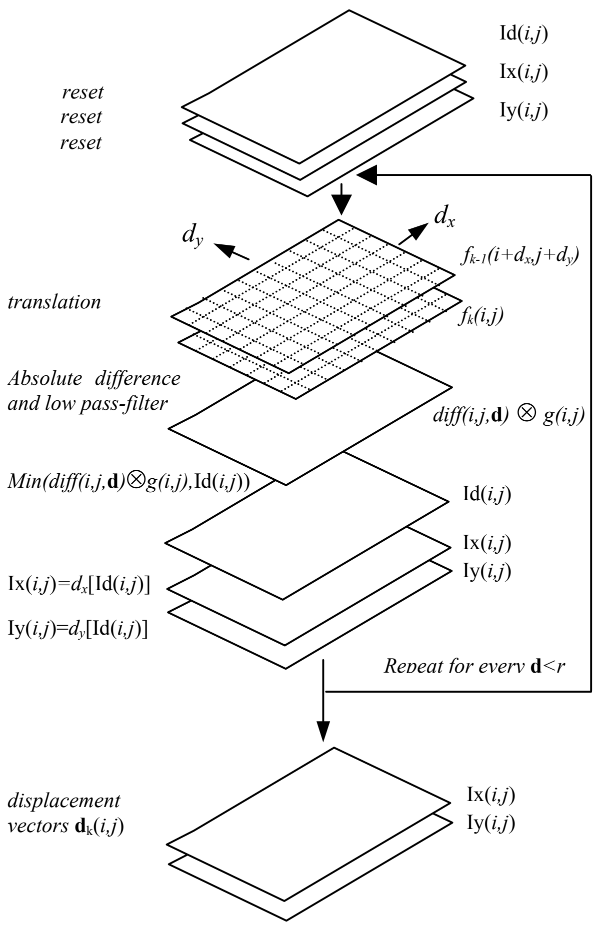

Figure 2: The figure shows a schematic representation of the procedure which we used to estimate the displacements between the pixels of two consecutive frames.

In order to estimate the motion, we used the algorithm which was developed by H. Bulthof et al. [1]. We used this algorithm since it appears to be a good choice for the implementation on a custom hardware and to obtain real-time performances. The algorithm relies on the assumption that optical flow is locally due to a pure translation. The first step is matching the local features of two consecutive frames f_{k-1} and f_k, where the considered features can be gray level intensities as well as edge locations or the output of any image filtering process. Three matrixes are used at this stage: the Id matrix where the minimum values of the absolute differences computed between the pixels of frames f_{k-1} and f_k are stored, the Ix and Iy matrixes where the displacement components which correspond to the values of Id are stored.

When the frame f_k is considered, the absolute difference $diff(i,j)=|f_k(i,j)-f_{k-1}(i,j)|$ is directly stored in Id(i,j) and every element of the matrixes Ix and Iy is set to zero. Then, the absolute difference $diff(i,j,\mathbf{d})=|f_k(i,j)-f_{k-i}(i+d_x,j+d_y)|$ is computed at every pixel of frame f_k where vector $\mathbf{d}=(d_x,d_y)$ is a two-dimensional vector whose magnitude is less than r (maximum estimated displacement). For every pixel (i,j) and for every value of $\mathbf{d}$ the absolute difference $diff(i,j,\mathbf{d})$ is compared with the corresponding value stored in Id. If the value $diff(i,j,\mathbf{d})$ is less than Id(i,j) then the element Id(i,j) is substituted with the value $diff(i,j,\mathbf{d})$ and the elements Ix(i,j) and Iy(i,j) are substituted with the corresponding displacement components d_x and d_y.

In order to cope with noise, the absolute difference map $diff(i,j,0)$ and the absolute difference maps $diff(i,j,\mathbf{d})$ are convolved with a normalized discrete box function $g(i,j)$ before being stored in Id and before the comparison process with Id, respectively. Therefore, the procedure finds, for each pixel, the displacement vector $\mathbf{d}_k(i,j)$ which corresponds to the displacement that provides the minimum value of the weighted mean difference $|(f_k(i,j)-f_{k-1}(i+d_x,j+d_y))|\otimes g(i,j)$. That is, the procedure finds the displacement vectors which minimise the absolute differences between the local distributions of the features of frame f_k and frame f_{k-1}.

As the vascular structure is supposed to be static it is clear that its apparent motion is equal to the probe displacement. The vector $\mathbf{D}_k$ which represents the global motion between frame f_{k-1} and f_k is computed as:

$$\mathbf{D}_k = \frac{1}{N}\sum\sum\nolimits_{i,j}\mathbf{d}_k(i,j)$$

In the presence of outliers (a few vectors $\mathbf{d}_k(i,j)$ which are very different from the mean vector) a poor reconstruction of the panoramic image can be obtained with the mean operator. However, the maximum displacement of every pixel is limited to a small neighborhood and consequently the presence of outliers cannot badly affect the estimation of the probe motion. The mean was chosen at this stage since it is the simplest operator, even though other operators, such as the median or other procedures of statistical analysis, appear to be more appropriate. Once every frame is aligned with the previous one they can be integrated with a mean operation to reconstruct the panoramic image.

3. System architecture

The system was developed on a Personal Computer. Once a video sequence is directly acquired from the ultrasound system, the PC analyses the frames and subsequently reconstructs and shows the panoramic image. If a critical error of the reconstructed panoramic image occurs, due to an accidental movement of the operator or to an incorrect choice of the parameters of the ultrasound system, the user can repeat the exam until a panoramic view with good quality is obtained. At the end of the exam the ultrasound sequence and the panoramic image are stored on a local digital archive.

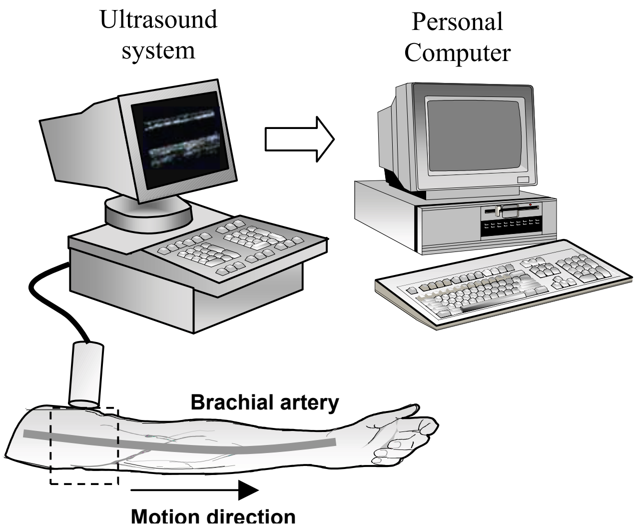

Figure 3: System architecture.

4. Evaluation of the procedure

In order to evaluate the performances of the procedure, three kinds of test sequences were used.

A sequence was synthesized with Field II. Field II is a software which has been developed at the Technical University of Denmark to simulate different fields of ultrasound transducers and different ultrasound imaging systems using linear acoustics. The software uses the Tupholme-Stepanishen method to compute pulsed ultrasound fields. Field II can compute the emitted echo fields of many different transducers simulating both pulsed and continuous wave systems. According to the authors, any kind of linear imaging can be simulated, as well as realistic images of human tissue. The image sequences were obtained by simulating the acquisition of B-scan images with a pure translation of a linear array probe. A synthetic image of a longitudinal section of a vessel with 100.000 superimposed scattering points was used as the starting image. From the original FIELD II sequence, we have obtained two more sequences by adding uncorrelated Gaussian noise with standard deviations *10* and *20* i.u..

Two test sequences were then acquired with a standard ultrasound system on a calibration phantom by using two different velocities of the probe during the acquisition phase. In this case, the probe motion was a pure translation and was achieved by using an ad-hoc mechanical support equipped with a never-ending screw.

The procedure was also applied on in-vivo images of the brachial artery of five patients which were acquired with a free-hand probe motion. In order to keep the quality of the image constant the scan correlation option of the US equipment was disabled. Sequences of 300 frames were processed by our system.

Since the displacement vectors of the probe were known in the case of Field II sequences, we were able to compute the mean percentage error on the two components d_x and d_y of the displacements $\mathbf{d}_k(i,j)$ computed at every pixel of every frame.

When using the image sequences acquired on the calibration phantom and those of the brachial arteries we compared the sizes of the reconstructed panoramic images with the distances which were covered by the probe.

5. Experimental results

The performances of the reconstruction algorithm when using Field II sequences with additional Gaussian noise are shown in Tab. 1.

σ_n	*A*	*mean_err_x (%)*	*mean_err_y (%)*
0	11	0,6	0,0
0	31	0,0	0,0
0	51	0,0	0,0
10	11	64,2	3,34
10	31	23,5	0,1
10	51	6,1	0,0
20	11	105,9	20,4
20	31	62,6	0,8
20	51	49,8	0,1

Tab. 1: The table show how the mean percentage errors vary when varying the noise standard deviation σ_n and the aperture *A* of the box function *g(i,j)*.

A	11	21	31	41	51	71
size_err (%)	5,2	3,9	3,8	3,8	3,2	2,4

Tab. 2: The table show how the percentage errors vary when varying the aperture *A* of the box function *g(i,j)*.

The mean percentage errors on the two components d_x and d_y of the displacements $\mathbf{d}_k(i,j)$ decrease when σ_n decreases and when A increases. The performances of the reconstruction algorithm, when sequences of the calibration phantom were used, are shown in Tab. 2. The percentage errors on the size of the reconstructed panoramic images do not change a lot when the aperture A varies. The size of the phantom was 115 mm.

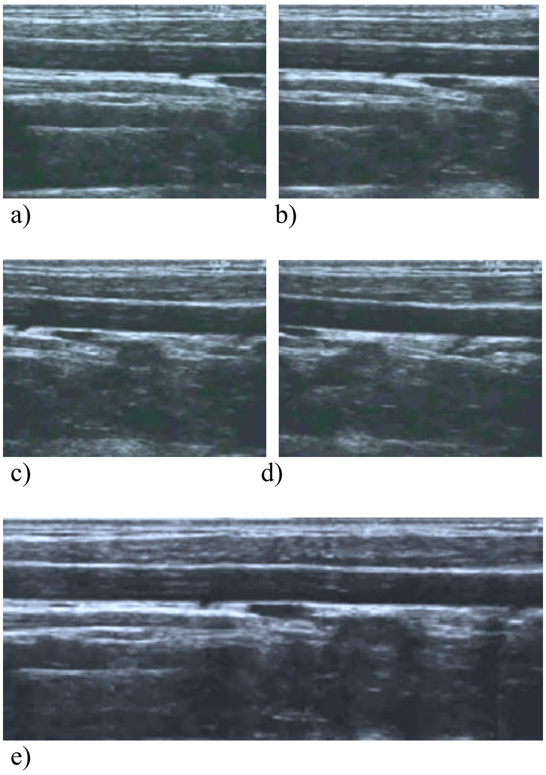

Figure 4: Example of a panoramic reconstruction. Images from a) to d) are still frames. Image e) is the reconstructed panoramic image.

When using image sequences of brachial arteries the percentage errors were inferior to 5% on distances of 8-10 cm. In Fig.4 the panoramic reconstruction of a brachial artery which was generated from 64 frames is shown. The image sequence was acquired with an ESAOTE ultrasound system and a 10 Mhz probe. The sequence was processed on a Pentium III 1200 Mhz with 256 Mb of RAM and the reconstruction process took 10 sec.

6. Conclusions

The panoramic image is an interesting tool and can play an important role in assisting physicians when they examine lengthened structures such as the vessels of the arm and/or the leg. The enlargement of the field of view can also be useful before or during a surgical intervention. A technique to generate panoramic images of peripheral vessels from echographic image sequences is proposed. An algorithm which was proposed previously by H. Bulthof et al. is used to compute the local motion. We have analyzed the algorithm in order to optimize the latter on a specific application.

It is not essential to estimate the local displacement of every single pixel of every frame of the sequences to compute the global motion of the frame when the latter is a pure translation. However, this algorithm was preferred to an algorithm which directly computes the global motion since we are planning to exploit the information on the local displacements in the near future. Our aim is to discharge those local motion artifacts which usually affect the panoramic reconstruction, to detect local deformations of the structures under observation and to adopt a more complex and realistic motion model. For example, probe rotations are needed when studying the thyroid and in this case a more general procedure is necessary to reconstruct a panoramic view of this organ.

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