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ORIGINAL PAPER

Strain and strain rate deformation parameters: from tissue Doppler to 2D speckle tracking

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Abstract Strain and strain rate deformation parameters based on Color Doppler Myocardial Imaging, and more recently on two-dimensional (2D) gray scale images, have evolved as important methods for the quantification of myocardial function. Although these parameters are already applicable in the research field, their acquisition and analysis involve a number of technical challenges and complexities. Accurate knowledge of the basic principles of those techniques, as presented in this article, will further enhance their applicability to clinical practice.

Keywords Imaging · Echocardiography · Strain

Abbreviations and Acronyms
2D 2 Dimensional
S Strain
SR Strain Rate

Introduction

The quantification of regional myocardial function remains a challenge in clinical cardiology. Traditional methods for the evaluation of regional myocardial function using echocardiography are subjective and only partially quantitative [1, 2]. Over the past 10 years, Color Doppler Myocardial Imaging (CDMI) and more recently 2D-speckle tracking have evolved as new clinical and research tools to quantify regional myocardial function. This approach seems to offer substantial advantages over standard gray scale echocardiography.

Although the theoretical background of Doppler imaging, based on the Doppler effect, was discovered by Christian Andreas Doppler (1803–1853) nearly one and a half centuries ago [3], it was implemented as an application in medical ultrasound procedures more than 100 years later, utilizing the basic principle for measuring velocities from the Doppler shift of a reflected ultrasound beam. However, many key prerequisite steps were accomplished by various investigators including Isaaz [4], McDicken [5], Sutherland [6], Fleming [7] Heimdal [8], Meunier and Bertrand [9], among others, establishing the evolution of strain echocardiography based on Color Tissue Doppler Myocardial Imaging or speckle motion/tracking, subsequently.

Color flow imaging and color tissue Doppler myocardial imaging

Color flow imaging takes the pulsed-wave Doppler concept a step further, by using Doppler frequency shift detection over a set of range gates, and along a

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number of acoustic lines. The result is a two-dimensional image, depicting flow, which is superimposed on the two-dimensional grey-scale conventional real-time b-mode image, generated from backscattered echoes.

The detection and quantification of blood flow velocity with ultrasound is based on the scattered echo from the red blood cells. The color Doppler blood pool algorithm is set to detect the high velocity/low echo amplitude moving blood. Myocardium is a relatively low velocity/high echo amplitude structure and can be distinguished from the blood by the implementation of appropriate thresholding and clutter filters, which represent one of the differences between color Doppler blood and myocardial imaging (Fig. 1).

In the pulse wave spectral Doppler systems, Doppler information is obtained from only a single sample volume. In the color flow system, each line of the image is made up of multiple adjacent sample volumes and the Doppler shift information for each line is obtained from several transmitted pulses.

Unlike spectral Doppler which relies on the Fast Fourier Transform (FFT) to extract the whole spectrum of frequencies that are present, color flow imaging uses a method, introduced in this context in 1985 by Kasai et al [10], known as autocorrelation.

The autocorrelation algorithm utilizes the echoes from two consecutive pulses to estimate the so-called autocorrelation function (ACF) of the data in a given line segment, by comparing the respective echo lines, in order to calculate the average Doppler frequency for a set of pixels along the axis beam. When the estimates are complete and mapped into the image, the next beam line is interrogated in the same manner [11].

If two RF signals (reflected ultrasound from tissues) are thus acquired at two time instances, t1 and t2, the phase shift ($\Delta\varphi$) between these two signals can be calculated. The velocity (V) of the tissue underlying the RF signal can be calculated as follows [12]:

$$V = \Delta\varphi \ PRF \ c/4 \ \pi \ fo$$

where c is the speed of sound, $PRF = 1/(t2 - t1)$ is the pulse repetition frequency, fo is the central frequency of the signal, $\Delta\varphi$ is the phase shift of the autocorrelation function of a data segment, which can be estimated from 2 consecutive Doppler demodulated pulse echo sequences in that segment, acquired at times t1 and t2.

The velocity information is then color-coded and superimposed on the conventional 2D grey scale image. By convention, Red is depicted as movement

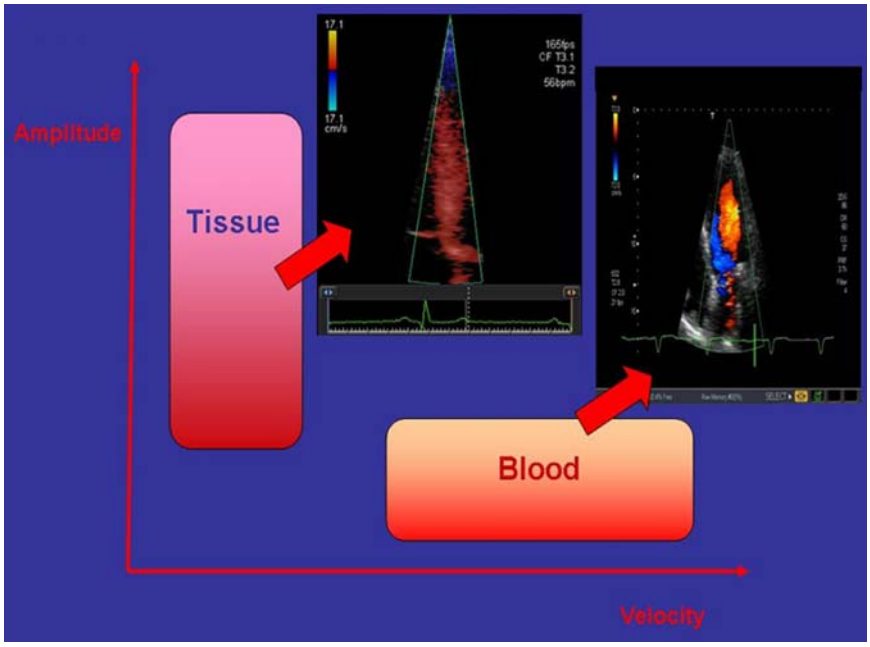

Fig. 1 High echo amplitude/low velocity tissue and low echo amplitude/high velocity blood pool

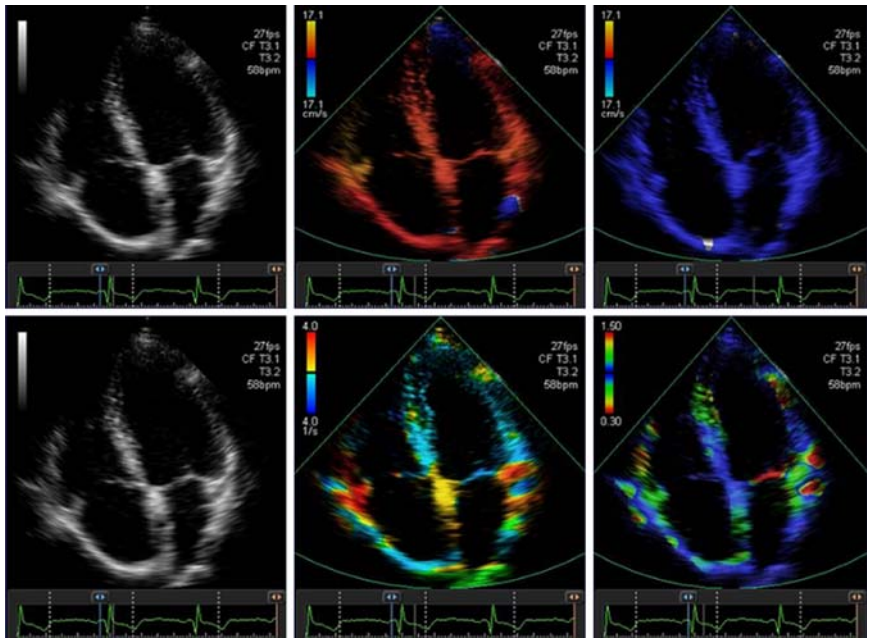

Fig. 2 *Upper row*: Superimposed color-coded velocity information on gray-scale 2D image. (Red: Systole, Blue: Diastole) *Down row*: Superimposed color-coded Strain Rate and Strain information on gray-scale 2D image

towards the transducer and Blue away of the transducer (Fig. 2).

One of the major limitations of any Doppler technique is the dependency of the measurements on the insonation angle between the ultrasound beam and the direction of motion (velocity vector). The estimated velocity is underestimated by 6% at an angle of 20°, 13% at 30°, 29% at 45° and zero at 90° [13]. This is also true for Doppler derived Strain and Strain Rate parameters, once velocity information is incorporated in the basic computational method (explained below).

For the above reason, estimation of only the axial component of velocity or deformation is feasible in clinical settings using Doppler techniques.

Motion -Velocity vs. Deformation-Strain

A moving object is not necessarily undergoing deformation so long as every part moves with the same velocity. Deformation occurs when different elements of the object move at different velocities so the object has to change shape during its movement.

In this case, the fractional change in length of an element of the object, compared to its original length, is called Lagrangian Strain**.** In a beating heart, the unstressed original length is difficult to measure, so end-diastolic length is most often used [14] (Fig. 3).

However, the deformation can also be expressed not only relative to the original length but also relative to the length at a previous moment in time (dt). In this definition the reference value is not constant over time, but changes during the deformation process and the strain is called instantaneous Natural Strain.

Lagrangian and Natural Strain are related via a nonlinear relationship, so in small deformations the two strains are approximately equal, but in large deformations the difference becomes significant [15, 16].

$$\text{Natural Strain (t)} = \ln\,(1 + \text{Lagrangian Strain (t)})$$

ln: the natural logarithm

Strain is a dimensionless quantity, and the resulting deformation of the material is generally expressed as a percentage %.

Strain rate (SR) is the rate at which deformation changes, i.e. change of strain per unit of time dt. It is the temporal derivative of strain and describes the rate of shortening or lengthening of an object or a part of the heart. As strain rate describes the speed of deformation, its measurement unit is (1/s).

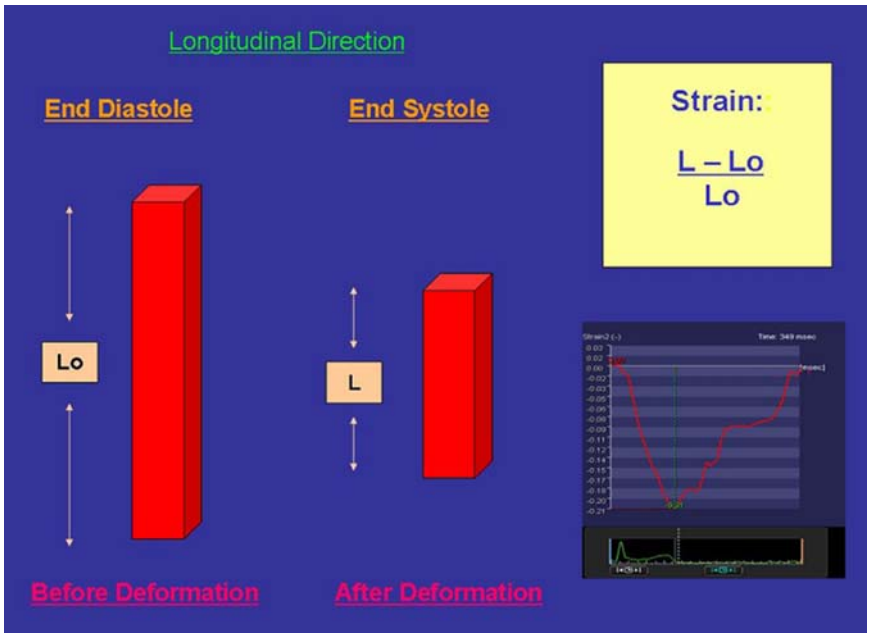

Fig. 3 Strain pattern of a region of the heart in the longitudinal direction. Lo, original length at end diastole; L, length after deformation

$$\text{Strain} = (L - Lo)/Lo,$$

$$\begin{aligned}\text{Strain Rate} &= d\text{Strain}/dt \rightarrow \text{Strain rate} \\ &= (L - Lo)/dt\ Lo \rightarrow \text{Strain Rate} \\ &= V - Vo/Lo\end{aligned}$$

L/dt = Velocity V, Lo = original length, L = length at time t.

In practice, rather than measuring instantaneous local velocities Va, Vb (estimated by the Doppler technique) in the myocardial wall, the spatial gradient of velocities in a predefined length (d) is determined by the Strain Rate of that region (Fig. 4). Thus, instead of measuring just the velocities (Va and Vb) at the ends of offset distance d, the strain rate is calculated from the slope of the

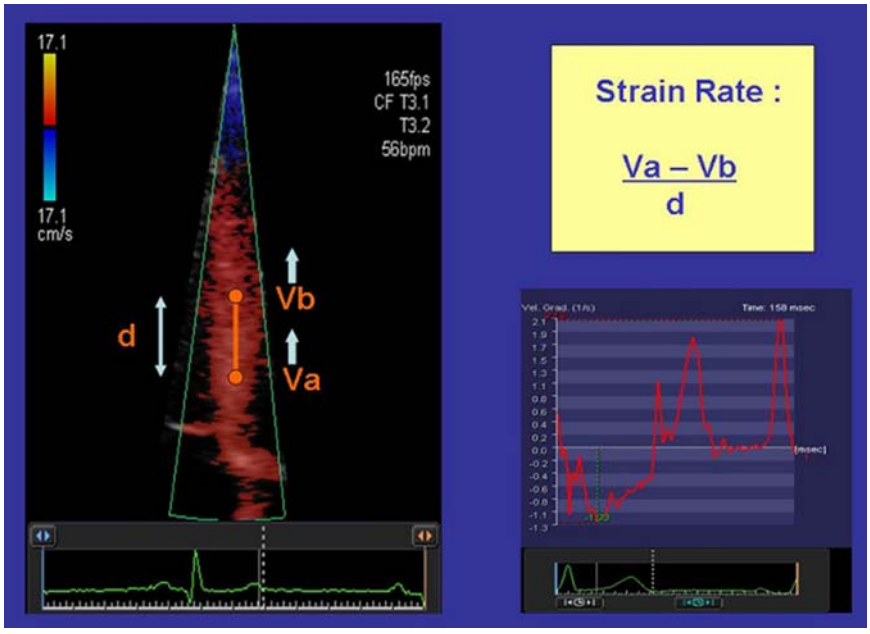

Fig. 4 Strain Rate estimation in a region of septal wall. Va and Vb are the instantaneous velocities in a predefined length d

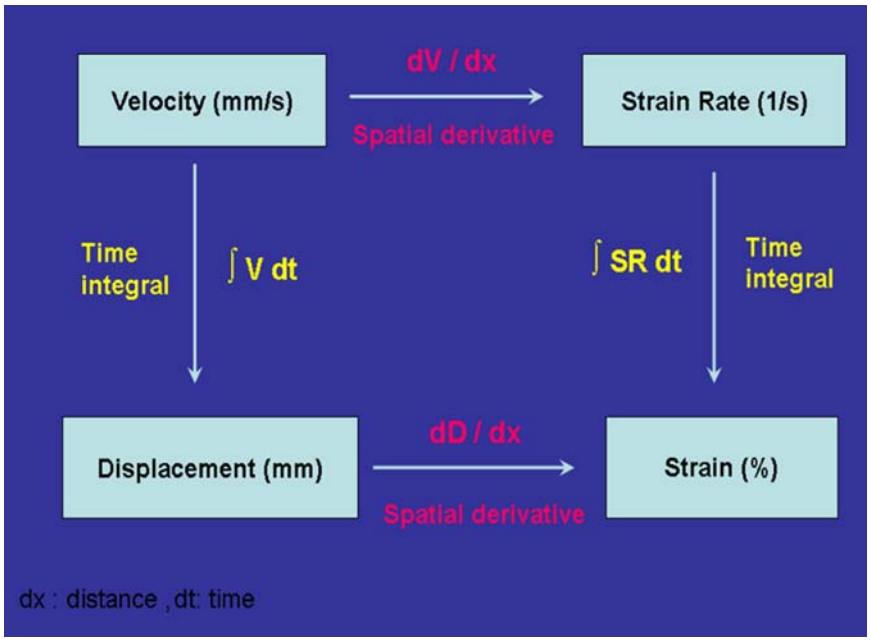

Fig. 5 The different paths in deformation calculation

regression line of all velocities along that distance (d) [17]. The temporal integral of the Strain rate is then estimated to give the Strain. The theoretical algorithm of strain and strain rate estimation is summarized in Fig. 5, where different paths are depicted. Velocities, strain and strain rate information may be color coded and can be superimposed on a 2D image (Fig. 2). Myocardial fibers shorten in the longitudinal and circumferential directions and thicken in the radial direction. By convention, shortening is indicated by negative SR and lengthening by positive SR values.

Deformation and ultrasound coordinate system

In order to define movement in space, a three-dimensional Cartesian xyz coordinate system may be set out. Every single spatial point is then described by three coordinates (Vx, Vy, and Vz) that can correspond to velocity or deformation.

Two coordinate systems can be defined, one in relation to the ultrasound beam and the other in relation to the myocardial deformation.

The ultrasound coordinate system consists of:

(1) Axial direction: Along the direction of beam propagation.
(2) Lateral direction: In the plane direction, perpendicular to the axial direction.
(3) Elevation direction: Out of plane, perpendicular to axial and lateral directions.

The myocardial coordinate system and its function, can be described with respect to the three principal spatially oriented motions-deformations, that the heart follows at any cardiac cycle: (Fig. 6)

(1) Longitudinal deformation: From base to apex.
(2) Radial deformation: Perpendicular to epicardium and to the longitudinal axis.
(3) Circumferential deformation: Perpendicular to radial and longitudinal axes.

Although for reasons of simplicity the deformation of a three-dimensional object can be described by the aforementioned longitudinal, radial and circumferential Strains, (Normal strains), in reality the complete deformation of a 3D structure is characterized by an additional 6 Shear Strains, which are referred to as relative movements of the borders of a volume element relative to each other (Fig. 6). Thus, the overall number of strain components of a three-dimensional object theoretically is nine. Thus, our current measurement approaches are a vast simplification of the true deformation of the heart.

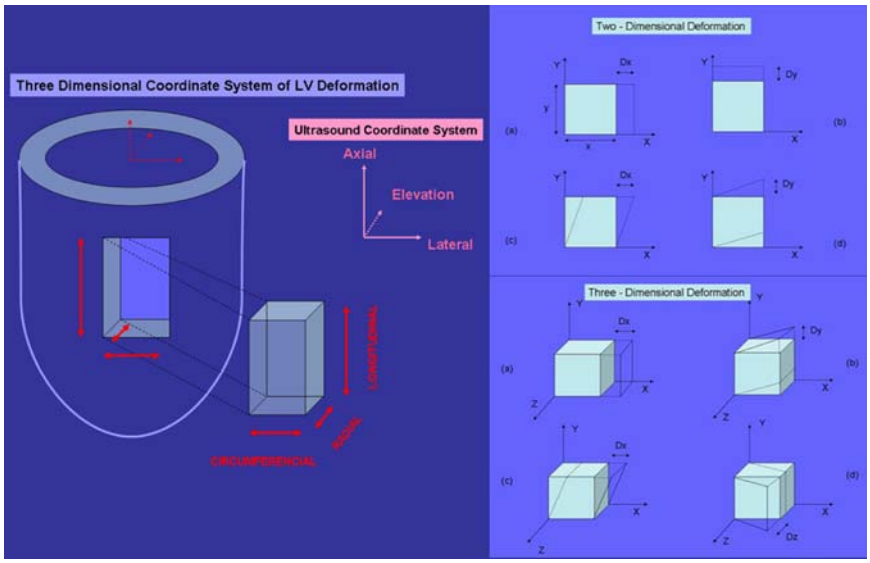

Fig. 6 *Left*: The three-dimensional coordinate systems of the heart and the ultrasound beam. *Right Upper*: Two Normal Strains (a, b) and two Shear strains (c, d) in 2D deformation. *Right Bottom*: Normal Strain (a), along with three Shear Strains (b, c, d), in 3D deformation

Regional strain/Strain rate measurements and contractility

In cardiac muscle physiology, strain is directly related to fiber shortening and SR to the speed of shortening, which is a measure of contractility. Several groups of investigators have demonstrated the superiority of SR and strain and better site specificity, over tissue Doppler velocity data for tracking local systolic function [18–22].

In principle, the superiority of deformation parameters for assessing cardiac function compared to motion-velocity-displacement parameters is related to the basic strain-algorithm, which subtracts the motion due to the contraction of neighboring segments (tethering effects and translational motion). Completely passive segments can show motion relative to the transducer due to tethering, but without any deformation, making velocity and displacement information completely unreliable for the characterization of such regions. Strain parameters on the other hand, are referred to as motion-deformation between two points in the myocardial wall, which is unrelated to the motion towards the transducer, and this fact discriminates the actual passive movement from true contraction in any myocardial region.

Color Doppler myocardial imaging has been used as a clinical and research tool for estimating deformation parameters with high spatial accuracy. The actual spatial resolution depends on many technical issues, but generally falls at best in the range from 1 to 5 mm [23].

Doppler-derived myocardial strain and SR have been validated using tissue-mimicking phantoms [24, 25], sonomicrometry [26], or magnetic resonance imaging [27].

Weidemann et al. [28, 29] showed that ultrasound-derived peak posterior wall systolic SR accurately follows changes in myocardial contractility induced by epinephrine, b-blockers, and pacing. On the other hand, peak systolic Strain correlates best with changes in stroke volume-EF and therefore is more closely related to changes in global hemodynamic than changes in contractility.

Greenberget et al. [30] demonstrated that peak and mean systolic SR are strong non-invasive indices of LV contractility, as confirmed by analysis of peak elastance. Using a closed-chest anesthetized pig model, Jamal et al. [31] demonstrated that peak radial systolic SRs in normal myocardium during dobutamine infusion were linearly correlated with peak positive dp/dt, a relative load-independent measure of LV contractility.

Region of interest (ROI)

The region of interest (ROI) is an area defined manually by the operator in which average velocity or strain measurements, rather than individual pixel values, are estimated. It provides a spatial average of the computed values, and although it gives a substantial reduction of noise in the samples it also

results in a loss of axial resolution as the ROI becomes larger (Fig. 8).

Frame rate

The precise measurement of peak values in strain/strain rate curves is one of the major technical considerations. The sample rate of interrogation is determined by the Frame Rate setting in the scanner, and defines the temporal resolution of the technique. The number (density) of scan lines constitutes another important parameter and regulates the lateral resolution of the measurements. As color flow imaging is a pulsed wave Doppler technique, the Doppler shift information for each line is obtained from several transmission pulses, thereby restricting the full range of frame rate values. The number of vector lines of velocity information which can be produced per second is restricted by the fact that a minimum two pulses must be transmitted for each beam line for an estimation of the mean velocity. Generally, approximately eight pulses are used [32]. In order to have a reasonable line density in each frame, the field of view width, the frame rate or the depth of penetration must be limited (Fig. 7).

As presented in Fig. 7 the frame rate is inversely related to the number of scan lines. A decrease in a number of scan lines can increase the frame rate, but the lateral resolution will also be decreased in that case. As most cardiac events are short-lived, the higher the frame rate the better the probability of accurately reconstructing the interrogated sample.

Choosing a narrow imaging sector is the most common way of increasing the frame rate and optimizing temporal resolution. Increased Frame Rate can be also accomplished by increasing PRF, decreasing line density, depth of penetration or the number of pulses transmitted along each line (Fig. 7).

Another specific method of getting high frame rates is called parallel beam-forming [33]. This is a method that requires the reception of multiple RF lines from a single transmit pulse. This is achieved by focusing in different directions during signal reception using different time delays on the array elements and thereby, increasing temporal and/or lateral resolution.

By increasing from 2 to 4 received parallel beams during multi-line acquisition, the temporal and lateral resolution can be increased from 100 to 161 frames per second and from 8 to 12 beams respectively [34].

Typically, frame rates above 150 Hz are adequate to resolve most of the regional strain rate curves during systole and rapid ventricular filling, but frame rates above 200 Hz might be needed to adequately interrogate isovolumic periods.

Effect of harmonics

The use of harmonic frequencies of the transmitted signal has been shown to reduce reverberation noise

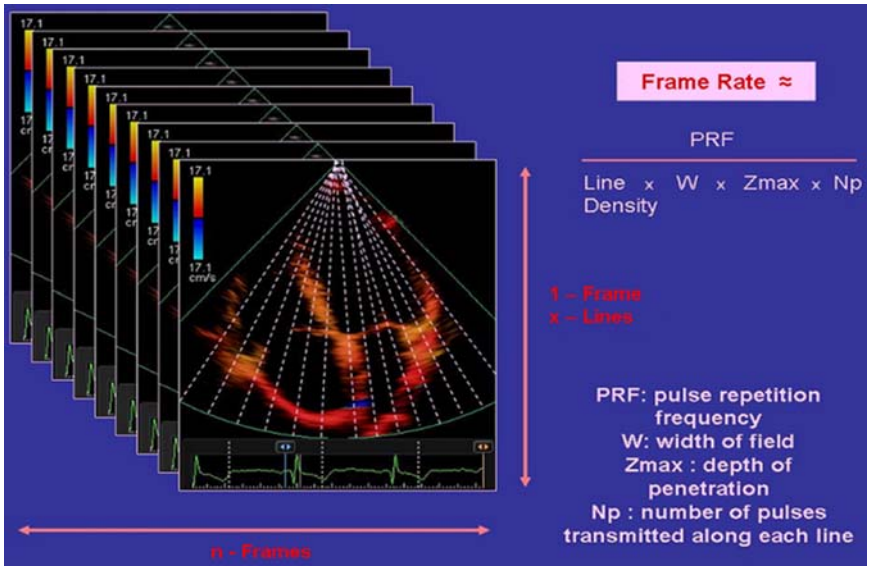

Fig. 7 Frame rate relations. The frame rate is proportionally related with the PRF and inversely related with the scan lines, width of field, depth of field and number of pulses transmitted along each line

and side-lobes, thereby improving the image quality of two-dimensional echocardiography [35]. To create the harmonic signal, high pulse strength is required. In general, adequate intensity is obtained near the axis of the transmitted ultrasound beam and this results a narrower beam profile compared to conventional fundamental imaging. Better quality of SR curves and more accurate SR estimation with a reduction in the variation of normal findings seem to be the benefits of harmonic settings for strain data.

Quantification of strain rate noise

Strain rate calculations are very dependent on the image noise and artifacts, and different calculation algorithms may provide inconsistent results [36]. The strain rate estimator is based on the linear regression of a number of velocity samples. When analyzing the longitudinal contraction of the myocardium from the apex, velocity is assumed to increase linearly along the length of the sample line. In a normal segment, velocity samples should be relatively close to a linear regression line, and the degree of variation from the regression is an index of signal noise [34]. The deviation of displacement from the regression line gives an estimate of the noise in the data.

Longitudinal and radial function

Longitudinal function, as exhibited by Doppler strain and SR data, can be measured in every segment of the ventricular wall, provided that the ultrasound beam is parallel with the vector of motion of the interrogated segment. By convention, shortening is indicated by negative S/SR, so that systolic Strain/SR curve is expected to be negative in the longitudinal direction (Fig. 8). The Radial function can be estimated from the posterior wall in parasternal long or short axis views. As in the radial direction the wall is thickened in systole, the S/SR curve is expected to be positive. Estimation of radial LV deformation parameters at the septum is limited, as myocardial structure in this region is constituted by both right and left ventricular fibers. Circumferential function theoretically could also be recorded from the short axis view at inferoseptal and lateral walls, but the influence of the radial component vector usually contaminates the data.

Two-dimensional strain – Speckle tracking

As stated previously, the Doppler technique is angle-dependent and this is the main reason why measurements are limited to certain segments, especially

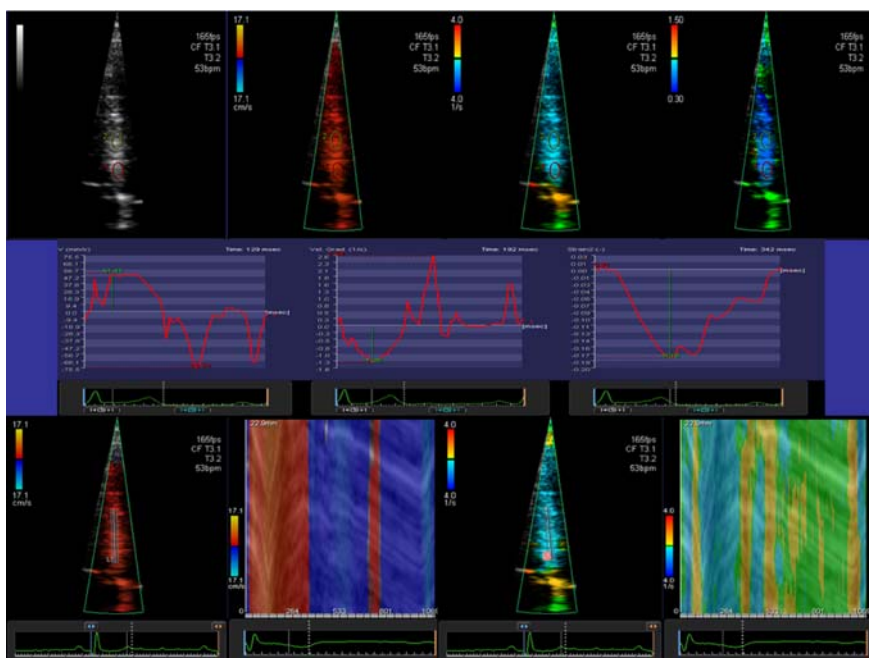

Fig. 8 *Upper row*: Septal interrogation along with regions of interest (ROI) depicted as red and green circles. *Middle row*: Longitudinal velocity, Strain Rate and Strain curves. *Bottom row*: Color coded M-mode representation of Velocities (left) and strain rate (right) as a function of time

when trying to estimate radial and circumferential function. An alternative method, using B-mode images, is the estimation of strain and strain rate by Speckle tracking.

The reflected ultrasound from the tissue is the result of interference by numerous reflected wavelets from the inhomogeneous medium. The interference pattern (resulting in bright and dark pixels in a B-mode image) (Fig. 9), remains relatively constant for any small region in the myocardium. This unique pattern (fingerprint) is called Speckle. In the speckle tracking technique, a defined region (Kernel) is tracked, following a search algorithm based on optical flow method [37], trying to recognize the most similar speckle pattern from one frame to another [38]. The algorithm searches for an area with the smallest difference in the total sum of pixel values, which is the smallest sum of absolute differences.

The technique is angle-independent as it is based on the displacement of speckles, defined in respect to the wall rather than the ultrasound beam, as Doppler techniques do, and has been validated by ultrasonomicrometry [39].

It provides information for *longitudinal*, *circumferential* and *radial* myocardial function, estimating directly Lagrangian Strain parameters, with better lateral resolution compared to tissue Doppler, due to a higher density of scan lines in the gray scale images.

Although it seems to be an attractive method, limitations are also present. The assumption that the speckle pattern remains constant throughout the motion, the re-orientation of myocardial fibers, the out-of-plane motion of scatterers, the resemblance between subsequent images, the ability to track with sufficient temporal resolution and the difference in axial and lateral resolution are all factors that modify the extracted values and their significance remains to be clarified.

Pitfalls and limitations

Strain techniques are generally very sensitive in determining their measured output, but at the same time they are extremely vulnerable to a number of potential pitfalls.

The first universal rule applicable to any strain technique is that the quality of images needed for strain analysis should be above average. Regarding Doppler strain, apart from the alignment with the beam mentioned above, noise incorporation, reverberation artifacts, dropouts, aliasing, lateral resolution limitations, insufficient tracking and inadequate frame rate constitute the main challenges of the technique.

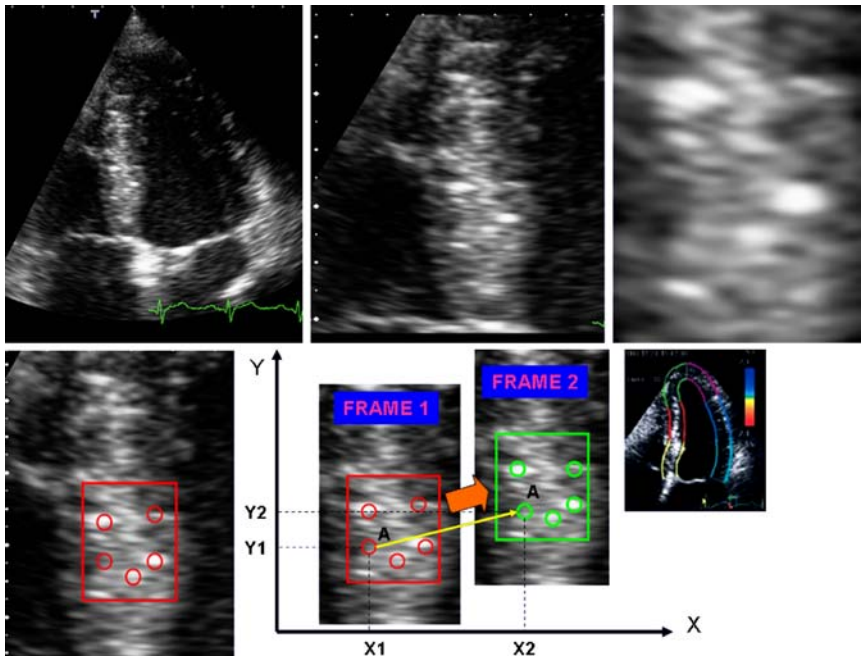

Fig. 9 The speckle pattern on the myocardial wall and the search algorithm by the kernel (red -green square) in two subsequent frames

For 2D strain determination, alignment is not a problem, but lateral resolution in the far field, out of plane motion, reverberations, dropouts and the quality of the B-mode image are important considerations. Frame rate limitations still exists under certain circumstances such as the displacement of the pixels needed to be searched at certain time intervals under the current available applications.

Clinical and research applications

Accurate knowledge of myocardial function is essential to the management of cardiac disease, and the quest continues for an optimal, quantitative technique to assess myocardial contractile function [40, 41]. Strain analysis seems to provide important information regarding quantitation of myocardial function, already applicable primarily in research, but also in clinical settings.

The application of strain analysis has been implemented in numerous experimental and clinical studies for exploration of cardiac function under various conditions, some of which will be cited representatively in this review.

Coronary artery disease

Strain analysis has been used to investigate whether the changes in myocardial deformation could accurately identify an acutely ischemic myocardium during coronary angioplasty. Systolic and post-systolic strain parameters measured during total coronary occlusion identified the existence of an acutely ischemic myocardium with a sensitivity of 86–95% and a specificity of 83–89% [42]. Edvardsen et al. [43] determined that the new Strain Doppler Echocardiography approach might be a more accurate method than Tissue Velocities for detecting systolic regional myocardial dysfunction induced by LAD occlusion.

Left ventricular function/Hypertrophy/ Cardiomyopathy

Serri et al. [44] found that despite an apparently normal left ventricular systolic function based on conventional echocardiography, all strain, longitudinal, radial and circumferential components were significantly reduced in patients with hypertrophic Cardiomyopathy (HCM). In another study [45], researchers were able to differentiate between HCM and hypertensive LV hypertrophy based on strain data, whereas, Palka et al. [46] showed that, in restrictive cardiomyopathy, the systolic Strain Rate (SR) and diastolic SR (E) during rapid ventricular filling, were lower compared to constrictive cardiomyopathy. Strain imaging has also been used to disclose early signs of infiltrative cardiac disease in familial amyloidotic polyneuropathy [47].

Tetrology of Fallot (TOF)

Weidemann et al. [48] studied the deformation properties (SR and S) of the right and left ventricles in asymptomatic patients after TOF correction, finding reduced values in the RV wall.

Valvular disease

Evaluation of global long-axis function by SR imaging was demonstrated to be a feasible method for detecting subclinical LV dysfunction and accurately predicted contractile reserve in asymptomatic patients with severe MR [49].

Stress echo/Viability

Hanekom et al. [50] sought to assess the feasibility and accuracy of strain and SR imaging (SRI) during the Dobutamine Echo for prediction of functional recovery in patients undergoing revascularization after MI and its incremental value to conventional wall-motion scoring (WMS). They concluded that a combination of WMS and SRI parameters augmented the sensitivity for prediction of functional recovery above the use of WMS alone (82 vs. 73%) although specificities were comparable (80 vs. 77%). In another study, normal values of exercise strain echocardiography were investigated [51].

Heart failure/Dyssynchrony

Strain echocardiography has been used to quantify dyssynchrony in patients with heart failure, and to

predict immediate and long-term response to cardiac resynchronization therapy (CRT) [52]. In another study, strain imaging was again proven to be a useful tool for predicting left ventricular reverse remodeling after CRT [53].

Animal experimental studies

Jamal et al. [54] investigated the ultrasonic strain rate and strain as new indices to quantify the contractile reserve of the stunned myocardium, whereas Marciniak et al. [55] quantified regional deformation in the normal, stunned or infarcted myocardium.

Future developments

The ultimate development almost certainly will be a 3D assessment of left ventricular deformation with ultrasonic equipment capable of providing sufficient spatial and temporal resolution for the task.

In the field of 2D strain imaging, new software based on RF data and the implementation of a cross-correlation technique, along with increased computational power of ultrasound scanners, may be expected to enhance our imaging potential in the near future.

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