Proceedings of the 28th IEEE
EMBS Annual International Conference
New York City, USA, Aug 30-Sept 3, 2006

Automated Doppler gate placement and velocity calculation based on a vessel angle estimate

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***Abstract*— Pulsed-wave Doppler (PWD) sonography allows the quantitative measurement of blood flow in vessels if the vessel's diameter and the angle (Doppler angle θ) between the ultrasound beam and the vessel's flow direction are known. In current clinical routine, these parameters are manually determined by the examiner. However the manual determination is time-consuming and a source of error in blood flow measurements because of inaccuracy and variability.**

To overcome these problems, we present methods for an automated Doppler gate placement and Doppler angle estimation based on image processing algorithms. The Doppler gate is determined by analyzing the intensity profile along the ultrasound beam. To calculate the Doppler angle, we present a multiscale approach which estimates the vessel's flow direction by principal component analysis. A first evaluation shows promising results. Our automated approach yields parameters which match the parameters determined by clinical experts very well.

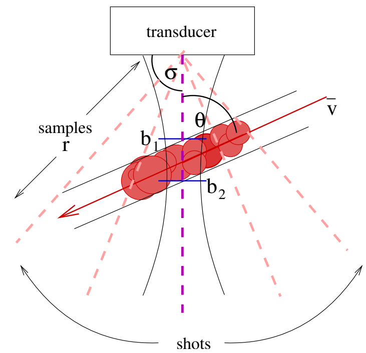


Fig. 1. The gate borders b_1 and b_2 are aligned to the vessel walls, the Doppler angle θ is defined by the angle between the ultrasound beam and the vessel's flow direction $\vec{v}$. The Doppler data are available in a polar coordinate system.

I. INTRODUCTION

Doppler sonography gained a widespread acceptance in medical practice in the last decade because it provides a non-invasive, convenient and cheap method for measurements of blood velocity and blood flow volume in vessels. Its application field reaches from cerebral hemodynamics [1] to echocardiography [2], gynecology and obstetrics [3]. As a special technique of Doppler sonography, pulsed-wave Doppler sonography has the advantage to provide Doppler shift data selectively from a vessel along the ultrasound beam. The Doppler gate, which specifies the sample volume, has to be aligned with the walls of the selected vessel. For the quantitative analysis, which is of fundamental importance for the assessment of hemodynamics, the selected vessel's diameter and the Doppler angle θ between the ultrasound beam and the direction of blood flow $\vec{v}$ within the vessel (see Fig. 1) has to be known [3]. With θ, the blood flow velocity v_{blood} can be calculated via

$$v_{blood} = \frac{v_{sound}}{2 f_t \cos\theta} f_d, \tag{1}$$

where f_t is the transmitted frequency, f_d is the Doppler shift, v_{sound} is the propagation speed of sound in the medium and θ is the Doppler angle.

Usually in current clinical systems the Doppler gate is determined by manually positioning the Doppler gate borders b_1 and b_2 to the vessel walls. The Doppler angle θ can be calculated after manually aligning a vessel axis marker along the vessel's flow direction $\vec{v}$. The operator has to select the Doppler gate and angle θ during real-time imaging with the one hand while holding the transducer with the other hand. This is a very time-consuming and inconvenient process, especially if adjustments in gate positioning or angle correction are necessary. Furthermore, clinical studies have proven, that the manual determination of the Doppler parameters constitutes a significant source of error and variability [4], [5]. The most critical factor that affects the accuracy of measurement is the Doppler angle θ. According to (1) the velocity measurement error increases exponentially with the angle error. E.g. with a Doppler angle of 60^o and an angle measurement error of $+5^o$, the velocity measurement error is approximatly 20% [3], [6].

An additional error source is the determination of the vessel's diameter which can be calculated from the Doppler gate size. Since this term is squared in the formula for calculating volume flow [3], any error in the determination is significantly magnified [6]. Errors in blood flow measurements may cause false diagnoses. Technical or computer assisted approaches can potentially improve accuracy, diminish variability and assure reproducibility [4].

Current solutions for determining θ are restricted to hardware-based or mechanical approaches. Ha and Hossak [7] use a transducer array to aquire a first cross-sectional Doppler data set of the vessel under examination. This array is then moved to a different angle to acquire a second cross-sectional Doppler data set. Then they calculate the true Doppler angle by using the known angle between the two arrays and the cosine scaled Doppler estimates.

In [8] Soustiel et al. use two ultrasound beams set at a fixed known angle and numerous small sample volumes are collected and analyzed by fast Fourier transform in real time.

The Doppler angle and the flow diameter are automatically computed allowing measurement of volumetric flow. This technique is at present limited to assessing superficial vessels such as the carotid arteries and is not available for maternal or fetal applications [3]. Experiments using a dual-beam vector Doppler system based on a modified linear array show that this system has low dependence on angle and similar reproducibility as single-beam systems [9]. These devices are also not yet available for clinical use in obstetrics.

In this paper we present a completely new approach to avoid errors in the measurement of blood flow. It is based on image processing and thus does not require technical modifications of the ultrasound devices. Our approach automatically determines the parameters b_1, b_2 and θ, the interaction of an operator being required only once in the beginning to select a target vessel V.

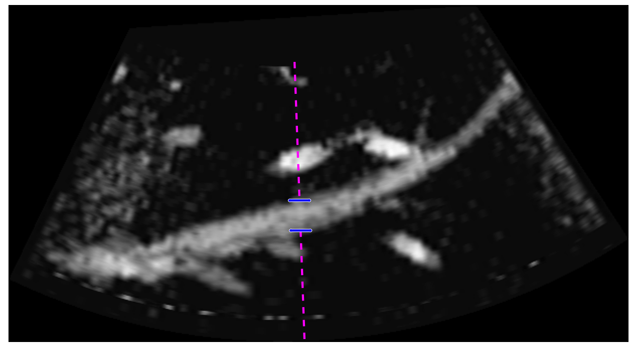
II. Methods

The methods we introduce in the following sections analyze 2D power Doppler flow data for determination of the Doppler parameters. The terminology of the data acquisition is as follows: the direction of a single ultrasound pulse defines the shot direction. The pulse echo of each shot is sampled resulting in n samples along every shot, n being constant. Several shots, originating from a common point O comprise an image. The direction of every shot is given by its shot angle σ, each sample is given by its radius r from O. This corresponds to a polar coordinate system (see Fig. 1). Thus the user-defined input point s has the coordinates $[r_s, \sigma_s]$. The input of s activates a fully automatic algorithm to determine the parameters.

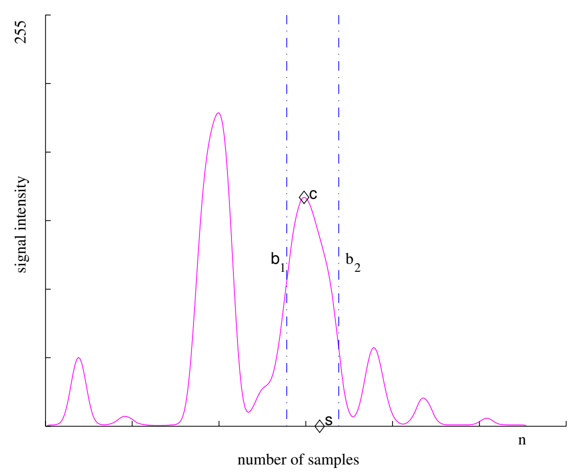
A. Doppler Gate Placement

The size of the Doppler gate is important for both the blood velocity measurement and the volumetric blood flow calculation. It determines the sample volume, i.e. how much of the vessel's blood flow will contribute to the velocity measurement. If the vessel is long and straight and the flow rate is constant, the distribution of flow velocities within the vessel diameter adopts an approximately parabolic profile. The velocity of the blood cells will be lowest near the vessel walls and greatest in the center of the vessel. The sample volume needs to be large enough to encompass the entire vascular cross-sectional area but has to exclude the surrounding structures. Thus, to obtain the mean velocity, the Doppler gate borders b_1 and b_2 have to be aligned correctly with the vessel walls [3]. If the vessel walls are known, the vessel's diameter can be calculated for the volumetric blood flow measurement.

In our automated Doppler gate placement method, the parameters c, b_1 and b_2 are determined by analyzing the echo intensity profile along the ultrasound beam defined by σ_s. In Fig. 2(a) an ultrasound beam is displayed and the corresponding intensity profile is plotted in Fig. 2(b). In power Doppler flow data the movement of blood is imaged bright and the background is dark. Considering the 1D intensity profile, all local maxima along the profile represent potential vessel centers. Thus the maximum next to r_s defines the center $c = [r_c, \sigma_c]$ of the target vessel V. Once, the gate center c is known, the gate borders can be determined in the next step. Starting with the gate center c, the adjacent local minima are calculated. Within this range an adaquate estimate for the vessel walls are the positions of extremal positive and negative first derivatives. Thus, the gate borders b_1 and b_2 are taken as these positions. In Fig. 2(a) and Fig. 2(b) the results of the automated Doppler gate placement are plotted.



(a) Ultrasound beam through the selected point s with detected gate borders b_1 and b_2.



(b) 1D intensity profile along the ultrasound beam.

Fig. 2. Result of the automated Doppler gate placement by analyzing the intensity profile along the ultrasound beam.

B. Doppler Angle Estimation

After the Doppler gate position is determined, we estimate the Doppler angle θ between the ultrasound beam σ_s and the flow direction $\vec{v}$ of the target vessel V. Therefore $\vec{v}$ has to be determined. We first consider a square region of interest (ROI). Reference point and center of this ROI is the Doppler gate center c. To achieve a reliable estimate of $\vec{v}$ we binarize the samples within the ROI using an adaptive threshold which is computed by the mean of all samples inside the ROI. The usage of such a local threshold for each ROI ensures an adaption to the local intensity mean.

Hence we obtain a set $\mathcal{V}$ of N samples ν belonging to the target vessel V. If we assume the positions of ν to be characterized by Cartesian coordinates, $\nu = \binom{x_1}{x_2}$, we can build a 2×2 covariance matrix C from the coordinates of all $\nu \in \mathcal{V}$, i.e.:

$$C = \sum_{i}^{N} (\nu_i - m) \cdot (\nu_i - m)^T, \tag{2}$$

m is the barycenter of $\mathcal{V}$. With a principal component analysis (PCA) of C, the main extension of V can be estimated as follows. One determines the eigenvalues λ_1 and λ_2, where $\lambda_1 \geq \lambda_2$, and the eigenvectors e_1 and e_2. The eigenvector e_1 corresponding to the largest eigenvalue λ_1 represents the direction with the largest variance in this local region and thus the flow direction of the target vessel.

For an efficient usage of the ROI, we have devised a multiscale approach which iteratively adapts the ROI according to the underlying vessel. The algorithm starts with a ROI of size $w \times w$ and performs the described procedure on the data within the ROI. We consider the directional estimate of the current scale reliable if the corresponding eigenvalue λ_1 is significantly greater than λ_2. If there is no reliable estimate based on the current ROI, the estimation is repeated with an enlarged ROI. Thereby the starting size of the ROI is of crucial importance for the algorithm. If it is too small, the algorithm needs too many iterations to yield a reliable estimation if the underlying data depicts a large cross section of a vessel. In the contrary, with a too large starting size the ROI may include surrounding vessels or bifurcations which make a reliable estimate impossible. According to this, an adaquate starting size for the ROI is the Doppler gate size. Thus the method permits an estimation of even a small vessel in a densely packed cluster of vessels. Once the flow direction vector of the target vessel is known, the Doppler angle θ between the ultrasound beam and the flow direction vector can be computed (see Fig. 3).

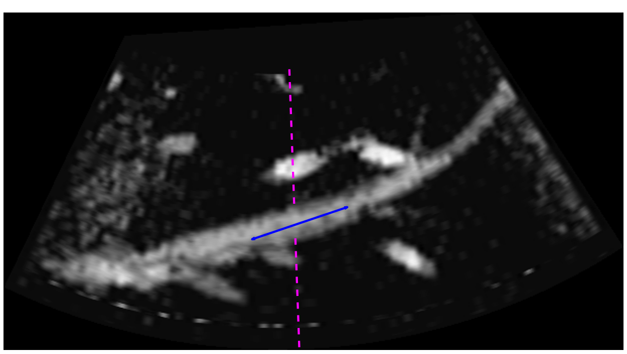


Fig. 3. Estimated flow direction of the target vessel.

Until now we assumed the positions of the samples were given in a Cartesian system. However, the underlying power Doppler data are available in polar coordinates (r, σ). The distances between the samples along an ultrasound beam (r) are constant, but the distances between the samples in lateral direction (σ) increase with the radius of the samples. To avoid a time-consuming resampling of the whole data to a Cartesian grid, we just transform the polar coordinates (r, σ) of the samples within the ROI by multiplying the σ coordinates with the (constant) radius r_c of the ROI center. The resulting space $[r, r_c\sigma]$ is still locally orthogonal and isometric to the Euclidean space for $r = r_c$ (the line passing through the ROI center). We can then apply the PCA to these transformed coordinates and obtain an angle which approximates the Doppler angle θ in the real world.

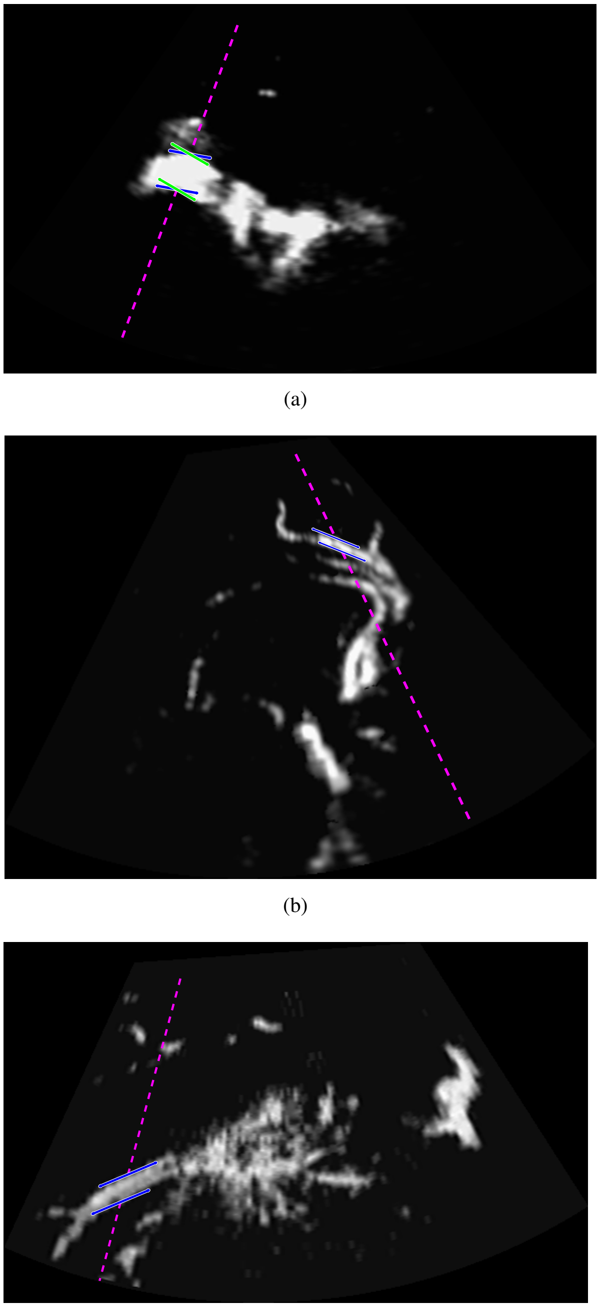
III. Results and Discussion

To get a first evaluation of our automated Doppler parameter approach we consulted two clinical experts. We chose 21 Doppler ultrasound images and set for each image an input point s. This point defined the target vessel V and the ultrasound beam σ. On these pre-defined positions, reference parameters θ, b_1, b_2 for each V were determined by discussion and subsequent agreement amongst the experts. We then performed our automated approach on all 21 images with the pre-defined input point s and compared the resulting parameters with the corresponding reference parameters. To evaluate the results we calculated the mean absolute difference Δ for each parameter type (Tab. I). Our

TABLE I

parameter	Doppler angle θ [^{o}]	gate border b_1 [pixels]	gate border b_2 [pixels]
mean abs. difference Δ	2.41	1.41	2.4

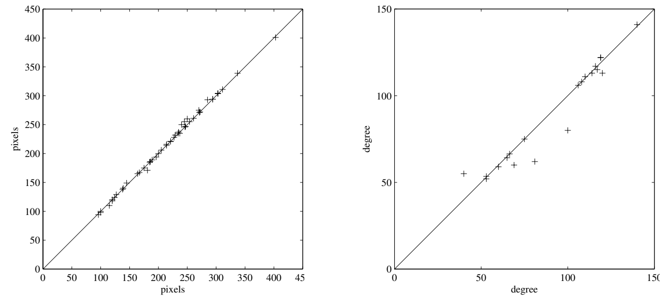
automated angle estimation yields $\Delta = 3.8^o$. In 16 cases, the automatically estimated Doppler angle diverged less than 5^o from the corresonding reference angle. However in three cases, the difference exceeded 15^o. Because of this large deviation we investigated these outliers in detail. In all three images, the target vessel had a slightly curved structure and the pre-defined ultrasound beam ran through the center of the curve. As an example we show one of these problematic cases in Fig. 4(a). Considering the left part of the vessel, our automated approach is correct, but the experts considered the main right part. Thus the large difference of 20^o can be explained. Excluding these outliers, our automated angle estimation yields $\Delta = 2.41^o$. Both for the gate border b_1 and for the gate border b_2 our automated approach reaches nearly in every case the reference parameters of the experts. In Fig. 4, two examples are shown, where our automated approach yields as well the reference θ as the reference gate borders b_1 and b_2. Fig. 4(b) additionally demonstrates the excellent performance of the multiscale approach on densely packed clusters of vessels. Fig. 5 shows scatter-plots of Doppler gate positions and Doppler angles determined by the experts versus the automatically determined parameters. * This first evaluation thus indicates that our automated approach yields parameters which match parameters manually determined by two clinical experts very well. Of course, there are still open issues concerning the operator's interaction. E.g. the three



(a) (b) (c)

Fig. 4. Results for automated Doppler gate placement and flow direction estimation. (a) shows one of the problematic cases with a angle difference of 20^o. The bright axes define the reference flow direction, the dark axes define the automatically determined flow direction. In (b) and (c) the automated approach yields as well the reference θ as the reference gate borders b_1 and b_2.

discussed problematic cases demonstrate that the selection of the input point s is of crucial importance for a reliable estimation of the flow direction. To yield the best results, s



(a) (b)

Fig. 5. Scatter-plots of Doppler gate positions (a) and Doppler angles (b) determined by the experts versus the automatically determined parameters.

has to be selected on a vessel part with a distinct direction. Furthermore, clinical studies have to show how "local" the estimation of the flow direction should be to obtain good flow measurements. According to this, the automatization can be adapted. Hence our approach could potentially assist examiners in clinical routine by replacing time-consuming manual selection of the Doppler parameters by just clicking on a vessel to obtaining reliable parameters.

IV. Acknowledgments

The authors would like to acknowledge the contributions of Armin Schoisswohl of GE Healthcare Kretztechnik in providing the 2DUS data sets and valuable discussions. This work has been carried out within the K*plus* Competence Center ADVANCED COMPUTER VISION. This work was funded by the K*plus* program.

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