Automated Blood Volume Quantification from Color Doppler Images while Tracking the Conduit Motion

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Abstract

Previous studies have shown that the quantification of volumetric blood flow from a color Doppler image (CDI) is a feasible approach to eliminate technical obstacles in the traditional pulsed wave (PW) Doppler method. This paper introduces an automated method that computes the volumetric blood flow through a conduit while tracking the movement of blood flow. Correct calculation of the blood flow vector will reduce the conduit overestimation caused by the axis misalignment between the actual and estimated blood flow direction. This correction immediately contributes to the improvement of measurement accuracy. A clinical evaluation of the developed technique was executed on 21 healthy subjects and 10 patients. The proposed method generated better measurement results than the existing technique when the outcomes were compared with blood volumes measured by the PW-Doppler method.

Keywords: Blood volume quantification, Color Doppler, Echocardiography, Angle-correction

1. Introduction

Stroke volume (SV) represents the amount of blood pumped from the heart into the body in one heart beat. This blood flow measurement provides clinically useful information for evaluating cardiac function – including left ventricular disorder, shunt flows, and valvular abnormalities [1, 2].

The PW-Doppler technique is the most widely used echocardiographic method for measuring blood volume through the conduit. The method computes the SV by multiplying the conduit diameter to the blood velocity profile through the conduit over an ejection [1, 3]. This clinically validated technique, however, includes incorrect assumptions that assume a uniform velocity profile across the blood flow and constant conduit diameter during an ejection [2, 4].

To correct these hypotheses, researchers have developed methods for computing SV from a composite image of the left ventricular outflow track (LVOT) [5, 6]. Volumetric flow is computed from CDIs by spatially and temporally integrating blood velocities encoded in each image throughout the ejection period. These methods attempt to reduce the consequence of false assumptions by using the Doppler velocity information encoded in each image. Furthermore, recent studies present techniques computing the SV with angle-corrected blood velocities. The conduit orientation in the heart and radial ultrasound beams forms an angle with the direction of blood flow. These global and local angle differences cause underestimation of the blood velocity resulting in underestimation of the SV [7-9].

Although newly developed methods correct the underestimated blood velocity caused by the angle differences, those methods compute the volumetric blood flow using a representative blood flow vector over an ejection period. The heart motion over the systole causes the translation of the LVOT, and thus variations of the blood flow direction. Any discrepancy between the actual blood flow direction and computed flow vector causes overestimation of the conduit diameter that leads to the overestimated SV.

This paper introduces a technique that computes the SV with multiple blood flow vectors. By defining blood flow directions for different time intervals, we can measure the SV while reflecting the motion of the LVOT over a systole. SVs measured by the proposed method were compared with those computed by a single blood flow direction and also with those computed by the PW-Doppler technique.

2. Computing Volumetric Blood Flow

In order to measure the SV from a CDI without human intervention, a series of tasks – identification of the ejection period, measurement of the conduit diameter, determination of blood flow directions, angle correction of the blood velocity, and computation of the SV – were implemented automatically.

2.1 Determination of the Blood Flow Direction

The ideal way to reflect the shift of LVOT for computing the SV will be determining the direction of blood flow separately for each image. But the nature of the CDI – low resolution and noise along the conduit border – makes it difficult to define the blood flow vector from an individual image. This is especially true of the images near the start and end of ejection where the conduit is partially filled with blood flow. To address this difficulty, the proposed method computes a representative blood flow direction based on a composite image that is created by multiple images within a time domain.

The images in a systole are divided into three different time intervals. The first time domain consists of images from the start of ejection (f_{start}) to the middle of ejection (f_{mid}). The second time interval includes frames from f_{mid} to the frame at the end of ejection (f_{end}). The other time domain is defined by frames within $f_{mid} \pm ((f_{end} - f_{start}) / 4)$.

To define a blood flow vector representing the direction of multiple frames, a composite image is constructed by adding together the blood velocity at each point in the scanner's region of interest (ROI) of images within the same time interval. In summing the velocities, movements toward the transducer are excluded to eliminate noise motion. Note that during an ejection, blood in the LVOT moves away from the transducer and is recorded as a negative velocity. By dividing each velocity sum by the number of images of the associated time domain, each point in the resulting composite image represents the average blood velocity at the corresponding location. This process can be formally expressed as

$$C(i, j) = \sum_{f_k \in t_m} V(i, j, k) / N_m \qquad (1)$$

where $C(i, j)$ is the blood velocity of composite image at (i, j), t_m represents a time domain, N_m is the number of images in t_m, $V(i, j, k)$ is the blood velocity at (i, j) in frame f_k where $V(i, j, k) = 0$ if $V(i, j, k) \geq 0$.

Each pixel $C(i, j)$ in the composite image includes a variety of movements within the ROI. A thresholding technique is applied to the composite image to identify the primary region of flow from noise movements. Unlike other image segmentation problems, it is not a critical issue to find an optimal threshold value to segment the primary blood flow. The blood flowing through a cylindrically-shaped conduit tends to form a series of concentric layers and thus, the blood flow through the LVOT will be symmetrical with respect to the conduit's centerline [10]. Sub-optimal threshold values will symmetrically eliminate velocities from the outer layers of the conduit.

An iterative threshold selection algorithm is used to define a threshold value to separate the primary flow in the composite image [11]. This method segments an image into two classes (object and background) by successively refining the threshold value. A threshold value for iteration is computed from the velocity histogram of the composite image as follows:

$$T_{k+1} = \left(\sum_{b=0}^{T_k} X \cdot n(X) / 2 \cdot \sum_{b=0}^{T_k} n(X) \right) + \left(\sum_{b=T_{k+1}}^{N} X \cdot n(X) / 2 \cdot \sum_{b=T_{k+1}}^{N} n(X) \right) \qquad (2)$$

where T_k is the threshold at the kth iteration, X is a value of the x-axis of the histogram, $n(X)$ is the number of pixels with value X for $0 \leq b \leq N$, and N is the maximum value in the histogram [11]. By applying the thresholding process, the composite image is transformed into a binary image. When a pixel's blood velocity is greater than the threshold value, the pixel is mapped to the flow region. Otherwise, the pixel is set to a non-flow area. The thresholding generates small isolated areas along the conduit border. Execution of region-growing algorithm eliminates extraneous regions from the binary image except a blood flowing area through the LVOT.

We detect two end-points of blood flowing area along each scanline of the binary image. A pair of end-points defines a midpoint of a scanline that will be located along the centerline of the LVOT. By fitting a line to these midpoints using the method of least squares, a line (L^{-1}) that is parallel to the direction of blood flow through the LVOT having the slope of m^{-1} is computed. The line, L^{-1}, represents the direction of blood flow associated with the selected time interval. A line (L) that has the inverse of the slope of m^{-1} serves as the perpendicular axis to the blood flow. This process is repeated to the composite images created from other time domains.

2.2 Identification of Conduit Boundary

The edge of LVOT conduit is identified by analyzing the blood velocity profile along the perpendicular axis, L. The blood velocity of two adjacent locations at the boundary of the tissue and the LVOT show significant differences in terms of magnitude and direction. Beginning at the intersection of the lines L^{-1} and L (p_{mid}), we walk outward along line L computing velocity gradients between each pair of adjacent points ($g_i = v_{mid-i} - v_{mid-i-1}$) where v_{mid-i} is the blood velocity of a location at p_{mid-i}. To detect the boundary locations of the LVOT, we execute a series of tests from the location having the largest gradient in descending order. The first point that satisfies all the following conditions is defined as the left edge of LVOT (p_l). The conditions are: $g_i < 0$; $v_{mid-i-1} \geq q_1$; all the blood velocity of locations between p_{mid} and p_l is smaller than q_2 where the threshold q_1 and q_2 were determined experimentally [6]. Remember the blood velocity within the LVOT shows a high negative value, while the movement of tissue has either positive, zero, or fairly low negative velocity. The right edge of the LVOT is identified in the similar way in opposite direction. This process is then repeated for each image in CDI to account for variations in the conduit diameter and for the translation of conduit within the image plane during the cardiac cycle.

2.3 Angle Correction of the Blood Velocity

Because of the nature of conduit location and orientation in the heart, the ultrasound beam commonly forms an angle with the direction of blood flow. The blood velocity underestimation caused by the angle differences between the direction of blood flow and the rays of the ultrasound beam was corrected. After locating the ultrasound beam origin (x_o, y_o) in CDI, a slope m_0 connecting two points between (x_o, y_o) and a point at (x_i, y_i) on the perpendicular axis, L, is computed (Figure 1). Under radial ultrasound beams, the angular difference (q) between the blood flow vector with slope m and m_0 can be expressed in terms of slopes m and m_0 as

$$\begin{aligned} tanq &= tan(a - b) \\ &= (tana - tanb) / (1 + tana * tanb) \\ &= (m - m_0) / (1 + m * m_0) \end{aligned} \tag{3}$$

$$q = atan(m - m_0) / (1 + m * m_0) \tag{4}$$

where $m \geq m_0 \geq 0$ and if $m * m_0 = -1$ then $q = \pi / 2$. The angle-corrected velocity (v_a) is

$$\begin{aligned} v_a &= v_m / cosq \\ &= v_m / (cos\,(atan\,(m - m_0) / (1 + m * m_0))) \end{aligned} \tag{5}$$

where v_m is measured velocity.

2.4 Stroke Volume Computation

The CDI is recorded with the imaging plane bisecting the LVOT parallel to the blood flow. The blood flow through the LVOT tends to form a series of concentric layers, and the velocity within each layer is relatively uniform [10]. The volumetric blood flow per unit time through the LVOT can be computed by the product of the cross-sectional area of the LVOT and the mean blood flow velocity per unit time [1].

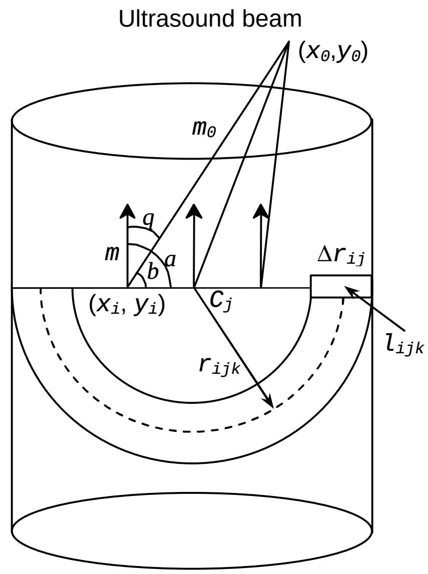

Figure 1. Angle-correction and SV Computation.

The conduit diameter is divided into n line segments of equal length. These line segments are rotated 180 degrees clockwise about the center point (C_j) to form a cross-sectional area consisting of a set of multiple semi-circular rings. Assuming a uniform blood velocity within each half of the layer, each point velocity within the LVOT represents the mean velocity for the corresponding semi-circular area. The flow rate through a semi-circular area derived from a line segment l_{ijk} is $r_{ijk} = (p \cdot \Delta r_{ij}) \cdot r_{ijk} \cdot v_{ijk}$ where Δr_{ij} is the width of line segment l_{ijk} and r_{ijk} is the distance from the center of line segment l_{ijk} to the center of the conduit C_j, v_{ijk} is the angle-corrected mean blood velocity of l_{ijk} (Figure 1). Integrating the instantaneous blood flow rates of the semi-circular rings yields the instantaneous flow rate (r_{jk}) for frame k. Multiplying the r_{jk} by the corresponding frame duration (t_k) yields the ejection volume for frame f_k. Finally, summing the ejection volumes over the ejection period gives the SV measured along sampling line s_j

$$cfsv_j = \sum_{f_k = f_{start}}^{f_{end}} ((p \cdot \Delta r_{ij}) \sum_{i=1}^{n} r_{ijk} \cdot v_{ijk}) \cdot t_k \tag{6}$$

3. Experimental Environments

To investigate the accuracy of the proposed algorithm, a clinical evaluation was executed on a total of 93 CDIs acquired from two different populations (1) 21 healthy male volunteers without clinical evidence of heart disease (2) 10 patients with clinical evidence of heart disease. The study protocol was approved by the Human Investigation Committee of Rush-Presbyterian-St. Luke's Medical Center.

Transthoracic echocardiograms were performed on all subjects using Hewlett-Packard 2500 cardiac ultrasound system. CDIs were recorded at the LVOT level sequentially for each subject from the apical long-axis view. Settings for CDI acquisition were: black/white suppression set to obtain maximal frame rate, minimal clutter filter to maximize low velocity information, and maximal smoothing to decrease the amount of color noise.

For each study, the SVs were computed from multiple locations along the slope of the blood flow m^{-1} to reduce the impact of measurement variations. The average SV within top 5% was computed as the SV for a CDI. The maximum SV among captured CDIs of each subject was reported as the subject's color flow stroke volume (CFSV). By choosing the maximal SV, we avoid SVs that are computed from locations which have significant aliasing or show underestimated conduit diameter when the imaging plane lies at an angle to the center plane of LVOT.

4. Results

The table below presents measurement results computed by three different methods. CFSV computations were performed in two ways using (1) a representative blood flow vector for an ejection (S-CFSV) (2) three flow vectors over a systole (M-CFSV). These results were compared to the SV computed by the PW-Doppler technique (PW).

Table 1. Evaluation Results.

	21 Healthy Subjects		
	PW	S-CFSV	M-CFSV
Mean ± SD (ml)	83.9±10.6	82.3±15.8	81.1±15.7
Bias ± SD (ml)	-	-1.6±10.4	-2.8±8.8

	10 PATIENTS		
	PW	S-CFSV	M-CFSV
Mean ± SD (ml)	81.5±10.8	89.9±21.2	85.7±19.1
Bias ± SD (ml)	-	8.4±12.8	4.2±11.6

Mean: Mean stroke volume SD: Standard deviation
Bias: Mean stroke volume difference between CFSV and PW-Doppler

5. Discussion

The SV computation reflecting the movement of blood flow direction improves measurement accuracy. In order to analyze the effect of the number of blood flow vectors on the measurement accuracy, the SV was computed with the blood flow direction per image. The blood flow vectors for each time domain are linearly interpolated to estimate the flow direction at each frame. The mean SV difference between CFSV with three blood flow vectors and the PW-Doppler was -0.53±10.16 ml. Meanwhile, the bias between CFSV with interpolated blood flow vectors and the PW-Doppler was -0.39±10.31 ml. Both methods produced nearly the same measurement results. This can be caused by not considering the amount of blood flow for each frame. As recorded on the PW-spectral Doppler image, the amount of blood flow for each frame is different, especially there is rapid blood volume change before reaching the peak ejection.

The developed system eliminates the blood flow vector that is significantly different from the slope computed by the images around the peak ejection. The post-experimental study reveals that there were noticeable differences between the blood flow vector by S-CFSV and flow directions by M-CFSV during some time internals. Frame-by-frame analysis shows that the LVOT conduit is partially filled with blood flow around the beginning and end of ejection. In addition, some CDI of cardiac patients shows improper blood flow caused by cardiac problems and blood velocity aliasing. These factors cause the difficulty in defining the direction of blood flow. Further research is required to estimate the flow vector from the images showing limited blood flow.

It is an important and difficult task to capture the CDI from the image plane that bisects the LVOT. When the CDI is recorded at a skew angle to the LVOT's center plane or from the plane located away from the center of the LVOT, both the conduit diameter and blood velocity will be underestimated. Post-experimental analysis shows that the significantly underestimated SVs of subjects were caused from the improper image-capturing locations. In the future, more systematic investigation about finding the most proper image-recording plane will be required.

6. Conclusions

This study introduces a fully-automated method for quantifying the volumetric blood flow from color Doppler images. The developed method measures the blood volume by computing multiple blood flow vectors during an ejection period. This approach reduces the blood volume overestimation cased by the transition of the LVOT. The clinical validation shows a close agreement between SV measured by the proposed method and that determined by the PW-Doppler technique. The proposed method improves the measurement accuracy compared to the quantification technique using single blood flow direction. This study also suggests several findings for improving further the measurement accuracy.

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