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Original Research Article

Strain examinations of the left ventricle phantom by ultrasound and multislices computed tomography imaging

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ABSTRACT

The main aim of this study was to verify the suitability of the hydrogel sonographic model of the left ventricle (LV) in the computed tomography (CT) environment and echocardiography and compare the radial strain calculations obtained by two different techniques: the speckle tracking ultrasonography and the multislices computed tomography (MSCT). The measurement setup consists of the LV model immersed in a cylindrical tank filled with water, hydraulic pump, the ultrasound scanner, hydraulic pump controller, pressure measurement system of water inside the LV model, and iMac workstation. The phantom was scanned using a 3.5 MHz Artida Toshiba ultrasound scanner unit at two angle positions: 0° and 25°. In this work a new method of assessment of RF speckles' tracking. LV phantom was also examined using the CT 750 HD 64-slice MSCT machine (GE Healthcare). The results showed that the radial strain (RS) was independent on the insonifying angle or the pump rate. The results showed a very good agreement, at the level of 0.9%, in the radial strain assessment between the ultrasound M-mode technique and multislice CT examination. The study indicates the usefulness of the ultrasonographic LV model in the CT technique. The presented ultrasonographic LV phantom may be used to analyze left ventricle wall strains in physiological as well as pathological conditions. CT, ultrasound M-mode techniques, and author's speckle tracking algorithm, can be used as reference methods in conducting comparative studies using ultrasound scanners of various manufacturers.

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Abbreviations: LV, left ventricular; CT, computed tomography; MSCT, multislices computed tomography; RF, radio frequency; DTI, Doppler tissue imaging; US, ultrasound; PVA, polyvinyl alcohol; PR, pump rate; SV, stroke volume; S, strain; SR, strain rate; RS, radial strain; RSR, radial strain rate; ECG, electrocardiograph.

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1. Introduction

During the nearly four last decades, echocardiography has become the most widely applied clinical local and global method of the left ventricle functioning assessment by measuring the changes of the interior heart dimensions as well as of the heart walls contractility. At first, isolated measurements of the distance between two selected points on the echographic image were examined and then the systems registering images and processing radiofrequency (RF) signals were applied. There were high expectations regarding the Doppler tissue imaging (DTI) technique but this method has not been wider adopted for heart wall elasticity assessment, as yet.

One of the most common applications of echocardiography in the clinical practice is assessment of the left ventricle (LV) function. During the last two decades attempts have been made to implement a technique that would allow to analyze the work of the cardiac muscle during a loading test and at rest and to assess types of contractility disorders. The two promising tools used during the last decade have been the techniques acquiring scattered echoes energy from the unprocessed (RF) signals and Doppler assessment of tissue movement using the DTI [1,2]. However, the first technique suffers from the lack of a uniform algorithm of data acquisition and processing. The limitation of the second technique is, as in all Doppler techniques, the dependence of acquired data on the angle at which an ultrasonic beam reflects from the moving tissue of the myocardium. Different parameters of DTI and deformation have been examined in order to assess the LV function. However, neither of these techniques have brought any remarkable progress. From among different techniques used to examine elasticity of tissues, the so-called *speckle tracking* modality deserves special attention. In 1991 Bohs and Trahey [3] have developed the two-dimensional method of soft tissue movement measurement using ultrasound, regarding as forerunners of this technique.

In 1993 Ryan et al. [4] have developed the visualization method of intravascular elasticity of artery walls using a rotating high frequency (42 MHz) ultrasound transducer and applying the *speckle tracking* technique. However, the analysis of artery wall movement was not possible in real time but only after completing the RF signals' acquisition. The authors applied the blood vessel phantom made of gelatin and subjected it to an intravascular change of pressure from 100 to 120 mmHg, next, the RF echoes were correlated between successive frames. In 1994 Berrioz and Pedersen [5] applied the correlation method to study the diversified rigidity of model simulating vessel wall atherosclerosis. In 1995 Chen et al. [6] studied the dependence of errors of the *speckle tracking* method on the influence of different factors related to different types of examined tissue.

Properties of elastic soft ''tissue'' were measured using specially fabricated models in the form of elastic pipes made of different materials with echogenicity similar to that of the human tissue. At first the mixture of agar and gelatin was used to produce tissue phantoms. However, phantoms were not resistant to applied pressure changes – they were ruptured when radial deformations exceeded 5% [7,8]. Next, polyvinyl alcohol gel was used to produce tissue phantoms [9]. In 2004 Langeland et al. [10] made the first attempt to apply this type of phantom to assess heart wall deformations obtaining linear dependences between longitudinal and transverse strain – they used the RF echoes' correlation method.

The dual-chamber ventricular phantom was used for ultrasonic examination of the left ventricle strains during an *in vitro* experiment [11]. The authors focused on testing of the physiological LV strains by imitating more ellipsoidal shape of the ventricle and on the dependence of left ventricle functions and its ''interconnection'' with the right ventricle.

The three dimension tensor analysis of cardiac wall strains on the base of the univentricular polyvinyl alcohol cardiac phantom was recently published in 2012 by Heyde et al. [12]. However, using the radial strain, the authors were not able to detect the region affected by the pathological process under a relatively small stroke volume.

During the last decade the *speckle tracking* modality has been widely advertised by the majority of echocardiological equipment manufacturers, however, without a thorough discussion of the applied algorithms and mathematical expressions. The lack of important definitions of the measured parameters makes conducting comparative studies using ultrasound (US) scanners from different manufacturers impossible. This problem was noticed by echocardiographers carrying out examinations in clinics already possessing different US scanners. Despite the fast development of new scanning machines and new algorithms, the objective method of verification of the results (especially obtained using different scanners) is missing.

The main purpose of our work was to develop the mathematical/numerical model and construct a simple phantom of the LV deformation with the acoustic properties similar to those measured in the real echography. The LV phantom can serve as a reference physical model of the left ventricle for different imaging modalities having realistic deformation parameters which are stable over time and can be used as a calibration tool for commercial imaging systems.

There is a commercially available product that can be used as both an US and CT phantom Model: PVAH-01 (Shelley Medical Imaging Technologie, London-Ontario, Canada) however, there is no strain and strain rate mathematical modeling related to it.

The basis of the concept of speckle tracking algorithm is a relationship between displacement of the tissue scatterers and displacement of the resulting speckles in the US image. The support of the *speckle tracking* technique is even more inexplicable because it was proved in 2008 by Tournoux et al. [13] that this method is less reliable than the method of tracking ultrasound image tissue contours. However, the speckle tracking technique is used in most of the commercial echo units. Recent studies have shown that the Multislices Computed Tomography MSCT [14,15] are effective in quantitative analysis of LV strain. The novelty of this article is the introduction of the quantitative comparison of the radial strain (RS) results from the dynamically moving LV phantom obtained by means of three different modalities; MSCT, ultrasonic speckle tracking and numerical modeling.

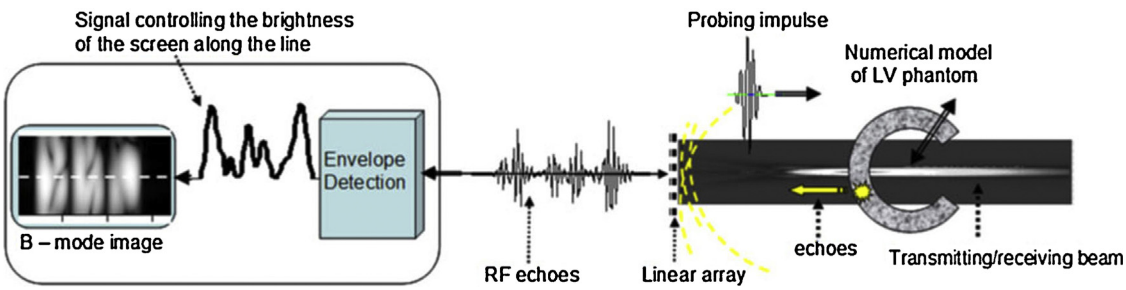

Fig. 1 – Acoustic track diagram of the LV phantom.

2. Materials and methods

2.1. Numerical model of speckle tracking

In this section we present some results obtained using mathematical and numerical modeling of phantom movement, previously described in [16]. The deformation of the phantom wall and resulting displacements of scatterers are modeled using the basic equations of the theory of elasticity [17] for the given boundary conditions. Modeling of the US imaging is based on the scattering theory [18]. Both models are integrated, especially in the context of description of deformation of the phantom wall, taking into account the random scatterers' distribution in the phantom wall material, resulting in the speckles pattern in the ultrasonic image (Fig. 1). The deformation of the region with heterogeneities (scatterers) results in a change in the position and brightness of the pixel along the image line. Changes in the material properties (pathological changes in the cardiac muscle) change both the size and spatial distribution of these inhomogeneities.

Fig. 1 schematically shows the steps of the numerical modeling: transmitting a probing signal; backscattering from the moving wall of the phantom, receiving the scattered field by the linear array; beamforming the RF line and finally constructing the echoes envelope line (white dotted line on the left).

Below we present an example of the numerical modeling of motion and deformation of the LV wall model. The calculations were made for a cylindrical ring with the boundary conditions similar to those of the real work of the left ventricle, Fig. 2.

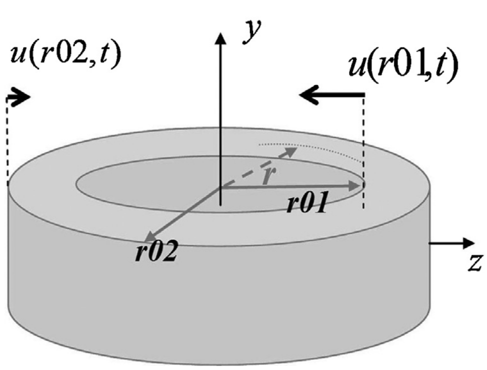

Fig. 2 – Numerical model of the left ventricular wall – boundary conditions in the diastole.

The model is described in cylindrical coordinates. The y-axis is a symmetry axis for the radially symmetrical ring deformation. In the cylindrical system of coordinates $R_r \equiv |\mathbf{R}|$, $R_r(r,t) = r + u(r,t)$, $R_\phi = 0 = R_y$, the radial displacement $u_r = u \equiv |\mathbf{u}|$, $u_\phi = 0 = u_y$ (the subscript r denotes the radial coordinate), t is time. $r01 \le r \le r02$, where $r01 = 18.8$ mm, and $r02 = 26.4$ mm are the rings of the internal and external wall boundaries in the initial state, respectively. We assumed the diastole for the initial state $u(r, t = 0) = 0$. The following boundary conditions for the displacements were assumed: $u(r = r01,t) = 0.5d1\cdot(\cos(\omega t) - 1)$, $u(r = r02,t) = 0.5d2\cdot(\cos(\omega t) - 1)$, d1 = 3.8 mm, d2 = 2 mm. In the numerical model of the LV three radially symmetrical layers, each 1.52 mm thick, positioned at the inner, middle, and external borders, respectively, were filled with randomly distributed scatterers, about five scatterers/mm^3 (see Fig. 3).

The distribution of scatterers in space was described by the scattering potential V:

$$V(\mathbf{R};t) \equiv \sum_{l}^{L} v_l \delta(\mathbf{R} - \mathbf{R}_l(t)) \tag{1}$$

where: v_l is the source strength of lth scatterer and $\mathbf{R}_l(t=0)$ is its initial random position.

The distribution of "energy" in the image is given by the formula:

$$|P^s(z,\mathbf{r}_s;t)|^2 = \sum_{l} |P^s(z,\mathbf{R}_l(t) - \mathbf{r}_s)|^2 + 2\sum_{l,m_{l\neq m}} Re(P^s(z,\mathbf{R}_l(t)) - \mathbf{r}_s)P^s(z,\mathbf{R}_m(t) - \mathbf{r}_s)^*) \tag{2}$$

where Ps is a complex point spread function (PSF) of a single scatterer in $\mathbf{R}_l$ for the scanner position $\mathbf{r}_s$, z-axis coordinate (PSF is the image of the point scatteter, given by an imaging system, describes the imaging system response to point input, and is analogous to the impulse response [19–22]. The second term corresponds to the interference pattern, * is a complex conjugate symbol.

The results of the numerical modeling of the detected backscattered signal from three layers in the phantom wall are shown in the right part of Fig. 4.

In the right part of Fig. 4 four phases of speckles' displacements in time (2), starting from the diastole at $t = 0$ up to a full contraction at $t = 0.5$ are shown – they were selected from 50 numerically computed subsequent images for the frame rate equal to 50 frames/cycle. The scans show the full range displacement of the wall. The central frequency of the scanning pulse was 3.5 MHz and consisted of four cycles with a

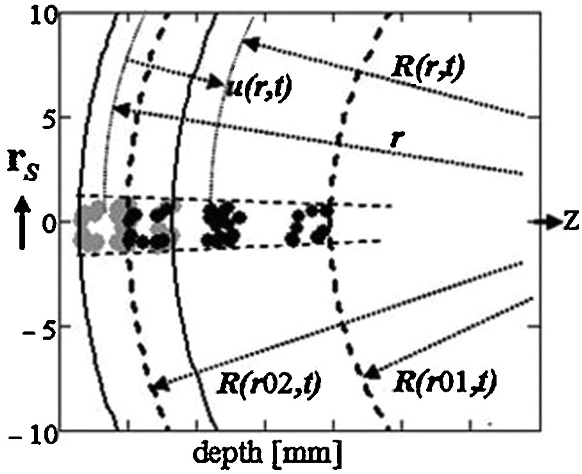

Fig. 3 – The initial state (diastole) and positions of the tissue scatterers in three layers are in gray. The final state (systole maximum contraction) and new positions of the markers are as black dots.

triangular envelope. In the left part of Fig. 4 the upper and lower solid lines represent the time course of the real position of the outer and inner model walls' boundaries.

For the time variation $t_m = m \cdot (1/50)$, $m = 0, 1, 2, \ldots, 50$, the positions of the black dots in a M-mode like plot correspond to the local energy maxima (2) for $r_s = 0$. As it can be seen, the positions of speckles (both imperishable and transitional due to the interferences) are in very good correlation with dynamic displacements of the scatterers in wall's layers. This can help to estimate the relative deformation (thickening or thinning) of these layers in different phases of movement.

2.2. Ultrasound model of the left ventricle

The left ventricular (LF) phantom was produced from 10% solution of the poly(vinyl alcohol) (PVA), suitable for processing. The material used to manufacture the phantom is a 10% (mass) aqueous solution of the PVA (molecular weight 89,000–98,000, 99 + % hydrolyzed, Sigma–Aldrich, St. Louis, MO, USA).

The solution was prepared by dissolving a weighted amount of PVA in water at temperature 80–85 °C. Magnetic stirrer mixed the solution until PVA became dissolved. In the next step the solution was cooled to the room temperature and placed in vacuum for 1 h to remove air bubbles suspended in the solution. After that, the solution was poured into the mold made from polymethyl methacrylate. Then the mold was placed in a freezer at -25 ± 0.5 °C for 35 h and then thawed for approximately 40 h to the room temperature [12,23]. This process was initially introduced by Surry et al. [23] and next applied by Langeland et al. [10] and by Leśniak-Plewińska et al. [11].

This material is durable and does not require additional venting before each experiment. The speed of sound in phantom material was equal to 1546 m/s and the attenuation was equal to 0.28 dB/cm (at 2.95 MHz).

Our model of the LV was driven by a computer-controlled hydraulic piston Super-Pump (Vivitro Inc., Canada) with adjustable fluid volumes and various loads. During a cycle of the pump, the stroke volume (SV) of water was pumped into the LV phantom and returned to the pump, resulting in changing the inner and outer diameters of the phantom.

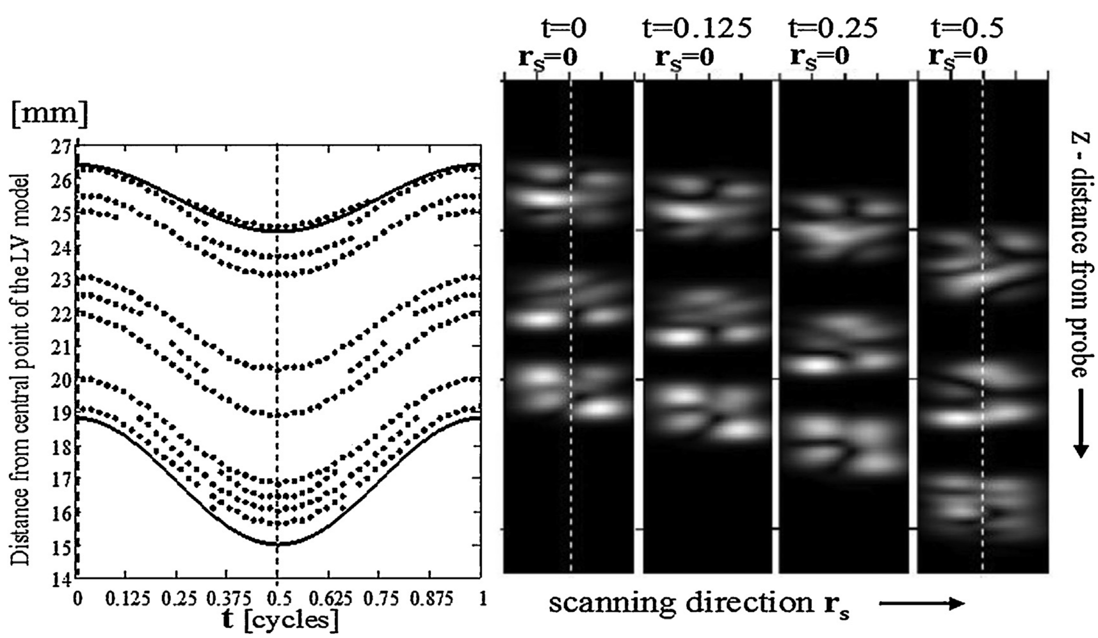

Fig. 4 – Modeling of speckle displacement during the diastole/systole phase. Speckles trajectories within the phantom layers are marked by dotted lines are located on the left part of the figure, continuous lines represent the time course position of the outer and inner model walls' boundaries. The speckle-like image of the distinctive layers within the wall for four phases of model contraction (from "diastole" at $t = 0$ to "systole" at $t = 0.5$, time is counted in cycles) is located on the right part of the figure.

Fig. 5 – LV phantom with a stiffened wall (darker patch) imitating the myocardial infarction.

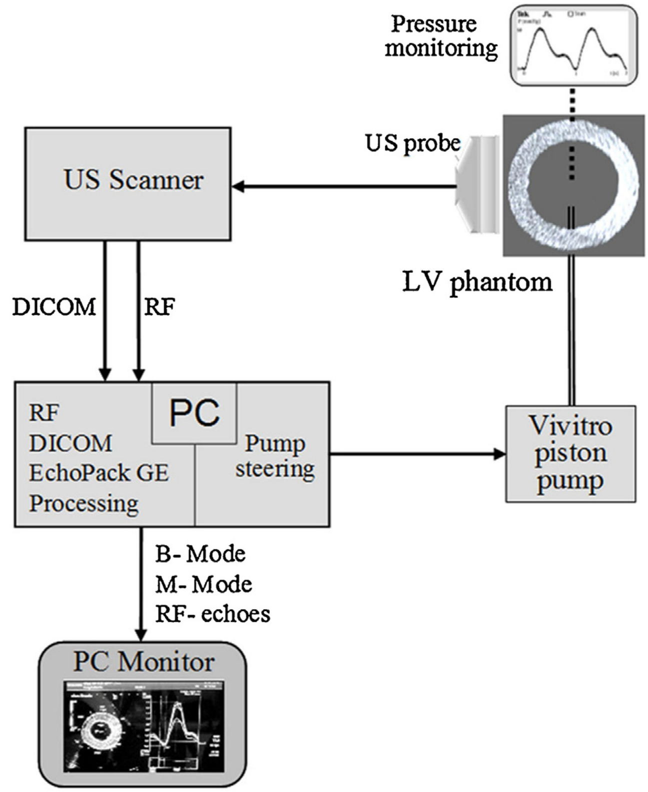

Fig. 6 – Setup for elastic properties' measurements of the LV model by ultrasound.

The LV phantom is presented in Fig. 5. The local hardness of the phantom wall mimicking the pathological changes was obtained by the drying process; a part of the wall material (the darker part at Fig. 5) was left in the dry for 48 h, while the rest of the material was immerged in water.

The percentage change of geometry (length, thickness) of examined targets is called the *Lagrangian strain*. The *strain* and the *strain rate* (SR) imaging enable segmental measurement of the myocardium for the assessment of its local and global functioning. If the left ventricle wall size increases by one-fourth, then the strain amounts to 25%, and if it decreases by three-fourths, then the strain amounts to −75%. The SR represents the changes in strain per time unit. Both parameters provide information about heart functions that complement each other. The *strain* and SR obtained in the echocardiography were verified *in vitro* and *in vivo* by using various methods. One of the simple ways of assessing the *strain* is by using the parasternal short axis M-mode view and calculating it by tracking the thickness changes of the LV wall phantom during the contraction.

2.3. *Ultrasound examination*

The repeatability of the parameters measured by using ultrasound scanners produced by different manufacturers is a major problem. The setup for elastic properties' measurements of the LV phantom is shown in Fig. 6. It consists of the LV model immersed in a cylindrical tank filled with water, hydraulic pump, the ultrasound scanner, hydraulic pump controller, pressure measurement system of water inside the LV model, and iMac workstation. The stroke volume can be changed from 10 ml to 100 ml. The fluid pressure was changed from 0 to 300 mmHg at the pump rate of 30–120 cycles per minute. The phantom was scanned using a 3.5 MHz Artida Toshiba unit at two angle positions: 0° and 25°. The previous measurements [25] were conducted for the Pump Rate PR of 40, 60, 80, 100, and 120 cycles per minute. The statistical analysis of the strain and SR measurements for two angle positions of ultrasonic probe was based on the non-parametric Mann–Whitney *U* test (the borderline value of the coefficient *p* of significance was set at 0.05).

The pressure inside the LV model was measured (Fig. 7) using the micro-tip catheter transducer (type SP-524, Millar Instruments, Inc., USA).

B-mode and M-mode images of the LV phantom posterior wall are shown in the left and right parts of Fig. 8. The RF echoes corresponding to the displacement of the posterior wall are displayed in the M-mode in diastole and systole (upper and lower parts of Fig. 8, respectively). The echoes were acquired using an Utrasonix scanner linear array, probe at 3.5 MHz with the RF output (SonicTouch Research Platform, Richmont, Canada).

The M-Mode was used to determine the maximum displacement of the inner surface of the phantom posterior

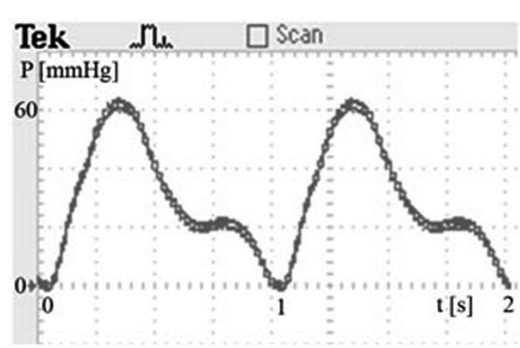

Fig. 7 – Pressure P course measured inside the LV phantom.

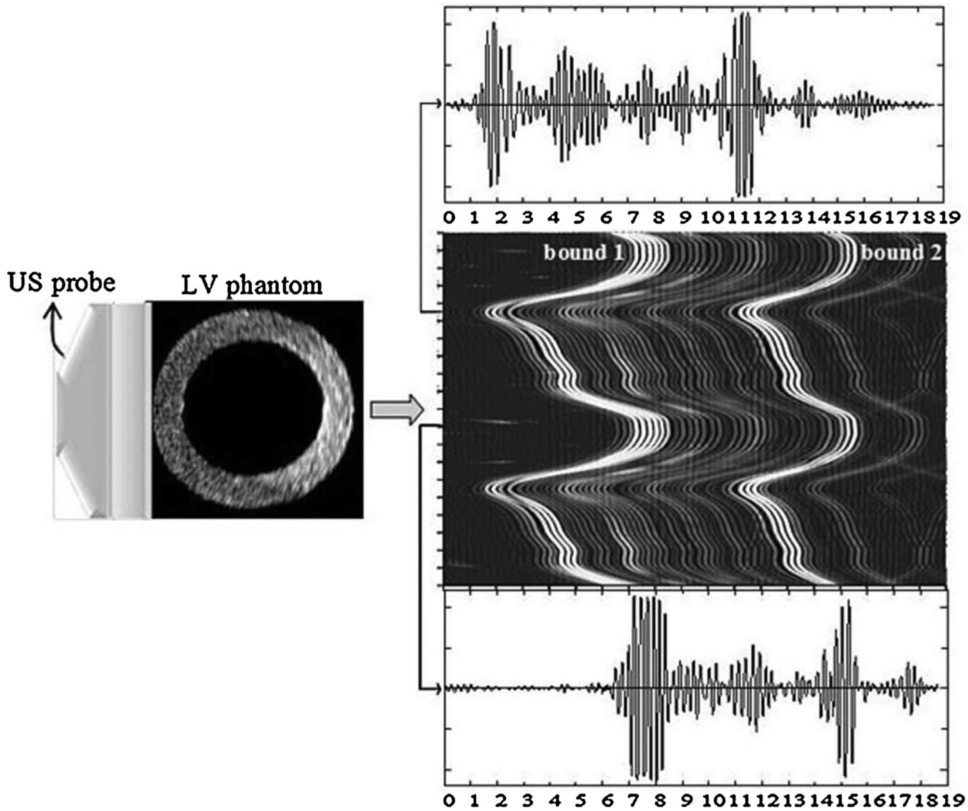

Fig. 8 – The M-mode presentation of moving surfaces of the posterior wall of the LV phantom (in the line scale of the brightness). The upper part of the figure presents the central RF line of the posterior wall of the LV phantom in the diastolic phase of the piston pump. The lower part of the figure presents the central RF line of the posterior wall of the LV phantom in the systolic phase of the piston pump.

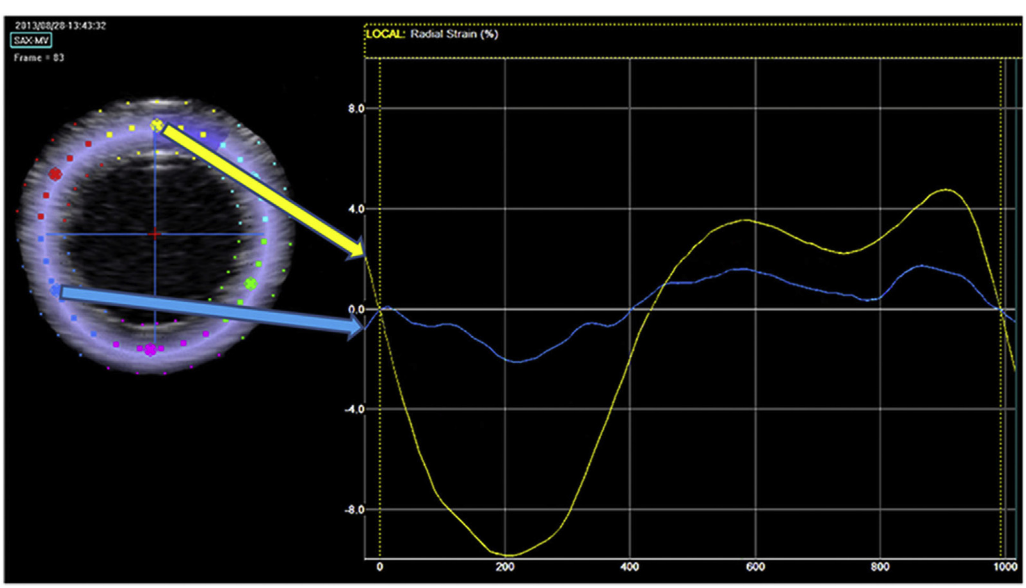

Fig. 9 – Radial strain traces of the LV wall phantom obtained using the Echo Pack PC GE software. The yellow trace corresponds with the elastic yellow segment of (radial strain = 15%), the blue trace represents the part of the blue segment (stiffer) of the LV wall phantom, imitating the myocardial infarction (radial strain = 4%). The horizontal axe is indicated in milliseconds.

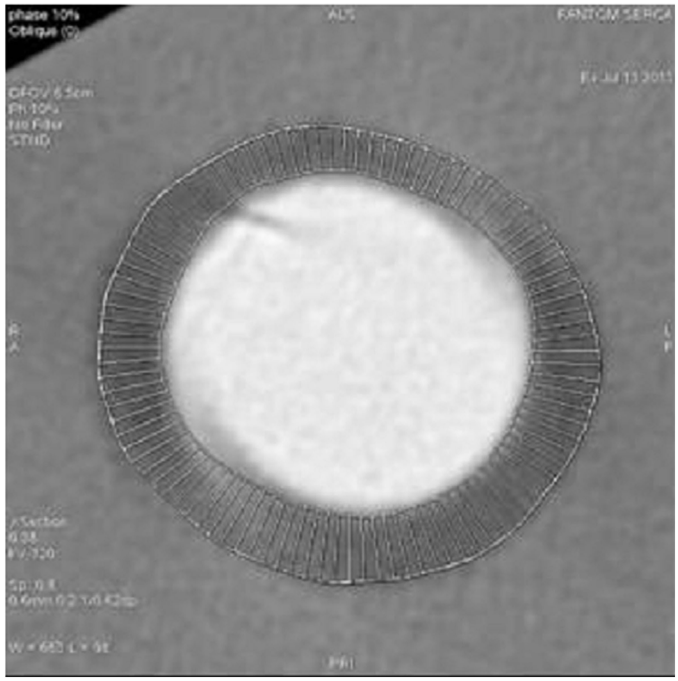

Fig. 10 – CT cross-section image of the LV phantom with the contrast agent at the diastolic phase of the piston pump cycle.

wall. It was assumed that the extreme position of the wall corresponds to the location of the first echo exceeding the half value of the maximum echoe – it is the first M-mode line whose brightness exceeds half of the maximum brightness. The difference between positions of the wall surface in diastole and systole is used to calculate our radial strain. In this work a new method of assessment of tracking quantitative speckles is also proposed.

In Fig. 9 two traces during one cycle = 1000 ms of the radial strain (RS) of the LV wall phantom, using the Echo Pack PC (GE Healthcare) software based up on scanner Vivid S5 (GE Healthcare) with 3.5 MHz linear array are presented.

2.4. *CT examination*

The comparison of the ultrasound scanning and CT techniques was performed and CT 750 HD 64-slice (GE Healthcare, 2012) machines. In both measurement sessions and in during numerical modeling the PR = 60 cycles/minute and the SV = 24 ml were used.

For the purpose of the CT scanning, the LV model was filled with the diluted iodine contrast agent in order to implement myocardium analysis protocol. The density of this solution (20% solution of the product Iodixanol 320 mgI/ml Visipaque 320, GE Healthcare) was similar to blood density in LV during a contrast enhanced cardiac study.

Retrospective ECG-gated MSCT acquisition Discovery CT 750 HD 64-slice (GE Healthcare, 2012) was performed with the 0.6 mm slice thickness.

The data were reconstructed every 10% of the R-to-R interval of the simulated ECG signal produced by the Vivitro SuperPump control device. The cross-section LV phantom images were obtained using a myocardium analysis protocol with manual settings of the valve and apex. The CT based measurements of the radial strain was performed for two from 10 MSCT images for the vertical axis of the LV wall thickness in the lower segment (the representative CT image of the LV model for diastolic and systolic phases are presented in Figs. 10 and 11, adequately). It can be observed between Fig. 11 and Fig. 10 the growing of the both internal and external LV phantom diameters and lower thickness of the LV phantom wall. The relative change of LF phantom wall gives RS.

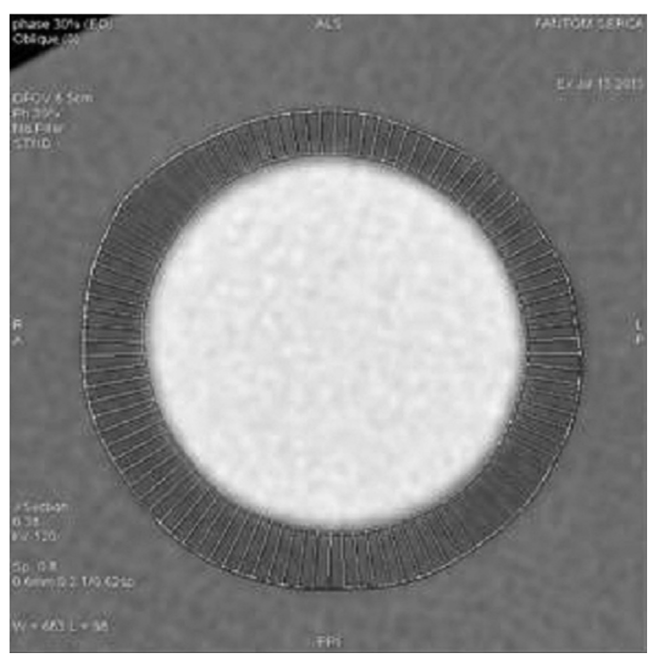

Fig. 11 – CT cross-section image of the LV phantom with the contrast agent at the systolic phase of the piston pump cycle.

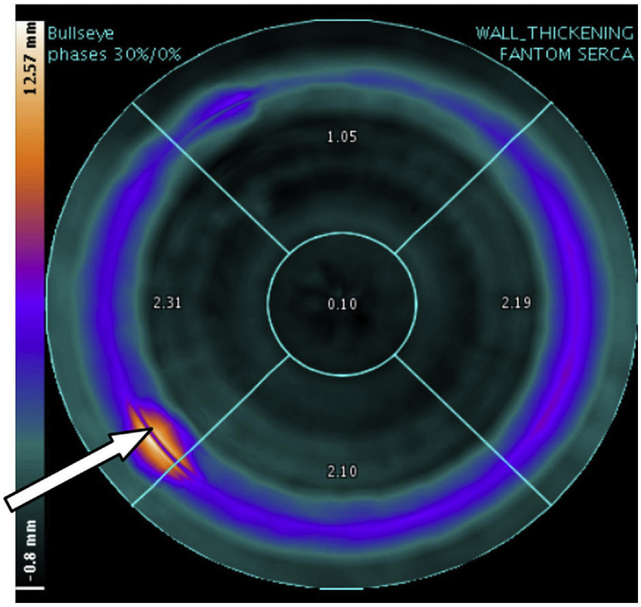

Fig. 12 – MSCT Bull's eye display of the LV model, arrow points to the stiffened wall region.

3. Results

The radial strain (RS_{nm}) calculated for the selected boundary conditions for the numerical model (upper and lower solid lines at an left side of Fig. 4) is equal to 19.1%. Applying of the speckle tracking algorithm to the results of numerical speckle formation (at an right side of Fig. 4) and their motion detection (dotted lines on right side of Fig. 4) resulted in the radial strain RS_{nST} = 19.3% and was very closed to RS_{nm}. The radial strain calculated from the M-mode recording of the phantom wall displaced (Fig. 8) was slightly lower, $RS_{M\text{-mode}}$ = 18.75%.) Under the EchoPack PC software environment, based on the results obtain using the Vivid S5 (GE Healthcare) scanner the cross-section of the LV phantom the yellow trace represents the radial strain course of the elastic wall (up to 15% at maximum) and the blue trace shows the impaired elasticity with a very low radial strain below 4% (Fig. 9). The radial strain obtained by the MSCT imaging between phases presented in Figs. 10 and 11 was RS_{CT} = 18.76%. MSCT Bull's eye plot [24] of the LV clearly shows the stiffened part of the wall of the LV phantom (arrow in Fig. 12).

4. Discussion and conclusions

MSCT imaging provides excellent isotropic resolution, decreases partial volume effect and motion artifacts, which makes it an appropriate reference method for echocardiographic measurements. Settings of valve and apex were positioned manually to reconstruct cross-section LV phantom images. In spite of these manipulations, further analysis involved time-consuming manual correction of inner and outer LV phantom wall borders, which is undoubtedly one of study limitations.

The proposed numerical model and phantom of the left ventricle are the realization of a fully controlled diagnostic environment allowing for the testing of algorithms that follow and analyze scattering of speckles, as well as improving the existing algorithms or supporting the construction of new ones. The study indicates the usefulness of the ultrasonographic LV model in the CT technique. Using ultrasound technique, the non-parametric Mann–Whitney *U* statistical test indicated that both the radial strain (RS) and radial strain rate (RSR) were independent from the insonifying angle or the pump rate [25]. The results showed a very good agreement, at the level of 0.9%, in the radial strain assessment between the ultrasound M-mode technique and multislices CT examination. However, the commercial software used in Vivid S5 scanner gave the mean value of the radial strain as equal to 15% (Fig. 9), being about 20% lower than the radial strain measured using our speckle tracking algorithm and the radial strain obtained by the MSCT imaging. The presented ultrasonographic LV phantom may be used to analyze left ventricle wall strains in physiological as well as pathological conditions [25]. The tree modalities MSCT, ultrasound M-mode and our modeling speckle tracking algorithm used in the comparative studies of the phantom displacement shoved very similar results differing by no more than 0.9%. It confirms our assumptions, that our simple LV phantom can be a useful calibration tool for the commercial scanners from various manufactures.

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