Live ultrasound-based particle visualization of blood flow in the heart

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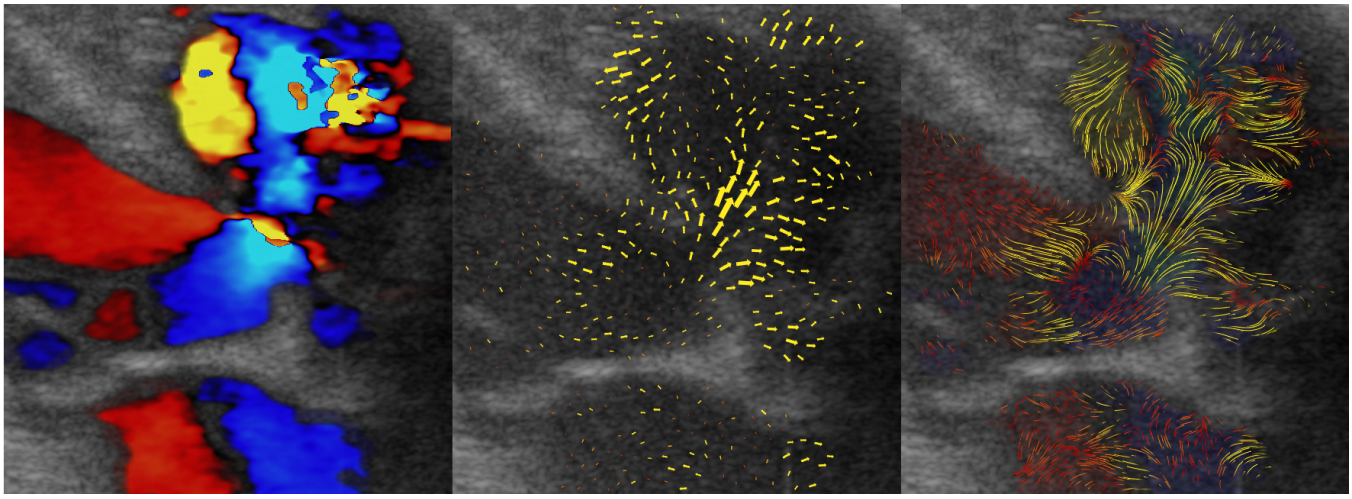

Figure 1: An instant of the systolic phase in a neonate with a perimembranous ventricular septal defect, visualized using color Doppler (left), arrow plot (center), and pathlets (right). The pathlets-based visualization is more effective during playback, and it can also better convey complex flow structures such as the shunt jet through the septum and the vortices in the right ventricle on the top right of the image.

Abstract

We introduce an integrated method for the acquisition, processing and visualization of live, in-vivo blood flow in the heart. The method is based on ultrasound imaging, using a plane wave acquisition acquisition protocol, which produces high frame rate ensemble data that are efficiently processed to extract directional flow information not previously available based on conventional Doppler imaging. These data are then visualized using a tailored pathlet-based visualization approach, to convey the slice-contained dynamic movement of the blood in the heart. This is especially important when imaging patients with possible congenital heart diseases, who typically exhibit complex flow patterns that are challenging to interpret. With this approach, it now is possible for the first time to achieve a real-time integration-based visualization of 2D blood flow aspects based on ultrasonic imaging. We demonstrate our solution in the context of selected cases of congenital heart diseases in neonates, showing how our technique allows for a more accurate and intuitive visualization of shunt flow and vortices.

CR Categories: I.3.3 [Picture/Image Generation]: Display algorithms—Line and curve generation;

Keywords: ultrasound medical visualization, real-time visualization, blood flow visualization

1 Introduction

The acquisition and visualization of in-vivo blood flow information in the heart has, until now, only been viable using either phase-contrast magnetic resonance (PC-MRI) [Bock et al. 2010] or ultrasonography, mainly by the Doppler method [Kasai et al. 1985] or derived techniques [Sengupta et al. 2012]. Of these two imaging techniques, only ultrasonography is capable of capturing and providing blood flow information in real-time during the acquisition. This makes ultrasonography an invaluable clinical tool for diagnosing pathological cardiac conditions.

Ultrasound scanners typically employ the Doppler method for capturing blood flow information, which identifies the direction and velocity of the blood flow along the ultrasound beam axis. Such data have been commonly visualized using a red-to-blue colormap where the red color represents areas where the blood is flowing toward the probe, while the blue color indicates blood that flows away

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from the probe. Recently, new methods have emerged to overcome this limitation of the Doppler technique, enabling to also capture the lateral components of the blood flow [Sengupta et al. 2012]. However, these new technologies also come with a number of drawbacks, most prominently the necessity of post-processing and the limited effectiveness of the currently employed visualization methods (see Section 2 for a discussion of the current state of the art). The limitations of currently available flow visualization techniques generally based on arrow plots and other instantaneous techniques, are also determined by the great amount of high-frequency data that is continuously acquired using ultrasound imaging. However, such data can be of great help in the diagnosis of certain conditions, for instance to evaluate congenital heart diseases (CHD) in neonates, where the blood flow pattern can be particularly complex. In such situations is highly desirable to study the non-instantaneous movement of the blood and assess complex flow patterns occurring due to abnormal shunt, dilated ventricles, and more complex anatomical disorders.

In this article we present a new method for the live visualization of cardiac blood flow based on ultrasonic imaging. Our method is based on a non-approximated directional flow information extraction method and a real-time integration-based visualization technique, which is specifically tailored to high-frequency streaming data. The acquisition technique is based on a custom high frame rate Doppler imaging sequence [Løvstakken et al. 2011]. It can be used to derive flow velocities even in the presence of complex flow. The method is able to capture the high systolic velocities of blood and does not rely on the use of contrast agents. This acquisition technique is coupled with a tailored particle-based flow visualization that is specifically designed to visualize continuously streaming unsteady flow fields. The goal is to convey both the instantaneous shape of the directional blood flow field and, at the same time, the time-dependent blood flow trajectories (see Fig. 1 on the right), which greatly helps to understand complex blood flow, in particular in certain challenging pathological cases.

The main contribution of this paper is a new, fully integrated and GPU based real-time solution that is capable of extracting and visualizing 2D vector blood flow velocities, instantaneous flow information, and trajectories by means of an integration-based flow visualization. The components of our solution have been carefully tailored for running on the GPU, and for doing so in an efficient manner that allows real-time processing of input ultrasound data. This approach also provides an improved descriptive pathlet-based visualization, which has been carefully tailored to this specific scenario, and may allow for a more accurate detection of flow patterns in pathologic cases which are otherwise invisible or unclear, thus having the potential to increase the diagnostic power of Doppler ultrasound over the current state of the art. This is demonstrated for neonates with congenital heart diseases in Section 4 where we present examples of situations where our method shows a clear advantage over the standard color Doppler imaging or the simpler flow visualization methods previously used in ultrasound imaging. Being orders of magnitude faster than related previous work, our solution also offers performance suitable for bedside examinations and new opportunities for live examinations of time-critical pathologies. This can greatly reduce the examination time, and could potentially lead to discoveries that, otherwise, would not have been possible due to limited time for post-processing.

2 Related Work

In the area of cardiac flow imaging the capture of vector flow information in the heart has been possible, until now, only by using time-resolved phase-contrast MRI (PC-MRI) in 2D and 3D [Bock et al. 2010], and ultrasonography. PC-MRI is severely limited by the duration of the acquisition, which in turn requires patients with a very regular heartbeat and flow. Moreover, this modality often requires the injection of a contrast agent to capture reliable measurements [Bock et al. 2010], which means that it is not suitable in certain applications, for example in pediatrics. For these reasons this imaging modality has not yet found widespread clinical use. This modality, however, has been used to produce detailed 3- and 4-dimensional datasets of healthy volunteers. Thanks to this, even if these datasets are usually limited to one cardiac cycle and a few timesteps due to the duration of the acquisition process, PC-MRI has stimulated considerable research in the field of 3D blood flow visualization.

Unlike PC-MRI, ultrasonography is capable of a much higher frame rate, and thus it is more suitable for imaging highly dynamic processes such as blood flow, not requiring, in particular, the combination of data from different heart cycles. Ultrasound color Doppler imaging [Kasai et al. 1985] has in fact been used for decades to assess cardiac conditions during clinical examinations. The Doppler method alone, however, is only capable of capturing the velocity and direction along the ultrasound beam within a 2D or 3D sector. This means that both cross-beam velocity components are not measured. A partial improvement over standard color Doppler can be found in the work of Løvstakken et al. [Løvstakken et al. 2004], where, by enhancing the speckle pattern acquired in color Doppler, the authors make it possible to visually track the blood flow speckle pattern from image to image. This method is limited by the Doppler pulse repetition frequency (PRF, the frequency of subsequent ultrasound pulses), thus showing only low flow velocities. However, the last years have seen the emergence of new methods capable of extracting directional blood flow information on a 2D imaging plane, thus substantially improving the major limitation of the traditional Doppler method. The oldest of these method is *cross beams* Doppler imaging [Dunmire et al. 2000], where the blood velocity is measured from multiple angles and triangulation is used to reconstruct the 2D (and 3D) velocity vectors. While this method is capable of producing correct directional information, it has two major shortcomings. First, it requires multiple angled emissions per estimate, which lowers the PRF and therefore the measurable velocities as well as the frame rate. Second, the estimates are accurate only where the angled beams meet, which leads to a relatively small area that can be reliably measured. For these two reasons cross-beams Doppler imaging is mostly used in vascular applications.

More recently, two newer techniques have emerged, which overcome some of the limitations of cross-beams Doppler imaging: *Echo-Particle Image Velocimetry* (Echo-PIV) [Kheradvar et al. 2010; Kim et al. 2004], and *Vector Flow Mapping* (VFM) [Tanaka et al. 2008; Garcia et al. 2010]. The Echo-PIV method is inspired by optical Particle Image Velocimetry, where a tracer is injected into the flow in order to identify and track flow particles. In Echo-PIV, Contrast-Enhanced Ultrasound (CEUS, see [Angelelli et al. 2011] for a more in-depth description) is used as imaging modality, and feature tracking algorithms are applied on a frame-to-frame basis to extract the flow direction from the microbubbles' movements. Performing whole frame-to-frame tracking limits the maximum velocity that can be derived. CEUS imaging also meets constraints with respect to its use, despite the relative safety of this imaging modality, as the injection of contrast agents is considered invasive (e.g., it generally cannot be used in pediatrics). Unlike Echo-PIV, the VFM method is based on the Doppler method, and does not need the injection of contrast agent. The reconstruction of the beam-orthogonal flow component (angular velocity) is done using the continuity equation under a planar flow assumption. The drawback of this method is that planar flow is almost never the case

in the heart, in particular in pathologic cases. This implies that the complex flow patterns result in inaccuracies in the reconstruction of angular velocities.

Our new solution makes use of the *Vector Flow Imaging* (VFI) method [Nyrnes et al. 2007; Løvstakken et al. 2011], a recent method for vector flow data extraction. Compared to the three methods described above, VFI is capable of deriving angle-independent directional information from the whole imaged area, without the necessity of using a contrast agent, and without any assumption on the flow pattern. In this work we develop this technique further, in order to achieve real-time performance. The benefits of combining the acquisition, vector extraction and visualization into one real-time pipeline are manyfold, most notably that the method can now be used for bedside examinations.

From a visualization perspective, different visual analysis methods have been proposed to study the complex flow of the blood in the heart and in the principal vessels. Most of these focus on the extraction and selection of flow features, extending the more classical feature detection methods. These approaches among others include methods based on clustering [Van Pelt et al. 2012], Boolean line predicates [Born et al. 2013b] and information visualization techniques to convey quantitative information within the spatial visualization [Markl et al. 2011; Angelelli and Hauser 2011]. A more complete overview of methods for the visual exploration of blood flow data has also been recently published [Preim and Botha 2013]. Some approaches have also combined more classical integration-based flow visualization techniques, such as streamlines, pathlines, stream surfaces or particles with illustrative techniques. In this area, selected relevant solutions include toon-shaded particles with speed lines [van Pelt et al. 2011] and illustrative abstraction of flow features using stream tapes and hatching [Born et al. 2013a]. A more comprehensive overview of illustrative methods in flow visualization is available as a separate review [Brambilla et al. 2012]. All of these advanced methods require user-interaction to define several visualization and analysis parameters, like the seeding body specification, the visual primitives to use, various thresholds and case-specific selection specifications. This is usually performed in a frame to frame fashion, which is possible because these methods are designed for offline data visualization (the data are visualized after the examination). This also releases constraints on the computational requirements to produce the visualizations/analyses or on the necessary time for inspecting/using them.

With the ability of continuously producing highly dynamic flow information, new challenges arise in how to visualize such data. User interaction as required by the advanced methods illustrated above is not viable anymore. Therefore, in all the work on ultrasound flow extraction discussed above, the visualizations employed are limited to arrow plots and color maps. These visualization methods are certainly simple, but being an instantaneous, non-occlusive depiction of the flow, they work reasonably well when the data are captured continuously. They also do not require the user to adjust any seeding specification or other parameters. The downside is the limitation in the amount of information conveyed, as trajectories and complex patterns in the flow field are not easily recognizable. The visualization method we suggest in this work attempts to overcome this limitation, while, at the same time, being designed to effectively convey continuous data input. Our basic approach is based on the use of short integral lines [Markl et al. 2011]. We introduce several improvements to adapt the use of this type of visualization entities to the high frequency streaming data, described in detail in Section 3.3.

3 Method

To best describe our method for the live visualization of blood flow, it is convenient to illustrate it as a pipelined processing chain. Each stage of the pipeline is executed on the GPU, the generated data is kept in the GPU memory, and memory copies are minimized in order to achieve real-time performance.

The first stage of the pipeline is devoted to the data acquisition. Here, a special imaging protocol is employed, in order to achieve the necessary imaging rate to reliably extract vector information. This stage is described in detail in Section 3.1.

After the ultrasound raw data for one frame has been acquired, it is propagated to the vector extraction stage. In this stage, a feature tracking procedure is performed along the ensemble direction. For each acquisition frame, a 2-dimensional velocity vector map is created. This stage is described in detail in Section 3.2.

The 2-dimensional vector map is then propagated to the visualization stage. This last stage is in charge of performing the pathlet integration and rendering. Here, each unit consists of the front part of a pathline, which is integrated across different timesteps. An overall blood-refilling effect is achieved by replenishing the areas of the image that have been seeded in precedence and from which pathlets have departed. After the pathlets have departed, these areas are constantly reseeded with new seeding points. This stage is described in detail in Section 3.3.

3.1 Acquisition

A custom real-time acquisition was set up using linear array transducers (GE 9L and 11L) on a GE Vivid E9 ultrasound system (from GE Vingmed Ultrasound, Horten, Norway). To achieve sufficient frame rates that enable the tracking of blood speckles in the ultrasound images our setup is based on plane wave (unfocused) emission and the beamforming of 16 image lines per single pulse, in principle amounting to 16 times the frame rate of conventional imaging. The principle of parallel compared to conventional imaging is shown in Fig. 2, where it can be observed that the broad field-of-view obtained by unfocused imaging allows for image lines to be generated in a wider field compared to focused imaging. The main downside is a reduced penetration. Therefore these methods need to be validated for any given clinical application. With the current setup we were able to image to depths of 6-8 cm, sufficient for vascular imaging as well as for imaging the heart of neonate and small children. More details about this acquisition technique have been recently published [Løvstakken et al. 2011].

With the given acquisition setup, near instantaneous color-Doppler images at high frame rates and with high quality can be obtained. One raw color-Doppler frame consist of a temporal ensemble of images used for velocity estimation. The frame rate within an ensemble corresponds to the Doppler pulse repetition frequency which amounts to thousands of images per second. This ultra-high frame rate is available for the part of the image covered by the group of parallel images lines, and a complete image is formed by several (6-9) of such groups. In vivo data was acquired from neonates with congenital heart disease in the context of a study approved by The Regional Committee for Medical and Health Research Ethics (REK). Informed consent was obtained from the parents. Regarding patient-security, measurements in this study setup meet the claims from the American Food and Drug Administration for mecanical and thermal indices in neonates.

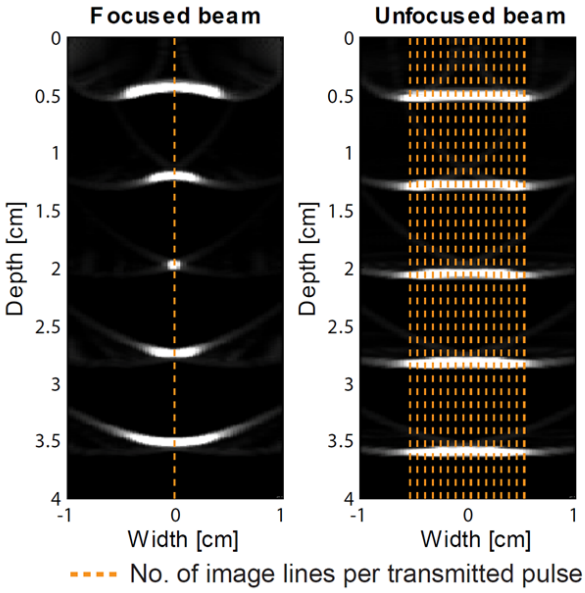

Figure 2: Conventional vs. plane wave ultrasound image acquisition. As illustrated, an unfocused pulse insonifies a broad field-of-view and allows for the generation of many image lines in parallel. The number of parallel image lines is basically limited by the available processing power and requirements of real-time imaging.

3.2 Vector Extraction

In order to generate our efficient pathlet-based blood flow visualization, the visualization stage needs to have estimates of the flow vectors in all regions of the image. The high pulse repetition frequency (PRF) used in the acquisition stage enables the robust extraction of velocity vectors by ensemble tracking methods [Løvstakken et al. 2011; Bohs et al. 1998]. The applied tracking method follows [Løvstakken et al. 2011], and the main points of this method are recapitulated in the following to keep the paper self contained.

In ensemble tracking, tracking is performed in the ensemble dimension (multiple consecutive samples of the same region) and only within a multiple receive beam (MRB) group. Assuming that the tracking approach uses a temporal lag of one pulse, this ensures that the kernel and the search regions are only spaced $1/PRF$ [s] in time. The speckle pattern within the same image region is thus highly correlated between consecutive ensemble samples. Another advantage given by the fast acquisition is the possibility to have larger ensemble sizes. This can be utilized to achieve more robust velocity estimates.

In order to extract velocity vectors, conventional template matching is used. An inherent challenge using template matching in echocardiographic imagery at such high PRFs is that much of the motion is subsample motion. This means the underlying structure has moved less than a sample between consecutive ensemble acquisitions. Straightforward template matching on speckle data will falsely report a zero velocity. One way to overcome this low velocity resolution is to upscale the data prior to template matching. This happens at the cost of higher computational demands for the matching. In addition, the results should preferably be refined by using subsample interpolation.

We have studied linear scan images. In the following description, dx and vx represent lateral displacement and velocity values, while dy and vy represent the axial direction. Our data were upscaled to achieve the preferred velocity resolution. A conventional template matching based on the sum-of-squared-differences (SSD) comparison criterion was performed, see equation 1. In order to reduce outliers, we utilized a large ensemble size and regularized the velocity vectors by taking the median of the results along the ensemble direction. To reduce the computational burden and prevent tapering the tracking results towards zero at the tissue boundaries, a segmentation based on the color flow image data excluding tissue areas with no flow was used to limit the region in which tracking was performed. Tracking results were refined using parabolic subsample interpolation [Lai and Torp 1999].

$$SSD(dx,dy,m) = \sum_{i=1}^{l}\sum_{j=1}^{k}\left(X_m(i+dx,j+dy) - X_{m-1}(i,j)\right)^2 . \quad (1)$$

In order to calculate the displacement for a particular point in the image, we first select a kernel region around it. A search region in the consecutive ensemble sample is also defined. Both the search and kernel regions are centered around the same image point. For each ensemble sample, m in the range from $m = 1$ to $m = M$,where M corresponds to the ensemble length, we calculate the displacements $(\tilde{dx},\tilde{dy})$ which minimize the SSD criterion 1.

$$(\tilde{dx},\tilde{dy})_m = \underset{(dx,dy)}{\arg\min}(SSD(dx,dy,m)) . \quad (2)$$

Due to the high acquisition rate, much of the movement is expected to be subsample movement. A parabolic interpolation scheme is thus adopted. Subsample interpolation is conducted separately in each direction. For the best match, the corresponding SSD value and its two nearest immediate neighbors in the direction of interest are used to fit the parabolic function. The argument yielding the peak value of the parabola is solved analytically and used to derive the subsample motion. Equation 3 shows the method for the x direction. The same equation is applied in the y direction, yielding subsample movement in the y-direction dy_m^{sub}.

$$dx_m^{sub} = \frac{SSD(\tilde{dx}-1,\tilde{dy}) - SSD(\tilde{dx}+1,\tilde{dy})}{2*SSD(\tilde{dx}-1,\tilde{dy}) - 2*SSD(\tilde{dx},\tilde{dy}) + SSD(\tilde{dx}+1,\tilde{dy})} . \quad (3)$$

The final displacements, after subsample interpolation, are written as $d\tilde{x}i_m = dx_m + dx_m^{sub}$ and $d\tilde{y}i_m = dy_m + dy_m^{sub}$. The final velocity vector estimate (v_x, v_y) can thus be written as:

$$(vx,vy) = PRF * (median(d\tilde{x}i_m), median(d\tilde{y}i_m)) . \quad (4)$$

Depending on the amount of upscaling and ensemble lengths, brute force template matching techniques may quickly become computationally unfeasible. To match the real-time characteristics of the visualization, the vector extraction step was sped up by implementing the SSD tracking on the GPU. Similar strategies have been shown effective also for other types of blood flow extraction problems, e.g., in Laser Speckle Imaging (LSI) for superficial blood flow [Yang et al. 2011]. However, we also chose to dynamically decimate and later interpolate the set of points to track in order to achieve the required performance for smooth bedside blood flow visualization. This decimation means that fewer image points are tracked, at the cost of a lower velocity field resolution. There is ongoing work to find better approaches for velocity vector extraction to shift this tradeoff between computational efficiency and vector field resolution more to the resolution side.

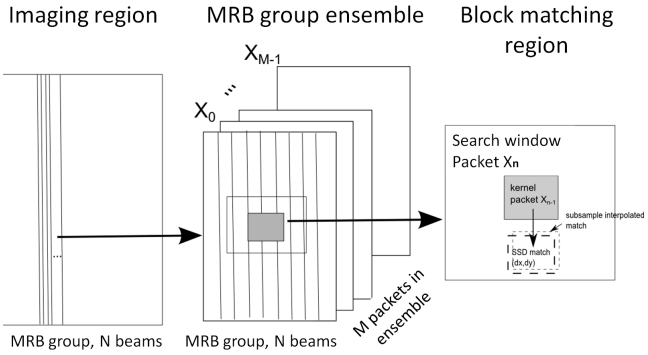

Figure 3: Illustration of ensemble tracking.

3.3 Visualization

At the end of the vector extraction stage, a vector field for each input ultrasound image is generated and passed forward. The goal of the last stage is to visualize the direction and velocity values in such a way that, at each point in time, not only the instantaneous direction of the blood flow is conveyed, but, unlike in previous works, also the time-dependent movement of the blood. The main benefit of having a visualization that conveys the blood trajectories is that it becomes possible to detect heart malformations that cause the blood to leak or spill into areas where this is not supposed to happen. Showing trajectories of the blood mass makes it also possible to detect vortices and other flow abnormalities in pathologic cardiac conditions. An additional challenge that must be tackled here is to enable the examiner to visually handle unsteady blood flow information that is continuously streaming from the acquisition stage.

To achieve the aforementioned goals while coping with streaming data, we propose to use a dense set of pathlets for the flow visualization. Pathlets are basically the leading short ends of pathlines, always of the same length as time proceeds. A pathlines is the spatiotemporal trajectory of an infinitesimal, massless fluid particle. Pathlines are very effective for conveying the movement of such (theoretic) particles, but they have multiple drawbacks in the presence of continuously updating data. The first is that, as time progresses, integrating longer and longer pathlines will crowd the visualization. Second, such lines convey only the trajectories of the particles they refer to. As the particles move over time, new fluid would take their place and flow thereon, but its trajectory would not be visualized. Third, the longer a pathline gets, the bigger is the discrimination between the time of the starting points of the pathline and the currently visualized time step. For instance, this can become confusing as the starting parts of pathlines may get increasingly non-parallel to the flow vectors of the currently depicted frame. Furthermore, integrating pathlines means accumulating small errors, for example from inaccuracies which naturally are contained in the data, in particular in reconstructed vector from imaging data. This means that, increasingly, starting points get visually associated with end points that – in reality – are not related. This error also increases exponentially, making long pathlines especially problematic.

In our visualization, the pathlets can be seen as the small, frontal parts of pathlines, that begin fading out when distance from the tip exceed a user-defined threshold. We generate the pathlets from a set of seeding points, and from their initial position they start traveling until they either end in an area with no flow, or their integration time exceeds a user-defined threshold. In addition, we use a user-specified number of seeding point sets. The number of seeding points in each set is kept the same, but the location of these points are determined using a Poisson disk sampling [Talmor 1997]. This introduces a certain amount of variability in the seeding positions, while preventing regions of the image to be left unseeded. We use this array of point sets so that, after a set of pathlets starts being integrated, regularly a new one is added after a certain time delay.

In addition to conveying trajectories which present a detailed picture of the flow structure, our method has several other benefits. First, a single set of pathlets does not clutter the visualization, in particular because these particles will eventually fade out. Second, our approach is designed to support highly dynamic scenarios, and our pathlets proved to effectively convey such continuously changing data (see Figure 4 and 5: the pathlets follow the axial direction reported by the instantaneous color Doppler). Finally, we gracefully handle possible data inaccuracies, caused by sampling a 3-dimensional flow on a 2-dimensional plane. One directional component is naturally missing, and in certain regions small or non-existant velocities can imply a flow in the normal direction. Our highly dynamic pathlets would suddenly become shorter and fade out in such regions, giving the impression of the particle going away from the plane.

During the method development, we also discovered that visualizing the trajectories at the actual acquisition rate, with the flow of the blood reaching velocities of over one meter per second under certain conditions, is particularly challenging to understand (or even to see). For this reason we included the possibility of slowing down the speed of the visualization. In our analysis, we discovered that visualization speeds of up to 0.4 times of the live acquisition speed are possible to follow. Such a lowered framerate does however impact the smoothness of the playback, in case that only the input frames are visualized. We therefore propose to generate additional visualization frames in order to match the maximum refresh rates of common computer displays (typically 60 or 75hz). To do so we employ linear interpolation to derive the data to visualize in between two subsequent images. In these frames we simply show the interpolated b-mode intensities, while we perform additional integration steps for the particles so that their movement appears smooth to the viewer.

This approach works well for recorded data, but in case of live streaming data it would introduce a dyssynchrony between the data captured by the acquisition and the data on the display. To solve this problem, we propose to define a fixed maximum amount of frames that can be buffered. When the visualization is performed at a reduced frame rate with respect to the acquisition frequency, after the buffer is filled, the data is discarded until the content of the buffer is completely consumed by the rest of the pipeline. At that point the process is initiated again, resynchronizing the second and third stage of the pipeline with the acquisition stage.

4 Demonstration

The vector extraction and flow visualization stages of the presented solution were prototyped in C++, supported by the OpenCL com-

Level of Detail	Tracking resolution	FPS
LOD0	138x299	3.6
LOD1	69x150	9.6
LOD2	35x75	17.8
LOD3	18x38	37

Table 1: Vector extraction performance. The performance increase appears to be quasi-linear despite the quadratic decimation of tracked vertices.

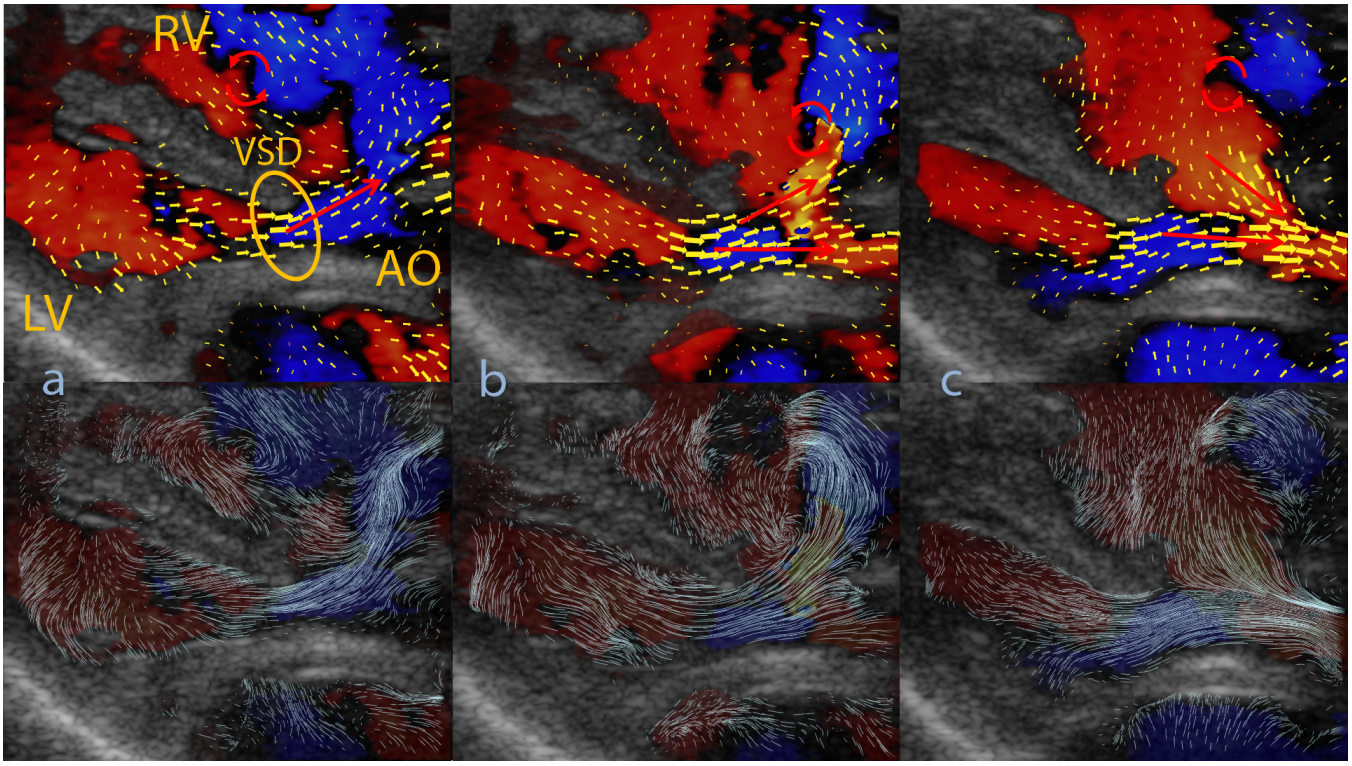

Figure 4: Three subsequent instants of the systolic phase in a neonate with double outlet right ventricle and a ventricular septal defect (VSD). Visualized using color Doppler and arrow plots (top row) and using pathlets (bottom row). Not only the pathlets are more effective during the playback, but also on still images they better convey the trajectories and complex flow structures such as bifurcations and vortices (illustrated in the top row). Our approach also matches well the flow structures highlighted in the color Doppler map.

puting framework and the OpenGL graphics library. The data acquisition was performed using a GE Vivid E9 ultrasound scanner (GE Vingmed Ultrasound, Horten, Norway), using the acquisition protocol described in Section 3.1. In our current prototype the vector extraction and visualization stages are run on a separate machine and not on the scanner itself. However, the process of integrating them into the scanner software is currently ongoing. The data is usually imported from a file for practical reasons, even if streaming it over TCP-IP directly from the scanner is a supported feature.

A performance analysis has been carried out on an Intel Core2 Quad 2.50Ghz workstation using a NVidia Quadro K5000 GPU. The data

Seed sets	Seeding points per set	Integration steps	FPS
2	1000	30	280
4	1000	30	220
6	1000	30	160
4	500	30	280
4	1500	30	160
4	1000	15	230
4	1000	45	160
4	1000	60	135

Table 2: Visualization performance. The number of seeding points (seeding sets and seeding points per set) seem to affect the performance linearly, while the length of the particle integration seems to have a nonlinear impact. The performance has been measured on pre-computed vector data to assess the cost of the sole visualization stage.

stream that has been used for this analysis consisted of ensembles of 12 images per frame, each of which has a resolution of 138x299 pixels, captured at a frame rate of approximately 40fps. The tables 1 and 2 report the performance results of the vector extraction and visualization stages respectively, using different resolutions for the points to track and different visualization parameters. From these numbers, the vector extraction stage is the most expensive in terms of computation. However, with a decimation factor of 8, it proved to be capable of processing approximately 37 fps, which is currently above the frame rate that we can use for a meaningful visualization (about 15fps on the same data stream). The performance of our approach is also orders of magnitude faster previously presented methods [Løvstakken et al. 2011], which allows the eventual deployment of a real-time solution that can be adopted bedside. In our experiments with the current implementation, the performance showed a linear increase with the quadratic reduction of the resolution of the tracking grid. We are currently still investigating the reason, but our first preliminary analysis suggests that the use of local memory in the GPU code has played a role in this. Concerning the visualization stage, we try to generate images at the same refresh rate as of the screen, typically 60Hz or 75Hz. This stage proved to be fast enough to keep up with the input data, even in presence of an amount of seeding points larger than what would be typically used in the visualization.

The presented technique has been used to examine the blood flow in CHD patients, and the results have been compared with color Doppler images, which is a method trusted by clinicians, as well as the gold standard in ultrasound flow imaging. Images from two neonates with congenital heart disease are shown in Figure 4 and 5. Figure 4 shows a neonate heart with a complex condition in-

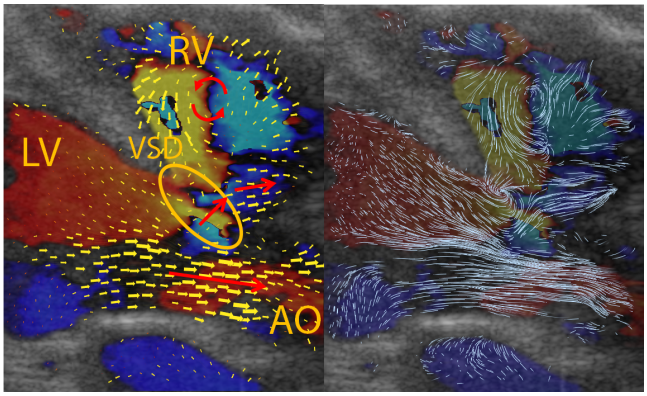

Figure 5: An instant of the systolic phase in a ventricular septal defect (VSD) patient. Visualized using color Doppler and arrow plots (left) and using pathlets (right). Pathlets help conveying the shunt flow going through the septum and the vortex in the right ventricle.

cluding a Double Outlet Right Ventricle (DORV, which means that the pulmonary artery and the aorta both arise from the right ventricle), hypoplastic left ventricle, and a large ventricular septal defect (VSD). The three images show three subsequent instants in the cardiac systolic phase, where the left ventricle contracts immediately before the right ventricle. In Figure 4a the left ventricle contracts, and there is a shunt flow (red arrow) moving the blood into the right ventricle. In Figure 4b the right ventricle also begins to contract, and pushes most of the shunt flow into the aortic outlet, but some flow still enters the right ventricle causing a vortex flow on the top, which is not apparent on the color Doppler. In Figure 4c the right ventricle finally contracts, and now it is noticeable how the blood flows out in the aortic outlet. These features in the flow are much more apparent and intuitive in the pathlet-based visualization (bottom row), which not only results more effective in still images, but in particular during the playback, when the time for interpretation is necessarily reduced. For the effectiveness of our method during the playback, we refer to the associated videos.

Figure 5, shows a case of a perimembranous ventricular septal defect. In this case there is an opening in the septum, close to the Atrioventricular (AV) plane. The figure shows an instant of the systolic phase, when there is a noticeable shunt flow going through the VSD into the right ventricle. The shunt flow is causing, also in this case, a vortex on the top. In both these examples, we superimposed our visualization technique over a semitrasparent color Doppler map, that is simultaneously acquired in an interleaved manner. By also allowing the color Doppler in the visualization, our pipeline can provide two different types of blood flow measurement and visualization together, thus increasing the level of confidence for the end user. Our method proved to have a very good agreement with the color Doppler information, in that the axial direction and velocity of our visualization matching the color Doppler. This is noticeable analyzing the particles' directions in the two images above.

We have also presented our method to selected cardiologists, gathering consistent feedback and informal evaluations, summarized below. The method has been very well received by the medical specialists, with the remarks that it provides a more descriptive but also more intuitive depiction of the flow, when compared to color Doppler and arrow plots. Figure 5 shows, for example, how the arrow plot leaves the task to mentally reconstruct the flow trajectory to the user and makes it more difficult to understand the flow especially during the live visualization. Our method has been considered to potentially enable the detection of a number of specific flow patterns which previously were not detectable. Among others, shunt flows which highlight leakages or septal defects in the heart, which is particularly important in neonates with CHD, as well as vortices and stagnating or recirculating blood, which have been previously associated with increased risk of thrombosis. It has been in fact reported that in more than 30% of the cases technical limitations are the cause of incorrect diagnoses [Benavidez et al. 2008]. It also helps to provide a more understandable picture to the patient, as it is easier to interpret than color Doppler.

5 Conclusion and future work

The ultrasound Doppler method has been deployed for decades as the most trusted examination and diagnostic tool in echocardiography. This method however has significant technical limitations that may reduce its diagnostic power, in particular in situations exhibiting complex flow patterns. Our solution aims at overcoming parts of these limitations, thus potentially increasing the diagnostic power of echocardiography. We demonstrated that our method is capable of deriving both the axial and the angular velocities from 2D ultrasound in real-time. We also demonstrated a tailored visualization technique that is capable of conveying continuously streaming data in an efficient way. Our method can also work in combination with color Doppler, to provide combined flow information and a higher margin of confidence to the viewer. Our visualization proved to qualitatively agree with the information coming from the well established color Doppler method. Offering real-time performance makes our method a candidate to be installed directly onboard on ultrasound scanners.

Our method has, however, still some limitations. The current unfocused acquisition strategy is limited to imaging down to approximately 10 cm in depth, which is suboptimal for adult echocardiography. It is also only capable of extracting the projected velocities on the image plane, meaning that an out-of-plane component is still missing. This is a general limitation of all 2D ultrasound blood flow imaging techniques, which can produce potentially misleading visualizations, but our method represents an improvement in this sense, even if not a definitive solution. Being aware of such limitation, we also tailored our pathlet-based visualization method to minimize its negative effect, e.g., by fading out particles in the proximity of low flow velocities. However coping or overcoming this limitation is also an exciting research challenge for the future, while being aware of the fact that the current solution still provides valuable insights to the examiner.

From this research we are also able to draw some interesting conclusions. First, enabling real-time visualization can make the difference between a posteriori analysis (too often not even possible due to time constraints) and immediate analysis, which can, in fact, also steer the acquisition as US imaging is steered by the examiner. Finding the "ideal" balance between direct flow visualization and integration-based flow visualization is challenging, especially when it comes to time-dependent flow visualization. Direct flow visualization is less prone to visualization errors, but less expressive; integration-based flow visualization is more sensitive, but clearly more expressive (also according to the feedback which we got). Finding the right level of "intelligence" in advanced visualization is crucial when it comes to the question of adoption, and our pathlet-based technique seems to be a promising compromise. Being able to capture directional flow information in real time, we also discovered that visualizing these data at actual speed becomes challenging for the viewer. At such speed, blood velocities in the order of magnitude of 1m/s that also change direction at a very high frequency proved, in certain cases, to be even not visible. It has been a com-

mon practice with Doppler imaging to slow down the playback of the acquired data to provide a visualization that can be understood by the viewer, technique which we also employ here. This is a general challenge when visualizing complex and highly time-varying flow fields, and is an interesting research topic that we plan to investigate further.

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