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Automatic Image-based Gating for 4D Ultrasound

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***Abstract*—Four-dimensional (4D) ultrasound can be a powerful tool in adult and fetal echocardiography, enabling physicians to visualize planes that are not easily accessible using 2D ultrasound. However, despite the advances in the transducer technology, gating is still required to avoid motion artifacts during acquisition using mechanical probes or even 2D matrix array transducers in case of extended volume or color Doppler acquisitions. In fetal echocardiography, where an ECG signal is not readily available, image gating is used, which can both be computationally expensive and/or require user interaction. We present a new image gating technique that has both low computational requirements and is fully automatic. It is not only suitable for real-time gating of 4D ultrasound data sets during acquisition, but also suits well for lower-cost systems where computing resources are at a premium.**

I. Introduction

Three-dimensional (3D) ultrasound has long remained a static imaging modality due to the technical challenges involved in the acquisition, processing and display of volumetric data sets in a quick and streamlined manner. This has restricted the use of 3D ultrasound to the visualization of static structures and, if the acquisition is gated, to the snapshots of moving structures (e.g., heart and vessel walls) only at a single point in the cardiac cycle. However, dynamic information from 2D ultrasound image sequences has long been utilized by physicians. One example is the detection and quantification of regional heart wall motion abnormalities during systole, which is indicative of ischemic heart disease [1].

Recent advances in ultrasound transducers as well as increases in the available computing power have made possible the acquisition, processing and display of ultrasound volumes dynamically, adding time as a fourth dimension to 3D ultrasound (i.e., four-dimensional (4D) ultrasound). However, volume rates leading to true "live" 4D echocardiography so far have only been possible with the use of 2D matrix array transducers [2], albeit with a limited field of view (typically less than a full adult left ventricle) and only in B-mode imaging. For extended B-mode volume or color-flow acquisitions, ECG gating is still needed and utilized [3]. In these acquisitions, data acquired over several cardiac cycles are combined together to visualize a single cardiac cycle using the ECG signal as a trigger source.

Image gating techniques have been employed in applications where an ECG signal is not readily available, e.g., fetal echocardiography, an area where 3D and 4D ultrasound is gaining a considerable clinical interest [4]. A simple yet effective means of image gating based on B-mode image information has been proposed by Treece *et al.* [5]. Although computationally efficient, the need for user interaction and the sensitivity of the algorithm's performance to user's parameter selection makes this algorithm unsuitable for clinical 4D ultrasound, where a continuous set of volumes need to be gated and reconstructed in real time or near real time. In this paper, we propose an enhancement that obviates the need for user interaction while still maintaining the effectiveness and simplicity of their algorithm.

Similar to Treece *et al.*'s algorithm, our algorithm relies on the generation of a temporal signal containing cardiac information, from which the cardiac phase is extracted. Treece *et al.* generate this signal by defining regions of interests (ROI) for each B-scan. These ROIs are selected manually for a number of scan planes, and the ROIs for the remaining scan planes are then interpolated from the user-defined ROIs. The user then selects an appropriate threshold for the data set, and the number of pixels falling within the threshold is computed to generate the signal containing the cardiac information. In our approach, we generate this signal by computing image centroids, details of which we discuss in the following section.

II. Methodology

A. Centroid Algorithm

When a number of structures within a 2D plane move with respect to each other, the centroid of the plane (i.e., center of mass) also translates within the same plane. We adopt this principle to determine the periodicities in image sequences. We assume that each pixel within a B-scan has a mass proportional to its intensity, and we compute the centroids according to (1) where $f(x,y,t)$ denotes the B-mode intensities of pixels at locations (x, y) in a B-mode frame acquired at time t:

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$$centroid_x(t) = \frac{\sum_x \sum_y f(x,y,t) \cdot x}{\sum_x \sum_y f(x,y,t)} \qquad (1)$$

$$centroid_y(t) = \frac{\sum_x \sum_y f(x,y,t) \cdot y}{\sum_x \sum_y f(x,y,t)}$$

Both the $centroid_x(t)$ and $centroid_y(t)$ are 1D temporal signals showing the x and y components of the instantaneous positions of the centroids. When the structures (e.g., heart valves) within a B-scan move periodically, the centroids will also translate in a periodic manner, adding a periodic component to both signals. We use a similar approach to that adopted by Treece *et al.* to extract the periodic component and its phase, i.e., we apply a bandpass filter to eliminate the components outside the typical cardiac frequency ranges, determine the dominant frequency, and apply matched sine and cosine filters for quadrature detection of the phase.

B. Data Acquisition

To evaluate our algorithm, we acquired 4D data using a commercial ultrasound scanner (Hitachi Hi Vision™ 8500) equipped with a mechanical convex probe (Hitachi EUP-CV524). Adult cardiac data from a healthy male volunteer were used to test our approach. A motor controller application running on the same machine was used to adjust the speed and the angle of the probe swing. A single, very slow sweep was utilized in all the acquisitions. Individual B-mode frames were stored in the cine memory of the system during the sweep and later written into the hard disk as uncompressed AVI files.

An ECG option was available in the ultrasound machine, and the electrodes were attached to the subject during the acquisitions in order to obtain visual ECG traces that could later be used for comparing ECG-based and image-based gating.

Due to the large footprint of the abdominal mechanical convex probe we used, standard parasternal and apical views of the adult heart were not available. Instead, subcostal window was used to acquire non-standard views that encompassed partial views of the heart. Since our objective was not to perform a proper diagnostic evaluation, any data set containing moving heart structures would be adequate for the purpose of demonstrating the feasibility. We acquired a total of 11 different cardiac data sets with varying sweep speeds and angles.

C. Data Processing and Analysis

All the acquired data were later processed in MATLAB. The AVI files contained the entire display area from the ultrasound machine including the captions and the ECG traces. Therefore, such parts were stripped off from each frame to leave only the internal ultrasound image areas. The centroid algorithm was then applied to each frame to compute both the $centroid_x(t)$ and $centroid_y(t)$. A bandpass filter with a passband of 0.5-2 Hz was then applied to each centroid signal, and a periodogram was computed to identify the dominant frequency, which was subsequently used to design the matched single-period sine and cosine filters. Instantaneous cardiac phase information was then reconstructed with quadrature detection using the matched filters.

After the phase reconstruction, the average number of B-mode frames per cardiac cycle was calculated, which was then used to determine the average phase increment per frame (in radians/frame). For each phase increment step from 0 to 2π radians, frames were clustered into bins by selecting the frames with the nearest matching phase, and a volume was reconstructed using the frames from the corresponding bin. Multiplanar images were then rendered from each volume and played through a cine loop containing one full cardiac cycle.

Also, in a subset of data sets, frame numbers containing the ECG R wave peaks were visually identified from the ECG traces that are visible in the original AVI files, and volume reconstructions were performed from these frames. Reslices from these volumes were then compared visually to the corresponding reslices from the image-gated reconstructions using the centroid algorithm.

III. Results and Discussion

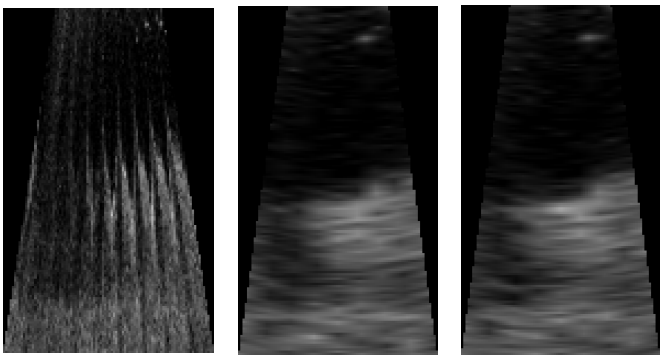
Figure 1 shows the reslices from one adult cardiac data set without gating, with ECG gating and with image gating. It can be seen how both the ECG- and image-gated reconstructions avoid the motion artifacts visible in the non-gated reconstruction. For gated reconstructions, both the ECG-based and image-based methods yield similar images, with almost negligible differences. These differences can be attributed to errors in visually identifying the peak in the ECG R wave. In all the data sets we have acquired, the $centroid_y(t)$ signal has almost always resulted in a clean sinusoid after bandpass filtering, while the $centroid_x(t)$ signal was noisier. We believe that the signal-to-noise ratio (SNR) for the y component was higher because of the finer sampling of motion in the axial direction than in the lateral direction. This is due to both the scan geometry and the axial vs. lateral resolution of the transducer. Therefore, we decided to use the y component of the centroid for all reconstructions. Figure 2 shows one example of $centroid_y(t)$-based phase detection for the same data set as in Fig. 1. The centroid algorithm performed well in all 11 data sets we acquired with no user interaction.

Although gated reconstructions avoid motion artifacts, there is visible loss of resolution in the elevational direction due to the sparser sampling of the anatomy in this direction.

The number of samples in the elevational direction is proportional to the number of 2D scan planes that are assigned to the same bin during the phase-based clustering of the original frames. Typically, in each bin, there will be as many scan planes as the number of cardiac cycles captured, which is approximately given by:

$$N_{cardiac} = \frac{f_H}{f_B} \times \frac{S_{buffer}}{S_{Bframe}} \quad (2)$$

where $N_{cardiac}$ is the number of cardiac cycles, f_H the average heart rate in Hz, f_B the B-mode frame rate in frames/s, and S_{Bframe} and S_{buffer}, in bytes, the size of one B-mode frame and the buffer size used to store such frames, respectively.



(a) (b) (c)

Fig. 1. Elevational slice (perpendicular to the plane of acquisition) after volume reconstruction with (a) no gating, (b) ECG gating, and (c) image gating with the centroid algorithm.

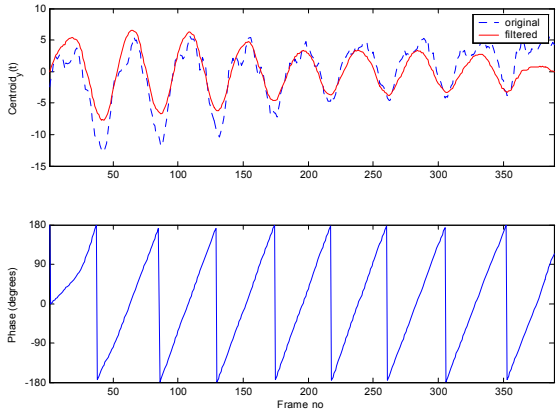


Fig. 2. Phase estimation using $centroid_y(t)$. Original centroid signal (after centering around the mean) and the bandpass-filtered signal are shown at the top while the detected phase after matched filtering is shown at the bottom.

Sampling in the elevational direction can be improved by increasing the duration of acquisition, i.e., lowering the swing speed, which will require increasing the total store buffer size, decreasing the size of B-mode frames, or decreasing the B-mode frame rate. In most cases, duration of acquisition cannot be arbitrarily extended since patient movement is more likely to occur during longer acquisitions, making the reconstructions more prone to artifacts. On the other hand, elevational sampling can also be improved by increasing the heart rate, which we do not have direct control over. However, for fetal applications, the heart rate is typically 2-2.5 times faster than the adult heart. Therefore, a 2-2.5 times improvement in elevational sampling would be possible.

Computationally, the most intensive part of the centroid algorithm is the calculation of the centroids for each scan plane. For a scan plane that consists of N pixels, one component of centroid calculation requires N multiplications and 2N-2 additions, followed by a single division. Most modern mediaprocessors support single-instruction multiple-data (SIMD) type of instructions, like multiply-add or inner-product operations, where more than one pixel can be multiplied and accumulated in a single instruction. Furthermore, most modern mediaprocessors come with multiple functional units, where several of these instructions can be issued in parallel in every cycle. For example, the recently introduced Texas Instruments TMS320C6455 mediaprocessor can perform eight of the required multiplications along with the necessary additions in a single clock cycle, and it can run up to 1 GHz of clock frequency. For a scan plane of 512 × 512, this would mean an ideal computation time of 32 μs excluding the cache misses. The post-processing required to extract the phase information from the centroid signals require only a couple of 1D FFTs and FIR filters, all of which can be performed at speeds that will almost be negligible compared to the centroid calculation, which is a 2D operation.

Many other ways were also proposed to estimate the cardiac cycle from ultrasound images. It is possible to estimate the average heart beat using temporal Fourier transforms [6] for all the pixels within a region of interest. However, this approach is both computationally very expensive and cannot accommodate variations in the heart rate during acquisition. Another proposed method is to use tissue Doppler signals for gating [7]. Although, this could be justifiable for color Doppler acquisitions, it would unnecessarily decrease the frame rates for B-mode acquisitions since additional firings will be required for each scan line. Recently, spatiotemporal image correlation [8] has been implemented in several high-end commercial ultrasound systems, and has been reported to perform very well in the limited-patient clinical studies. However, real-time image correlation requires a lot of computing resources.

Treece *et al.*'s approach is both computationally simple and can accommodate variations in the heart rate similar to the more sophisticated spatiotemporal image correlation. However, it is not suitable for real-time processing due to the need for user interaction. On the other hand, our approach enjoys the computational efficiency of Treece *et*

al.'s algorithm in tracking heart rate changes, but does not require any user interaction. This makes it clinically practical for use on 4D systems. Computational savings offered by our algorithm compared to spatiotemporal image correlation can free up some of the resources, which can be allocated to other compute-intensive tasks, such as the volume reconstruction and rendering.

While 3D and 4D fetal echocardiography clinically has almost exclusively been used with the mechanical probes, experience with the 2D matrix array transducers has also been reported. However, "live" 4D acquisitions with these have been found to be inferior in image quality to those acquired using mechanical probes with image gating [9]. Also, while high-end systems are more likely to be equipped with 2D matrix array transducers, mechanical probes are more likely to be favored in mid-end systems due to their lower cost. As 2D matrix array transducer technology improves, this could change. However, gated acquisitions would still be needed in the foreseeable future for extended volume and color Doppler acquisitions using these probes.

IV. Conclusion

3D and 4D ultrasound is finally making its way from research labs into clinics. Its success will depend not only on the image quality and the diagnostic information it can provide, but also on the tools that are employed to visualize, interact and analyze the volumetric data. Given the amount of data in a 4D volumetric scan and ever-increasing numbers of imaging exams, it is almost unrealistic to expect the users to optimize the acquired data sets manually and achieve the patient throughput requirement at the same time. On the other hand, full automation almost always requires high computational complexity, leading to higher system costs. In the era of distributed diagnosis and home healthcare in the future, where lower-cost equipment will be an important requirement, this trade-off will definitely be of concern to the system developers, while at the same time, providing exciting opportunities for the research community.

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