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CARDIOLOGIC PARAMETERS EXTRACTION FROM ECHOCARDIOGRAPHIC IMAGES USING VIRTUAL INSTRUMENTATION

Mihai-Dan STERIU*, Nicolae PAUN, Josef FRIEDMANN*****

**Electronics and Telecommunications Faculty, "POLITEHNICA" University of Bucharest*
steriu@golana.pub.ro, dan.steriu@dce.pub.ro
***University of Medicine and Pharmacy "Carol Davila"-"CARITAS" Hospital, Clinic of Cardiology, Bucharest*
****Center for Microscopy – Microanalysis and Information Processing, "POLITEHNICA" University of Bucharest*

Abstract. *Color Doppler M-mode recordings are digitally post processed. A very simple method for extracting blood flow velocities from RGB components of echographic images, stored in .bmp files, is proposed. Instantaneous pressure distribution is obtained by solving Navier-Stokes equation. Different 3D and 2D representations are used for extracting cardiological parameters like: minimum of pressure at base and apex, $P_{min\text{-}base}$, $P_{min\text{-}apex}$, intraventricular pressure gradient between apex and base, $G_{apex\text{-}base}$, maximum velocity delay, TIV_{max} and minimum pressure delay TIP_{min} between apex and base, TAV_{max} and TAP_{min} , times for maximum velocity and minimum pressure at base. A simple patient data base and all images processing was made by a program developed in LabView 6.1 (National Instruments)*

Key Words: *Color Doppler Ultrasound, Image Processing, Virtual instrumentation*

1. Introduction

Left ventricular diastolic dysfunction, the inability of the left heart to fill with blood at an acceptable low pressure, is an important cause of cardiac morbidity and mortality. To avoid the inconvenience and the risk of invasive measurements, non-invasive approaches such as Doppler echocardiographic characterization of ventricular filling have been explored. Color M-mode Doppler (figure1) is a pulsed Doppler technique that allows obtaining a spatio-temporal blood velocity map. A complete picture of the filling pattern can be acquired. The velocity of a discrete blood sample is provided by the value of its corresponding image pixel. Some clinical parameters can be extracted from ***v(s,t)*** distribution, but the main information comes from temporal and spatial fluctuations of pressure. The present study proposes a methodology to map accurately velocity and pressure during the filling of the left ventricle (**LV**).

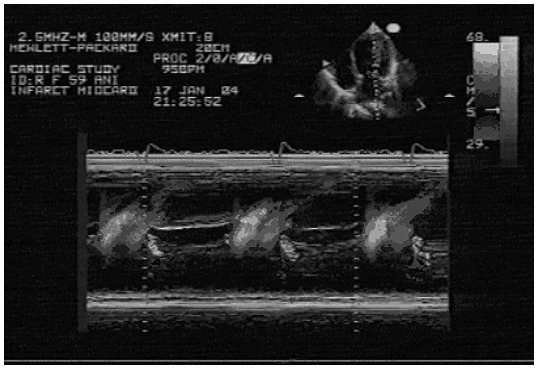

Figure 1- M-mode Doppler Color echocardiography

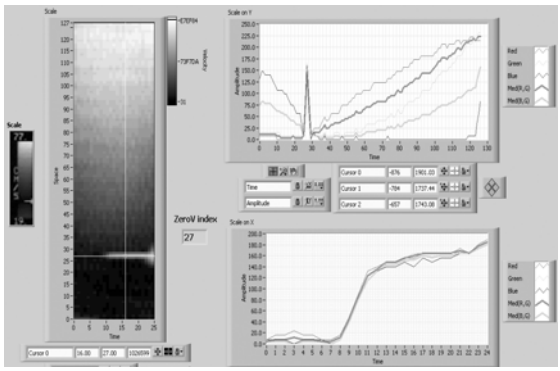

Figure 2 - Velocity scale analysis

2. Velocity decoding and fitting

Velocity decoding (RGB pixel -> velocity value) is made using very simple relations established by scale analysis (figure 2 and 3). Velocities toward the transducer (positives) are red/yellow, and velocities from the transducer (negatives) are blue/cyan (figure 3):

$$v(i,j) = \frac{\left[R(i,j) + G(i,j)\right] \cdot v_{Max}}{2 \cdot RG_{Max}} \quad , for\ R(i,j) > G(i,j)$$

$$v(i,j) = \frac{\left[B(i,j) + G(i,j)\right] \cdot v_{\min}}{2 \cdot BG_{Max}} \quad , for\ R(i,j) \le G(i,j)$$

where v_{Max} and v_{min} are extremities of the velocity scale, RG_{Max} is the maximum of ***Average(R,G)*** and BG_{Max} is the maximum of ***Average(B,G)*** – figure 3. A very good linearity is observed (figure 4)

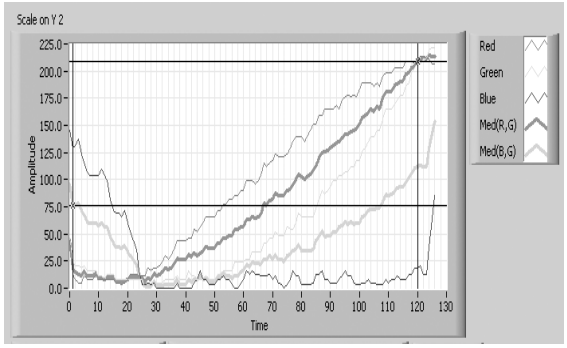

Figure 3 - RGB components of velocity scale

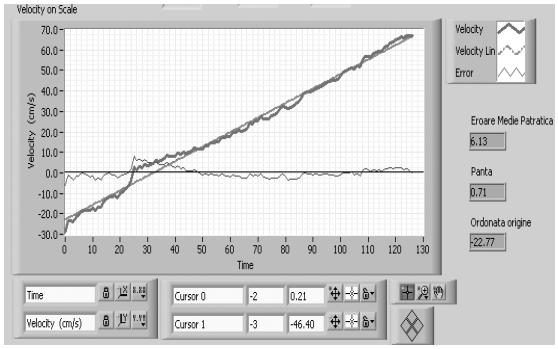

Figure 4 – Linear fitting of velocity

Due to the lack of flow (no Doppler phase-shift), ultrasound drop-out (no Doppler signal), transport of information on different supports (VCR) results in a very noisy representation of the velocity (figure 5). This renders the simple finite-differences method and velocity-linked parameters extraction highly inaccurate. Digital image filtering (figure 6):

$$v_m = \frac{\sum_{i-n\le k\le i+n} \sum_{j-n\le p\le j+n} v(k,p)}{(2n+1)^2}$$

and smoothing by polynomial fitting (figure 7) are two alternatives to improve accuracy. In both cases the results are very close.

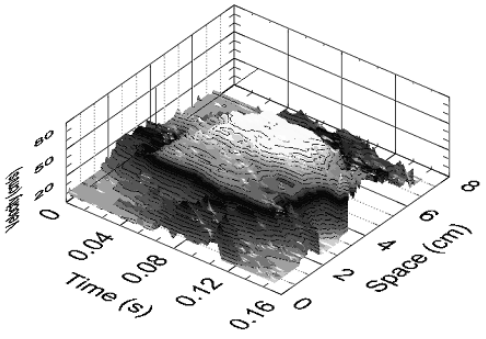

Figure 5 –Directly extracted v(s,t)

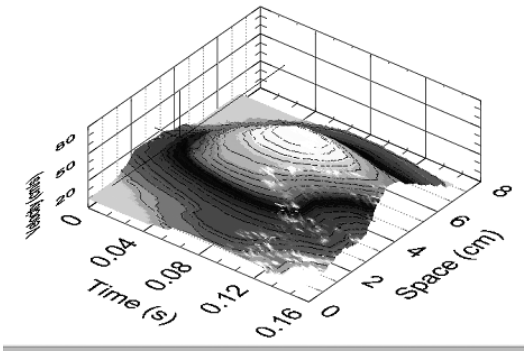

Figure 6.- Filtered v(s,t)

3. Fluid dynamics of diastolic intraventricular flow

Relationship between spatio – temporal mapping of intracardiac pressure, ***p(x,y,z,t)*** and blood velocity ***v(x,y,z,t)*** is governed by *Navier-Stokes* equation:

$$-\nabla\left(\frac{p}{\rho}+g\cdot z\right)+\left(\frac{\eta}{\rho}\right)\cdot\left[\nabla\times\left(\nabla\times v\right)\right]=\frac{\partial v}{\partial t}+\left(v\cdot\nabla\right)v$$

where ***g*** is the gravitational acceleration, z is the local vertical position, and, ***ρ*** and ***η*** are the blood density and the dynamic viscosity. The viscous term and hydrostatic contribution are very small and if the flow is assumed to follow a single spatial direction (s) represented by ultrasonic streamline, then the *Navier-Stokes* equation becomes *Euler's* equation:

$$\frac{\partial p}{\partial s}=-\rho\left[\frac{\partial \mathrm{v}}{\partial \mathrm{t}}+v\frac{\partial v}{\partial s}\right]$$

The pressure gradient is computed from the sum of the convective ($-\rho v \frac{\partial v}{\partial s}$) and inertial ($-\rho \frac{\partial v}{\partial t}$) components. Then, the pressure, reported at a reference point (ex. **base**) is given by:

$$\Delta P[t,s] = \int_s^{base} \frac{\partial p}{\partial u} du$$

We have used two methods to solve *Euler's* equation. The first one is fully numerical and it is based on the evaluation of partial derivatives by convolving ***v(s,t)*** with Sobel operators. The second one is based on the polynomial fit of ***v(t)*** at distance ***s***:

$$v(t)\big|_s = \sum_{k=0}^{m} a_k \cdot t^k$$

and on the formal integration of the convective term and the numerical integration of the acceleration (results in figure 8):

$$\Delta P[t,s] = \rho \left(\frac{v_s^2 - v_{base}^2}{2} + \int_s^{base} \frac{\partial v}{\partial t} ds \right)$$

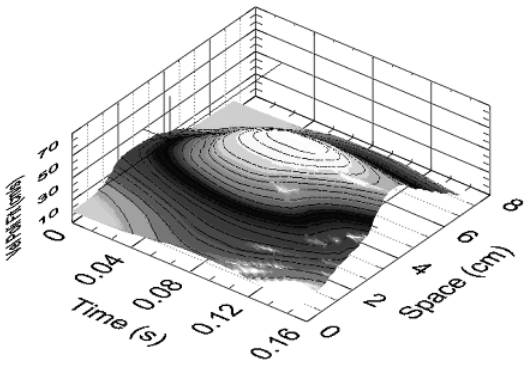

Figure 7 – Polynomial fit of velocity

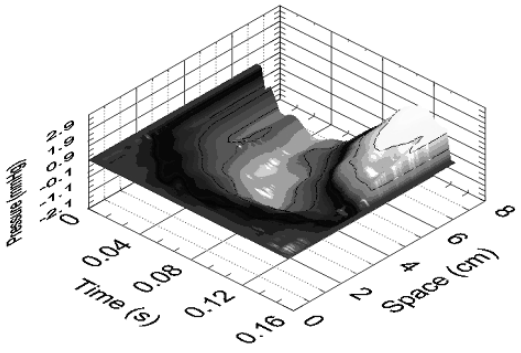

Figure 8 – Pressure distribution

4. Results

It was possible to extract some echographic specific parameters from different 2-D sections, by using spatio-temporal maps of blood velocity and pressure. The differences between healthy subjects and patients with diastolic dysfunctions can be highlighted.

Patient with diastolic dysfunction (figure 9): Delay of V_{max} between base and apex (cursor 0-cursor1), TIV_{max}*=0,079s*. Time for V_{max} at base (cursor 1), TAV_{max}*=0,081s*.

Healthy subject (figure 10): Delay of V_{max} between base and apex (cursor 0-cursor1), $TIV_{max}=0,037s$. Time for V_{max} at base (cursor 1), $TAV_{max}=0,0071s$.

Patient with diastolic dysfunction (figure 11) Minimum pressure at base, $P_{min\text{-}bsza} = -0,74\ mmHg$. Minimum pressure at apex, $P_{min\text{-}apex} = -1,07\ mmHg$.

Healthy subject (figure 12) Minimum pressure at base, $P_{min\text{-}bsza} = -1,40\ mmHg$, Minimum pressure at apex, $P_{min\text{-}apex} = -3,10\ mmHg$.

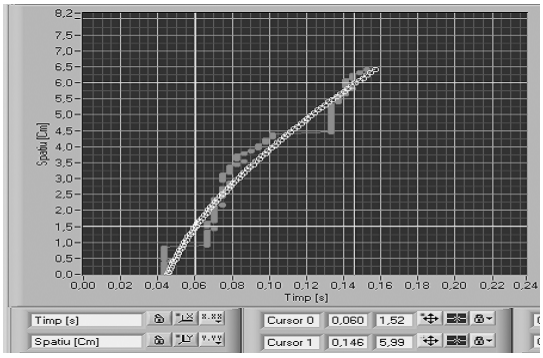

Figure 9

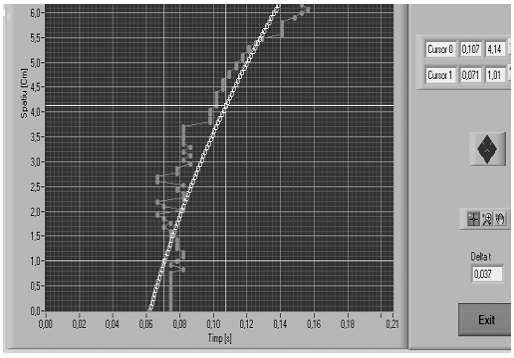

Figure 10

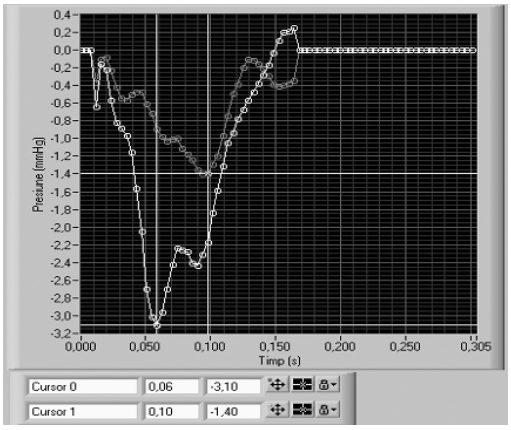

Figure 11

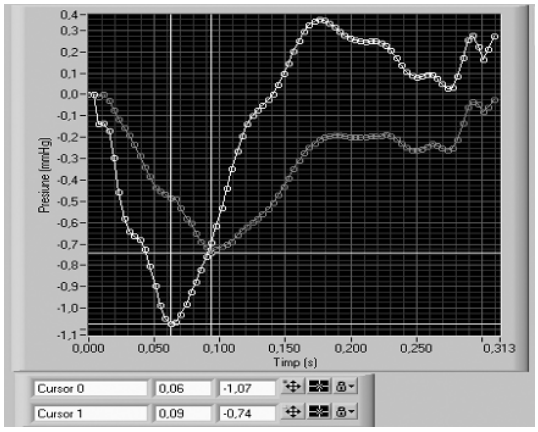

Figure 12

Patient with diastolic dysfunction (figure 13): Delay of P_{min} between base and apex (cursor 1-cursor0) , TIP_{min}= 0,086 s. Time for P_{min} at base (cursor 0), TAP_{min}= 0,060 s.

Healthy subject (figure 14): Delay of P_{min} between base and apex (cursor 1-cursor0), TIPmin=0,036s. Time for P_{min} at base (cursor 0), TAPmin=0,051s.

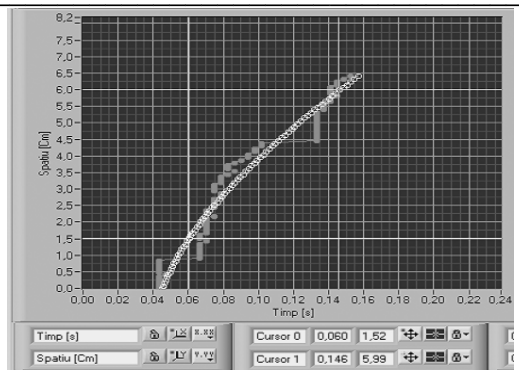

Figure 13

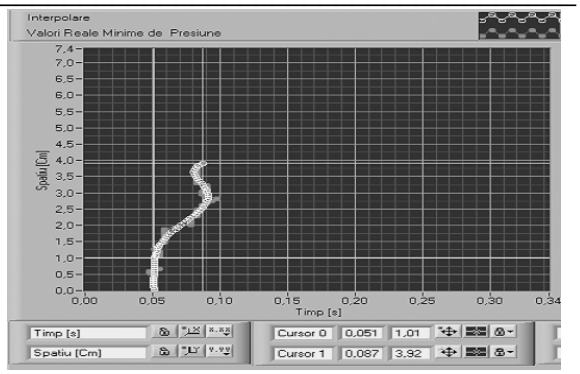

Figure 14

5. Conclusions

Virtual Instrumentation makes possible to provide unique information on **LV** filling dynamics using an entirely noninvasive method. This method reduces medical care costs and makes this kind of investigation absolutely non-traumatical.

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