Automated cardiac flow measurement by digital colour Doppler echocardiography

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Quantification of cardiac flow volumes, such as stroke volumes and cardiac outputs, has had a major impact on the management of patients with cardiac diseases.

In the 1980s, the development of Doppler echocardiography, especially pulsed wave Doppler techniques, provided significant progress in determining cardiac flow volumes. The pulsed wave Doppler method is based on a simple mathematical principle, which multiplies a flow area by its velocity during the flow time (velocity–time integral) to estimate a flow volume. This Doppler method assumes a spatially flat flow velocity distribution and a constant flow area during the entire time of flow. However, the spatial velocity profile is not necessarily flat and the flow area may change during the time of flow. In the 1990s, a digitally automated cardiac flow measurement method was developed to overcome these shortcomings of the conventional pulsed Doppler method. This new method uses not only a single central point but also multiple Doppler points in space along the diameter of the flow; it also uses several imaging frames during the time of flow, accounting for the temporal change in the flow area (that is, not assuming a constant flow area). In addition, the calculation does not require manual tracing of the Doppler velocity nor the measurement of flow diameter. Thus, the actual use of this automated method in clinical settings is simpler and easier than the conventional pulsed Doppler method.

**PRINCIPLE OF AUTOMATED CARDIAC FLOW MEASUREMENT**

When we try to calculate a cardiac flow volume (for example, stroke volume) from a patient, we only need to image colour Doppler flow through a corresponding flow tract (the left ventricular outflow tract in a five chamber view or an apical long axis view for stroke volume). Then, for calculating a flow volume, a region of interest is selected, as seen in fig. 1. The size of the region of interest is typically 33 mm in width and 5 mm in depth, but it can be altered to cover the entire flow field. Each region of interest includes five discrete velocity profiles sequential in depth, parallel to each other across the diameter, each with a thickness of 1 mm if we select the size of the region of interest at 5 mm in depth. Each discrete velocity profile includes 115 Doppler velocity data points across if the width is selected to bed the 33 mm. These data points are derived from 42.6 scan lines at the first profile closest to the transducer and 38.1 scan lines at the fifth furthest profile using the computed interpolation.

The flow rate (not volume) from each one of the five velocity profiles is calculated by integrating the Doppler interrogated velocity across the flow diameter in digital cine memory, assuming a half circular symmetrical flow distribution for each radius from the centre of the flow. As seen in fig 2, the actual flow rate (Q) is equal to the product of the velocity of the flow (V), which is parallel to the direction of the vessel, multiplied by the area (S) which is the area of flow perpendicular to the vessel. This flow rate is also equal to the product of velocity Doppler determined velocity (V Doppler = V cosø) and the corresponding area of flow (SDoppler = S/cosø) perpendicular to the Doppler determined velocity; flow rate is calculated by using Doppler determined velocities = V cosø S/cosø = V S = Q. Any obliquity to the flow direction increases flow cross sectional area in proportion to the decrease in computed velocity. The five different flow rates from the five sequential velocity profile rows in the depth domain of each colour frame are then averaged to determine the representative flow rate. The calculation of the flow rate described above is performed consecutively for each frame during the time of flow. This way, the computer assisted method accounts for temporal changes in flow area during the period of flow. All of these flow rates during the selected time of flow are added together and multiplied by the time interval of each frame to obtain flow volume.

**TECHNICAL PROBLEMS AND THEIR SOLUTIONS**

This method, as attractive as it may sound, is not yet widely accepted or used in clinical settings. One major problem seems to be the availability of this method itself. So far, this colour Doppler software is equipped with ultrasound systems from a Japanese company (Toshiba Co, Tokyo, Japan). No other cardiac ultrasound companies have implemented this software capability in their ultrasound systems. There are some technical and methodological limitations, however.

First, the colour gain setting significantly affects the accuracy of calculation as in any other colour Doppler method. Therefore, one must be very careful to obtain appropriate colour signals—that is, not having any colour in the area of the non-flow area nor having black velocity points (namely zero velocity) on the area of the actual blood flow. Another important technical point is the velocity filter. When the Doppler velocity filter is set too high, especially for low velocity flows, one may underestimate the flow volume.
and the Doppler interrogation if it is more than 30°. One can needs to be corrected for the angle between the flow direction obliquity is oval shaped. Thus, the flow measurement still of the flow). However, the actual area of flow imaged with flow area (actually a half circle for each radius from the centre system uses only one axial dimension (r), assuming a circular area in proportion to the decrease in computed velocity, the obliquity to the flow direction increases flow cross sectional underestimation. Although as explained above (fig 2), any obliquity of the Figure 2 Principle of the digitally automated cardiac flow measurement (see text for detail). In essence, any obliquity of the Doppler interrogation to the flow direction increases flow cross sectional area in proportion to the decrease in computed velocity, cancelling each other. Appropriate filter setting should be chosen to avoid any underestimation. Although as explained above (fig 2), any obliquity to the flow direction increases flow cross sectional area in proportion to the decrease in computed velocity, the system uses only one axial dimension (r), assuming a circular flow area (actually a half circle for each radius from the centre of the flow). However, the actual area of flow imaged with obliquity is oval shaped. Thus, the flow measurement still needs to be corrected for the angle between the flow direction and the Doppler interrogation if it is more than 30°. One can perform this by turning the angle correction knob equipped within the ultrasound system. Also, this method can be applied properly only to circular vessels because it assumes a half circular axi-symmetric flow velocity profile. Thus, this method may not be accurate for measuring blood flows through asymmetrical areas. Use of two orthogonal planes may be recommended to calculate flow volumes through the mitral annulus and even the pulmonary artery. Finally, because it is impossible to assign proper colour flow velocities in aliased areas, one should use on-line baseline shift to avoid aliasing, especially for high velocity flows.

CLINICAL APPLICATIONS
When one is attentive to the above mentioned limitations of this method, it is possible to apply it clinically. Several in vitro and in vivo studies have confirmed its validity for determining cardiac flow volumes, especially stroke volumes. Experts applied this method clinically as well, in patients with mitral regurgitation, aortic regurgitation, postoperative tetralogy of Fallot, and atrial septal defect. Based on these studies, this new digital colour Doppler method has been recognised as a means of determining regurgitant volumes such as mitral, aortic, and pulmonary regurgitant volumes and shunt flow volumes clinically. In my opinion, this method has a great potential for clinical use, when appropriate steps are taken as discussed, especially for determining stroke volumes and cardiac outputs.

REFERENCES