INTERACTIVE RECONSTRUCTION AND VISUALIZATION OF DYNAMIC 3D ULTRASOUND DATA SETS

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Abstract: In this paper we describe techniques which have been developed in order to interactively visualize dynamic 3D ultrasound (US) volume data sets acquired using a small animal 2D US scanner. The 2D frames are acquired at different spatial positions and gated based on the ECG signal. For the reconstruction process a user interface is specified, which allows the user to select 2D frames acquired during different ECG phases at different spatial locations. Thus it is possible to reconstruct and visualize a volumetric model based on several 2D frames for different phases of the heart cycle. To improve the interactive visualization, we facilitate filtering techniques, which have been implemented on the GPU.

1 INTRODUCTION

Medical imaging has the intent to harness the ever-growing graphics computing power of modern architectures for the visualization of medical data. A major goal is the visual analysis of images of biological objects acquired with different medical scanning technologies such as CT, MRI or PET. One important application of medical imaging is cardiac and cardiovascular research. According to the World Health Organization, cardiovascular diseases (CVDs) are the number one cause of death globally, representing 30% of all global deaths (Organization, 2006). Therefore researchers need to better understand the functions and processes of the heart in vivo. Non-invasive small-animal imaging techniques have a great potential to allow a better understanding for both researchers and physicians in the inquiry and prophylaxis of cardiovascular diseases. The potential of a system allowing an easy reconstruction and visualization of medical data is immense and lies primarily in its ability to support the members of the physicians in their decision-making process concerning proper patient treatment and disease prophylaxis by visualizing internal structures. In addition, a 3D visualization has benefits in contrast to classic 2D visualization systems. For instance, when visualizing a dynamic 3D heart, a physician is able to make accurate and reliable 3D observations of heart movement and ventricle deformation during a heart cycle as well as volumetric measurements.

Although US technology is used in a large number of biomedical imaging areas, the registration and visualization of moving internal structures such as heart ventricles and blood flow or prenatal fetal facial features are among the more prominent examples. Consequently, the automatic generation of 3D US volume data from moving biological structures scanned at a high resolution adds to the complexity of the task of volumetric visualization primarily because of the incoherency of the single 2D scan slices with respect to creating a static volume out of them, and secondly, because of the relatively high speed of movement of certain organs. Therefore, a certain additional alignment measure has to be introduced in order to be able to create coherent volume representation of organs from slice scans of moving tissue.

In this paper we propose an interactive visualization system, which allows to acquire dynamic volumetric ultrasound (US) data sets by using a 2D scanner. In particular we describe the preprocessing as well as the visualization of raw US data and explain how a user can interactively create and manipulate volumetric data sets originating from 2D US B-scan slices. Therefore the 2D US B-Mode scan data and metadata is analyzed and prepared for display. De-

pending on the type of the scanned data, the user may choose between automatic generation of a volume data set or an interactive generation. Due to the relatively noisy nature of the US data, several filtering techniques are applied in order to improve the visual quality of the data and remove artifacts introduced by the acquisition procedure.

2 RELATED WORK

US has been used in many medical subdomains, as for instance prostate and heart examinations (Gee et al., 2003). Fenster and Downey give an extensive overview over developments in 3D US technology and visualization (Fenster and Downey, 2000). Thus, several techniques for US data acquisition with the purpose of creating volumetric data sets have been developed up until now. Gee et al. (Gee et al., 2004) divide these techniques into two major categories. On the one hand, volume data sets can be generated by utilizing a dedicated US 3D probe which can scan a fixed-size volume from a short distance. On the other hand, conventional 2D US transducers can be employed for the generation of arbitrarily-sized scalar fields. But the latter case requires the process of volume reconstruction and temporal and lateral 2D frame synchronization. In addition, the irregular spatial distribution of the 2D slices adds additional complexity to the process which may result in different scan results of the same anatomical structure under differing directions of sonification. Gergs et al. (Gergs et al., 2004) have shown that besides 3D US data sets, 2D data sets have also the potential to show clear silhouettes for certain application cases.

The *side-fire probe* which is used for acquiring circular B-scan slices forming cylindrical data sets (Gee et al., 2003) is similar to the technique described in this paper. However, an additional positional sensor is attached to the transducer head whose constant input is used for receiving the relative positions and angulation of each 2D slice and then aligning them within the volume cuboid during volume generation (Fenster and Downey, 2000).

To improve the visual quality of the acquired data sets, different filtering techniques can been used. In (Sakas et al., 1994), Sakas et al. have attempted to evaluate the deployment of a combination of several low-pass filters for improving the quality of visualized US data sets through filtering. In their work, they have used a 3D Gaussian filter for noise/speckle reduction and a 3D median filter for additional contour smoothing and small gaps closing caused by misalignment of consecutive B-scan slices. They claim

that the results of this undertaking has been encouraging, but still see room for improvement by modifying the filtering technique and adjusting it to the nature of the US signal. Besides the application of Gaussian filters, the BLTP filtering technique (Sakas and Walter, 1995) has been developed specifically for the purposes of filtering US volume data. However, discussing these filters in detail is beyond the scope of this paper and we refer to (Gonzalez and Woods, 2001) for more information.

3 SYSTEM SETUP

In our system setup we use a VisualSonics Vevo 770 small-animal US scanner for scanning mice in vivo. It provides most common US scan modes, and supports ECG-gated acquisition of 2D slices. It allows multiple types of physiological measurements on animals, from which we have used the ECG signal. Additionally, the Vevo 770 scanner performs respiration gating by acquiring data only during the rest sub-period in the respiration cycle. In doing so, excessive moving artifacts caused by the heaving of the chest cavity of the animal are obviated.

We use a special scanning mode which is a proprietary variant of an ECG-gated scanning procedure. The so-called EKV (ECG-based Kilohertz Visualization) mode is a scanning technique, during which multiple B-Mode images from consecutive heart periods are acquired and interpolated into a single, complete heart period of the animal (VisualSonics, 2005). Thus a complete heart period is represented by a set of frames for each lateral position. Therefore, in using the EKV-Mode in this paper, the creation of precise volume "stills" of the scanned organ simplifies the frame synchronization routine significantly. Thus, by choosing the corresponding frames out of each synchronized frame set recorded at each lateral position, a volume data set can be created for each phase of the heart cycle. Subsequently, the volume data sets reconstructed in this manner comprise a whole dynamic data set collection.

In order to acquire an EKV cine loop of the heart period, the scanner needs proper ECG-data from a probe attached to the animal. This is accomplished by sensing the animal's heart rate from four small fields on the warmer plate where the mouse is fastened. Constant monitoring and supervision of the animal's vital signs and telemetry is required. Additionally, the animal's temperature is monitored through a signal delivered by a small rectal probe connected to a module responsible for monitoring the overall physiological condition of the scanned animal. Additionally,

the respiration rate of the animal is derived from its ECG signal and also delivered to the monitoring unit. For proper ultrasonic acoustic transmission, the space between the animal's chest and the US transducer is filled with a layer of water-based US gel in order to diminish the attenuation of the reflected wave signal.

4 INTERACTIVE DATA PROCESSING

In this section we describe our approach allowing interactive 3D reconstruction of 3D US data sets based on 2D slices acquired with the system setup described in Section 3. After the reconstruction has been performed (see Subsection 4.1), image processing techniques are applied in order to enhance image quality (see Subsection 4.2).

4.1 Volumetric Reconstruction

We support two approaches for reconstructing a 3D US data set out of 2D slices. First, the standard ECG-gated automated technique, and second, an interactive technique. As mentioned in Section 1 the automatic reconstruction, which can be considered as a well known standard technique, has several drawbacks. With this technique, 2D scans of the heart are captured only at designated locations within the ECG heart cycle, thereby ensuring a relatively coherent volume data set. However, the assumption that heart form and deformation during each heart cycle is identical during the whole duration of the acquisition process is misleading. Firstly, especially in cases where the heart of the subject is contracting in an irregular manner either due to immediate application of a medication or to some kind of a heart disease, each ECG heart cycle differs considerably from the other, and this results in a large number of incoherence artifacts in the resulting volume. Secondly, additional external factors such as respirational movement or movement caused by instability of the biological object can also cause irregularities in the single scans, resulting again in incoherent data sets.

Our alternative interactive approach alleviates these problems during the volume reconstruction by integrating the user input into the process of selecting the required number of slices and situating them at their respective locations in order to construct a volumetric data set. In order to do so, the ECG curve of the heart rate during which the 2D slices have been recorded is shown. Based on this display, the user can select an arbitrary slice depending on its position as well as its location within the heart cycle.
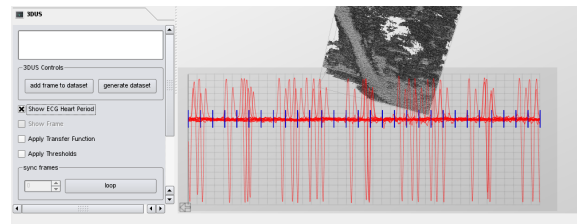

Figure 1: Visualizing the entire sequences of ECG cycles leads to visual clutter and makes it difficult for the user to identify a single frame within the whole cycle.

However, displaying the ECG cycle is a critical task. Therefore, in order to extenuate incoherency artifacts in the ECG cycle, the Vevo 770 US scanner is configured to record high resolution ECG data throughout the whole B-Mode scanning session. However, this accumulated high bandwidth of ECG frames alongside 2D US scans poses another problem. The length of the recorded ECG data exceeds largely the length of the period of B-scan recording because the US scanner keeps as many sequences of ECG heart cycles as possible for purposes of interpolating them all into a spatially precise heart cycle synchronized with the animal's ECG. However, this excessive amount of ECG heart cycles should not be displayed entirely because the large number of single ECG frames clutters the display (see Figure 1). In order to create a usable display of the long ECG sequence of multiple heart periods, the interval of ECG frames is cut off at the points which coincide with the subinterval between the time stamps of the first and last recorded frame. Thus we obtain an ECG display with considerably less cluttering, as it is seen in the ECG overlay in Figure 2. The blue lines on top of the ECG curve represent the slices which have been recorded. The information about each slice is extracted from the meta data, which is generated by the scanner. Besides other, this meta data also contains X- and Y-resolution of each frame, start and end timestamps of the ECG data portion so that ECG samples can be extracted, B-Mode ECG frequency enabling the accurate positioning of frames in the ECG heart cycle graph, heart period length and each frame's dimensions in *mm* with respect to the scanned object.

After a slice has been selected by clicking on the overlay display, its content can be displayed. In the case of US slices, a greyscale image containing the intensity values of the US 2D frame is generated and inserted into the display. This gives the user the control over which slices to select for the data set and at which positions to situate them. Moreover, the additional display of the slice contents allows the user to additionally "prefilter" those slices which, although situated at a suiting position, contain a large amount of
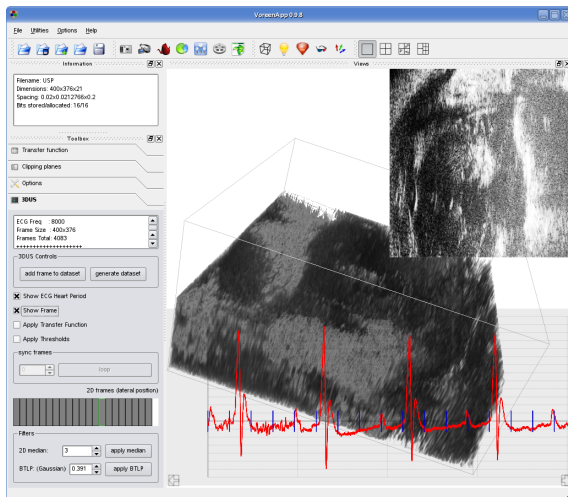
Figure 2: The user interface supporting interactive US volume reconstruction. The ECG curve is reduced to avoid clutter and a B-scan frame display overlay is shown, whereas the frame is enhanced by applying histogram equalization.

noise and their insertion in the volume would merely worsen the volume generation with respect to the geometrical consistency of tissue surfaces. After the user selects the required number of slices, the contents of each slice are concatenated, combined into a data set and can be stored and later on visualized.

Figure 2 shows the user interface which is used for the interactive 3D volume reconstruction. The text box at the top contains information pertaining to the current object, be it a volume data set or a single frame. The checkboxes below are used for activating different auxiliary tools such as ECG heart period overlay, B-scan frame overlay, application of the transfer function and the thresholding of the voxel values. All these features can be applied interactively by providing immediate visual feedback.

Besides these standard volume visualization controls, the system provides the interaction capabilities necessary for the volume reconstruction. The shown synchronization box is relevant for data sets acquired through Vevo's EKV-mode and can be used for playing a data set collection dynamically. The custom widget underneath has been developed to aid the positioning of single slices within a volume data set. By selecting one of the rectangular boxes, the user selects to display all frames which the US scanner has recorded at the corresponding lateral position of the scanhead. The currently selected position is highlighted by the green box contours and only those frames are visualized on top of the ECG signal, which have been acquired at the corresponding lateral position. After a frame has been selected and its contents

inspected, if necessary, the user can add its raw contents to the data set to-be-created by pressing the "add frame to data set"-button above. The current lateral position is then disabled and the reconstruction can be continued.

4.2 Image Processing

The US signal has a lower signal-to-noise ratio than other medical scanning paradigms such as CT or MRI, and, as a result, visualizing the raw, unmodified US data delivers data, which can be optimized by preprocessing. In order to deal with this effect, our system allows to apply different image processing filters to the selected slices. The filters can be applied interactively, and the type of filter as well as the filter parameters can be chosen using the user interface (see Figure 2).

We have integrated two kinds of filters: linear as well as non-linear spatial filters. Although (Sakas and Walter, 1995) suggests their usage only in a preprocessing step due to their resource-intensive computation, we were able to perform them interactively, since the dimensions of the generated US data sets are relatively small.

The linear filters operate in the spatial domain of the input signal and, as such, are used for blurring and noise reduction of the image data. Considering the nature of noise artifacts in the US signal, this family of filters is the most natural approach towards improving the visualization of US data. What is more, they remove small artifacts and noise from images which are very similar in structure and origin to common US noise and speckle.

Characteristic for all linear spatial filters is that they compute an intensity value of each pixel based on the intensities of a number of neighboring pixels selected by a square filter mask and a linear filtering function. Based on the latter, these filters are sometimes called averaging filters (Gonzalez and Woods, 2001). Since noise and image artifacts are substantiated in the greyscale intensity distribution of an image pixel grid as abrupt changes in grey levels, the averaging function applied during filtering brings all values to an approximate image-wide average whose value is determined by the surrounding pixel intensities and the width of the filter mask. Hence, the main effect of this type of filtering is blurring of image regions which differ radically from the rest of the image. The disadvantage of using such filters is that along with the noise removal, particular image segments such as edges or contours similar to noise but corresponding to the edge of a structure are also blurred, although they are parts of the image that should not be mod-

(a) no filtering

(b) arithmetic mean filtering (3×3 kernel)
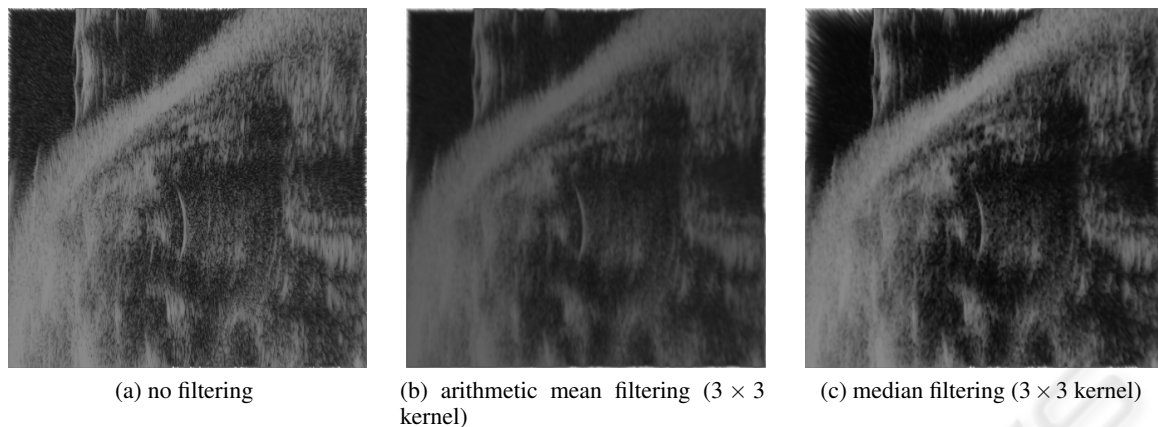
(c) median filtering (3×3 kernel)

Figure 3: Application of different filtering techniques to a single lateral slice. From left to right: no filtering, arithmetic mean filtering and median filtering.

ified. A simple linear spatial filter is the *arithmetic mean* filter, which computes the average sum of the pixels under the filter mask and assigns the resulting value to the center pixel, in the case of an equilateral filter mask and in the two-dimensional case:

$$I(p_j) = \sum_{i=1}^{k^2} p_i, \quad p_j \in P. \qquad (1)$$

In the above formula, p_i, p_j denote pixels in the set of pixels P belonging to an image and k denotes the side length of the equilateral filter mask. A slight modification of the arithmetic mean filter is using a mask which assigns to each pixel a different weight and thus different importance of each pixel with which it participates in the sum. However, both filter versions deliver similar results due to the fact that the filter mask is small relative to the size of the noise grains in the image. Its application is shown in Figure 3(b).

The median filter (see Figure 3(c)) is a non-linear spatial filter, which operates by ordering the pixels selected by the filter mask according to their grey level intensities, then choosing the *median* out of them and replacing with it the pixel intensity at the center of the filter mask. In comparison to a linear filter a median filter with the same sized filter mask delivers better results in noise reduction and results in less blurring. Thus sharp edges can be better retained in comparison to the mean filter (see Figure 3). Thus, we have reduced the functionality of the user interface to be able to apply the median filter instead of the mean filter.

Additionally, we have integrated the BLTP filtering technique originally proposed by Sakas et al. (Sakas and Walter, 1995). However, unlike their implementation developed for surface extraction, we have modified the opacity values to be binarized such that they are not all positive voxel values but, instead, they are selected through intensity thresholds which are dynamically controlled by the user. As a result, the criterion for removing a voxel value is determined interactively by a lower and upper threshold within the interval of the data set intensities. Thus a certain amount of noise can be removed manually, thereby improving the interactive rate of the filter's response.

To adjust and improve the contrast of the separate US B-scan images whose grey values are positioned within a low dynamic range, we additionally apply a histogram equalization.

5 APPLICATION EXAMPLES

The application examples for this paper have been prepared from US data acquired from scanning several laboratory mice specimen with genetically induced LQTS (long qt) syndrome for the purposes of medicament-based disease prophylaxis. Every animal's cardiac section has been extensively scanned while under light sedation, and the procedure has been executed in strict compliance with cardiac research norms dictated by the specifics supplied with each medicament prototype.

When generating the volume data sets for this paper, the scanhead was mounted onto a rail in order to improve mobility and produce a B-Mode image at several lateral positions within the animal (see Figure 4). The spacing between the B-scan mode slices along the travel axis was 0.25*mm*. Since the scanned cardiac region of the mouse is relatively small, at least twenty slices in traveling direction were required for sensible dimensions of the generated volume data sets. A major problem when creating the data set was aligning the acquired lateral frames from the mouse
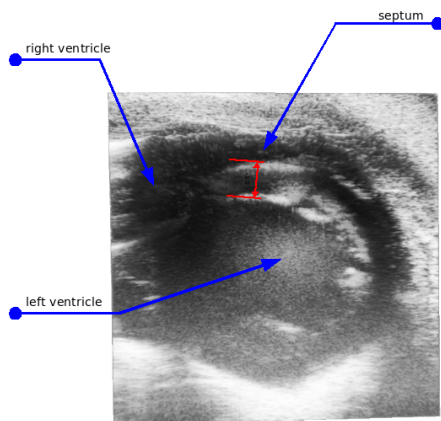
Figure 4: A single lateral slice showing a cross-section view of the heart.

heart precisely so that a volume image of a still heart could be created while the mouse was showing a heart pulse of 300-450 bpm in an anesthetized condition. After positioning the US scanhead over each lateral location, the image acquisition was initiated by switching the scanner to EKV mode. Considering the quality of the scans required, the EKV mode took several minutes per lateral position, extending the duration of anesthesia. In order to alleviate that, additional measures of maintaining the animal's temperature within the normal range were undertaken and the animal was placed under an infrared heating lamp.
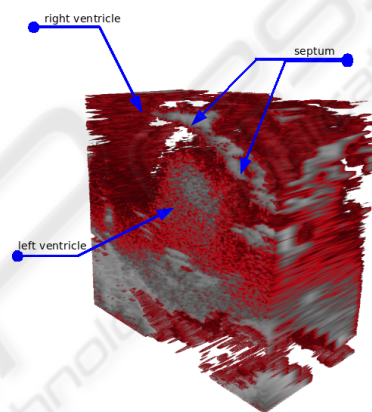
5.1 Application Case 1

One of the possible utilizations of medical US in cardiac research is the inquiry and measurement of the heart septum, i.e. the wall separating the left and right heart ventricle. One complication when this separating wall is damaged is known in the medical profession as Ventricular Septal Defect (VSD) (Lung and Institute, 2006), which results mostly from a small perforation located between the lower chambers of the heart. When untreated, this heart defect degenerates into, among others, congestive heart failure, arrhythmia and pulmonary artery hypertension. Cross-sectional slices can show the septums' condition only at the particular lateral position of the scan (see Figure 4). A 3D view of the complete septum allows much easier diagnostics of VSD and precise hole measurements in contrast with 2D US. Therefore we have conducted a scanning procedure in order to diagnose VSD in different mice.

In Figure 5(a) a rendered volume data set is shown which is clipped up to the same cross-section as in
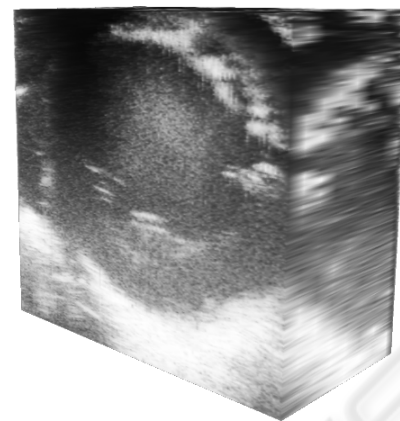
(a) Raw clipped data set (no filtering).

(b) Application of a BLTP filter.

Figure 5: 3D data sets acquired during application case 1, showing the same cross-section as shown in Figure 4. No image-processing is applied (top), a BLTP filter as well as a modified transfer function is applied (bottom).

Figure 4, such that this 2D slice is situated in the data set. It shows that the septum wall in the upper right corner of the volume cube is barely visible. In order to be able to show more details of the septum wall, several techniques are applied to the volume image.

The resulting data set is displayed in Figure 5(b), after applying a BLTP filter with underlying Gaussian with σ value of 0.621, resulting in a 1D kernel mask of $[1\ 3\ 1]$ and a BLTP threshold value of 0.15, thus removing small speckle regions from the data set. Although the filters and the thresholds have removed parts of the original image, the septum can still be distinguished in the bottom image. What is more, the applied transfer function allows for easier demarcation of each wall side of the septum, thereby revealing some of its anatomical properties such as width in both clipping directions and location within the heart structure. Additionally, by repeatedly changing the clipping parameters of the volume, the user can in-

spect the dimensions and spread of the septum wall in the whole heart volume interactively.

The abrupt discontinuities visible especially in the right upper side of the volume where the septum wall is located are caused by the insufficient amount of lateral frames that constitute this volume data set (in this case ~ 20). This is due to the fact that with spacing of 0.5*mm* between each lateral position, the granularity of the scanning installation is rather coarse in contrast to the longitudinal size of a mouse heart of approximately 1*cm*.
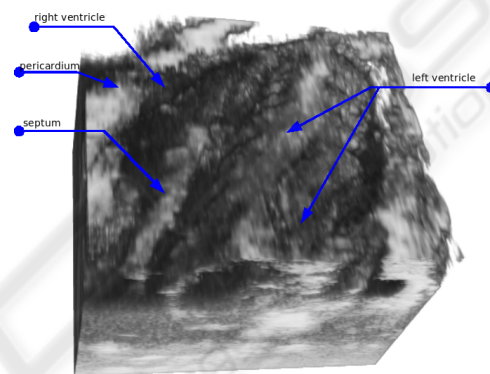
5.2 Application Case 2

In order to somewhat alleviate the shortcoming of too few lateral slices, the granularity of the scanning installation has been increased with an additional scale in sub-millimeter precision. In Figure 6 two renderings of thus generated data sets are shown, the lateral slices have been acquired with a spacing of 0.25*mm*. This improves the overall visualization quality and in particular the coherency of the single structures. The effect is especially visible in Figure 6(b), where the enclosing wall of the left heart ventricle can be viewed as an almost coherent and contiguous structure. What is more, the ridge labeled as septum can be followed from the frontal beginning of the cross-section to the very end of the heart - the apex. Additionally, the same filtering is applied to the data set: BLTP with underlying Gaussian of σ value of 0.621 and BLTP threshold of 0.42. In this case, the filter does not remove significant parts of the heart structure but merely blobs and artifacts hindering the proper display of the inner anatomy.

6 CONCLUSIONS

In this paper we have presented techniques comprising a possible 3D US preprocessing and visualization scheme. With the proposed system we are able to reconstruct 3D US data sets from 2D slices by applying a new interactive reconstruction approach. To facilitate reconstruction, the 2D lateral scans are visually organized based on their position as well as their phase within the heart period, which is recorded during the scans by using ECG. In Contrast to existing techniques, which reconstruct a volume automatically, the user has additional influence when using our approach and is thus able to reconstruct volumetric data sets having a higher quality. In addition to the proposed reconstruction methods, we have applied filtering techniques in order to reduce artifacts.
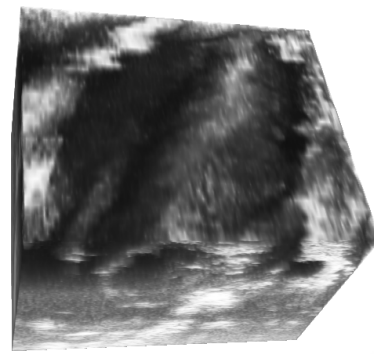
(a) Raw clipped data set (no filtering).

(b) Application of a BLTP filter.

Figure 6: 3D data sets acquired during application case 2. No image-processing is applied (top), a BLTP filter as well as a modified transfer function is applied (bottom).

Thus, the proposed interactive approach towards volume reconstruction encompasses a flexible way for exercising maximum control when constructing volume data sets out of 2D slices by allowing the user to select each single slice manually. By displaying the contents of each slice and its temporal position in the heart ECG, it provides for a user-based "pre-filtering" of the final volume contents.

REFERENCES

Fenster, A. and Downey, D. B. (2000). Three-dimensional ultrasound imaging. In Beutel, J., Kundel, H. L., and Metter, R. L. V., editors, *SPIE Handbook on Medical Imaging - Volume I: Physics and Psychophysics*. SPIE Press.

Gee, A., Prager, R., Treece, G., and Berman, L. (2003). Engineering a freehand 3D ultrasound system. *Pattern Recognition Letters*, 24(4-5):757–777.

Gee, A., Prager, R., Treece, G., Cash, C., and Berman, L. (2004). Processing and visualizing three-dimensional ultrasound data. *British Journal of Radiology*, 77(2):186–193.

Gergs, U., Boknik, P., Buchwalow, I., Fabritz, L., Matus, M., Justus, I., Hanske, G., Schmitz, W., and Neumann, J. (2004). Overexpression of the catalytic subunit of protein phosphatase 2A impairs cardiac function. *J. Biol. Chem.*, page M405770200.

Gonzalez, R. C. and Woods, R. E. (2001). *Digital Image Processing*. Addison-Wesley Longman Publishing Co., Inc., Boston, MA, USA.

Lung, N. H. and Institute, B. (2006). Ventricular septal defect. Online; accessed on 20-March-2007.

Organization, W. H. (2006). Fact sheets: Cardiovascular diseases. Online; accessed on 23-March-2007.

Sakas, G., Schreyer, L.-A., and Grimm, M. (1994). Case study: visualization of 3D ultrasonic data. In *VIS '94: Proceedings of the conference on Visualization '94*, pages 369–373, Los Alamitos, CA, USA. IEEE Computer Society Press.

Sakas, G. and Walter, S. (1995). Extracting surfaces from fuzzy 3D-ultrasound data. In *Computer Graphics Proceedings, Annual Conference Series SIGGRAPH 95*, pages 465–474.

VisualSonics (2005). Vevo 770 high-resolution imaging system: Operator manual.