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Scientific Paper

High frame rate strain rate imaging of the interventricular septum in healthy subjects

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Abstract

Objective: In the present study the feasibility was assessed of a new strain rate imaging method with a very high frame rate of around 300 frames per second. *Methods:* Digital radio-frequency (RF) data were obtained in nine healthy subjects using a sector of 20–30° in an apical four chamber view. The RF data were analysed using a dedicated software package that displays strain rate images and profiles and calculates strain rate values. With the new method, it is possible to study events and spatial-temporal differences in the heart cycle with duration down to 3.5–3 ms, including the pre-ejection period and the isovolumic relaxation period. Since the interventricular septum (IVS) is of crucial importance for the left and right ventricular function, we assessed changes through the heart cycle of the strain rate in the IVS. *Results:* Mean peak systolic strain rate in the healthy subjects was -1.65 ± 0.13 s^{-1}. Mean peak diastolic strain rate during early filling was 3.14 ± 0.50 s^{-1} and during atrial systole 0.99 ± 0.09 s^{-1}. We found individual differences in the strain rate patterns, but in all subjects, the ventricular contraction started simultaneously in all parts of the septum. After the ejection period, the elongation started before aortic valve closure, in the midinferior septum and propagated towards the apex. *Conclusion:* High frame rate strain rate imaging makes it possible to study rapid deformation patterns in the heart walls.

Keywords: Strain rate values; Heart cycle; Rapid deformation patterns

1. Introduction

The regional function of the left ventricle can be visualised in real-time using the strain rate imaging method (Heimdal et al., 1998). Deforma-

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tion or strain of a tissue segment changes with time during the heart cycle. The rate of this deformation (the strain rate) is equivalent to the velocity gradient, and can be estimated using tissue Doppler data. Strain rate imaging is a new ultrasound method for evaluating regional function and agrees fairly well with standard echocardiography in grading regional wall function (Stoylen et al., 1999) as well as with reference to coronary angiography (Stoylen et al., 2000). Strain rate imaging provides both quantitative measurements of local deformation rates, and semi-quantitative information about the regional function.

With magnetic resonance imaging it is possible to measure wall deformation in three dimensions (Beyar et al., 1990), but the current frame rates of 20 frames/s (FPS) are too low to resolve myocardial mechanical events and peak velocities. An advantage of strain rate imaging is the high frame rate. With a decreased spatial resolution, a frame rate of 130 FPS is possible when acquiring tissue Doppler images of the whole left ventricle. Ultrasound tissue Doppler techniques can thus be used to reveal the complex spatial-temporal patterns of the deformation events in the ventricle during the heart cycle.

The longitudinal motion of the heart shows a descent of the base towards the apex during systole, with a reverse movement during the two main phases of diastole; i.e. early filling and atrial systole, while the apex remains almost stationary throughout the heart cycle (Jones et al., 1990). Although longitudinally directed fibers comprise only a small portion of the myocardial mass, several studies have shown that the systolic displacement of the atrioventricular plane towards the apex is an important component of the pump function of the left ventricle (Simonson and Schiller, 1989; Alam et al., 1992).

Strain rate imaging is optimal for studying longitudinal deformation of the left ventricle. This makes the method suitable for studying the time-course of mechanical events in the heart, as they generally progress along the length of the ventricle. The interventricular septum (IVS) regulates ventricular stroke volumes to maintain proper balance between systemic and pulmonic circulation (Lundback, 1986). The IVS is therefore of crucial importance for both the left and right ventricular function. Endocardial mapping in humans in sinus rhythm with normal left ventricles has also shown that one out of two endocardial breakthroughs of the ventricular activation sequence is on the midinferior septum (Cassidy et al., 1984). This means that the IVS is of particular interest when the timing of events is studied.

In order to study the complex patterns of events in the left ventricle during the heart cycle, the frame rate should be as high as possible. We have developed a new strain rate imaging method with frame rates around 300 FPS and spatial resolution equal to a tissue B-mode image of high quality. The aim of the current study was to evaluate the feasibility of this imaging technique to study timing and fast changes in regional function through both the systolic and diastolic phases of the heart cycle. Since the new method depends on a narrow image sector, we chose to study changes in strain rate in the IVS, as this wall is the most readily accessible, as well as the most interesting from a physiological point of view.

2. Materials and methods

2.1. Strain rate imaging

The strain rate (SR) is equivalent to the spatial gradient of the velocity. The strain rate is estimated from the velocity, v, at two points along the ultrasound beam, as described in Eq. (1):

$$\mathrm{SR} = \frac{v(r) - v(r + \Delta r)}{\Delta r}, \tag{1}$$

where r is the distance along the beam, and Δr is the small offset between the two points illustrated in Fig. 1. Negative strain rate corresponds to a shortening of the tissue segment, whereas positive strain rate corresponds to an elongation. The strain rate is visualised using a colour scheme where red corresponds to negative and blue corresponds to positive strain rate values. Strain rate values around zero are visualised using a green colour.

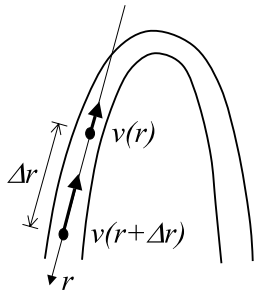

Fig. 1. Illustration of the velocities used in the definition of the strain rate estimate in Eq. (1).

2.2. Data acquisition and signal processing

A new acquisition technique was used, where the strain rate images were calculated from the same RF-data as the tissue B-mode images. A System Five scanner (GE Vingmed Ultrasound, Horten, Norway) with customised software was used to acquire digital radio-frequency (RF) data for off-line processing. The digital raw data were stored as complex base band signals where the in-phase and quadrature signal samples were represented as 16 bit integers. The data were transferred from the scanner and processed on a standard personal computer using MATLAB software (The MathWorks, Inc., Natick, Massachusetts, USA). The tissue B-mode images were calculated by amplitude detection of the complex signal followed by logarithmic compression to increase the dynamic range. The strain rate images were calculated using an estimator described by Heimdal (1999). The offset distance Δr in Fig. 1 was 6 mm, and we performed spatial averaging over 4 mm in the radial direction and across three neighbouring beams. The calculated image data were written to files on the GE Vingmed EchoPac format, and analysed using a prototype software package from GE Vingmed Ultrasound (Horten, Norway). The processing chain is summarised in Fig. 2.

When the strain rate values are calculated from the same pulse transmissions as is used for the B-mode image, there is a continuous stream of data with constant sampling intervals in the temporal direction. The strain rate estimates were thus calculated using a sliding window technique where each estimate was based on three data samples in the temporal direction. With this acquisition technique, the pulse repetition frequency (PRF) becomes equal to the B-mode frame rate and is thus considerably lower than what is used in conventional tissue Doppler techniques. Tissue velocity estimates calculated from these data would therefore suffer from aliasing artefacts. However, strain rate estimation involves calculation of a difference between two velocity estimates, and aliasing of the strain rate estimates can be avoided as long as the difference between the two velocities is below the Nyquist limit (Heimdal, 1999). Since the velocity difference is small for small offset distances, no severe aliasing artifacts in the strain rate images were experienced with frame rates exceeding 300 FPS.

Two-dimensional images of the IVS were obtained from a four-chamber apical view and the image sector was placed with the IVS in the centre giving a small angle between the IVS and the ultrasound beam. An image sector of 20–30° was sufficient to cover the whole width of the IVS, and frame rates above 300 FPS were obtained. Without using separate pulse transmissions to acquire the strain rate and B-mode data, and by using second harmonic data acquisition, the reverberation noise (clutter) was reduced compared to conventional strain rate data acquisition (Heimdal et al., 1998). Within the narrow image sector, we

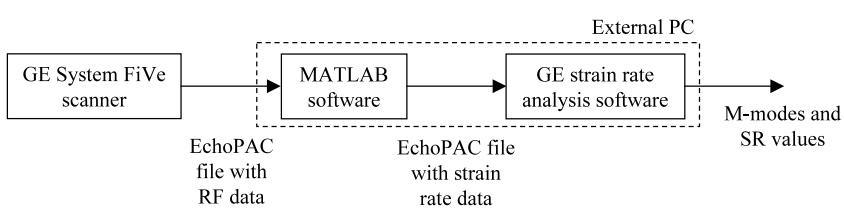

Fig. 2. Processing steps from data acquisition to strain rate (SR) analysis.

thus obtained high quality strain rate estimates with spatial resolution equal to the tissue B-mode image.

2.3. Study subjects

Nine healthy and physically active male volunteers (mean body surface area 1.96 ± 0.06 m^2; age 28.2 ± 2.6 years) without any evidence of cardiac disorders were studied. All volunteers were in normal sinus rhythm and gave informed consent to participate in the study. Standard phonocardiogram was used to determine the aortic valve closure and standard two-dimensional echocardiography was used to visualise and detect mitral valve opening since the anterior mitral valve leaflet was seen in the strain rate images.

2.4. Data analysis

The generated EchoPac files with strain rate data were analysed in a prototype software package from GE Vingmed Ultrasound. With this software we could create anatomically curved strain rate M-modes. Curves of strain rate versus time could also be plotted. Peak systolic strain rate, peak diastolic strain rate during early filling and during atrial systole were calculated as a mean value of velocity gradients over different parts of the IVS.

3. Results

In all the healthy subjects we were able to acquire strain rate images of the IVS with a higher temporal resolution than previously published. In all the subjects, the isovolumic contraction with shortening of the IVS started almost simultaneously at all levels from base to apex, followed by a slight, equally simultaneous recoil (Fig. 3). Similarly, the shortening during ejection started almost simultaneously. The elongation of the IVS started in the midinferior septum, and even before the isovolumic relaxation as shown by the phonocardiogram in Fig. 4. The tissue M-mode in Fig. 4 is included to be able to relate the strain rate values to the movement of the tissue

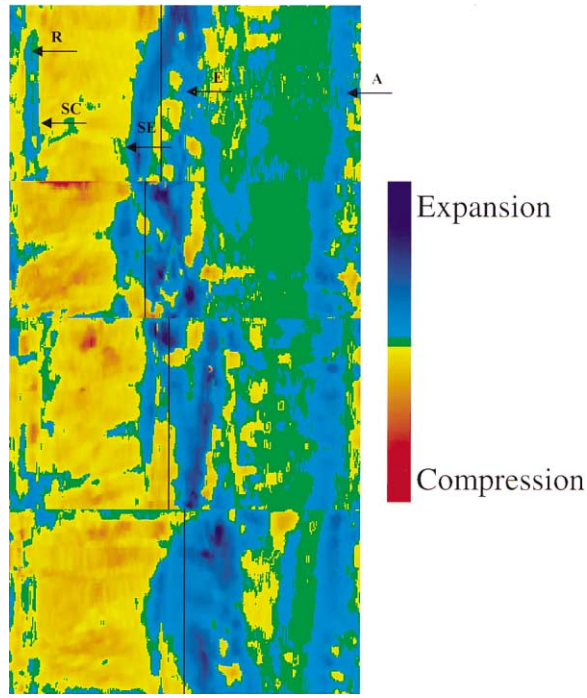

Fig. 3. Strain rate imaging M-modes of the IVS for one heart cycle in four healthy subjects. The apex is at the top of each image, and the basal part of the IVS is at the bottom. Start is at the R-wave of the QRS complex and the vertical line marks the mitral valve opening determined from the 2-dimensional image. To the right is the colour scale where cyan to blue corresponds to positive strain rate values, orange to red corresponds to negative strain rate values and green corresponds to strain rate values around zero. R, recoil; SC, start of contraction (ejection); SE, start of elongation; E, early filling phase; A, atrial systole.

structures during the cardiac cycle. The two main phases of diastole; i.e. early filling phase and atrial systole, could be recognised in all subjects. However, the deformation patterns were not similar in the subjects. In diastasis (the slow filling phase) the strain rate was almost zero. Mean peak systolic strain rate in the healthy subjects was -1.65 ± 0.13 s^{-1} ($\pm$SD). Mean peak diastolic strain rate during early filling was 3.14 ± 0.50 s^{-1} and mean peak diastolic strain rate during atrial systole was 0.99 ± 0.09 s^{-1}. In Fig. 5 we have compared a high temporal resolution (high frame rate) strain rate imaging M-mode with M-modes of lower frame rates. The M-mode was calculated from data acquired with a frame rate of 323 FPS

(a) and these data were then decimated down to frame rates of 65 FPS (b) and 25 FPS (c). Linear interpolation of the strain rate values was used to get as many data points in the decimated M-modes as in the original M-mode.

4. Discussion

This study shows the feasibility of studying rapid deformation patterns in the heart with a new strain rate imaging method with very high frame rates. The method has a superior temporal resolution as illustrated in Fig. 5. The frame rate of 25 FPS corresponds to the frame rate of data recorded on video tape. Comparing the M-modes in Fig. 5, we see that the details of the recoil at the start of systole are visualised only with the highest frame rate. We also see that the details in diastole are better captured with the highest frame rate.

It is a fundamental problem with the currently available strain rate methods in ultrasound that only one of the three normal component is available at one time, and that the six shear components cannot be measured (D'hooge et al., 2000). However, the clinical relevance of the method is promising (Stoylen et al., 1999, 2000, 2001) and with frame rates above 300 FPS the potential for quantification of the regional myocardial function is even greater than with the currently available methods. The spatial resolution with this method is equal to the B-mode resolution and is superior to the resolution that is possible with conventional tissue Doppler acquisition.

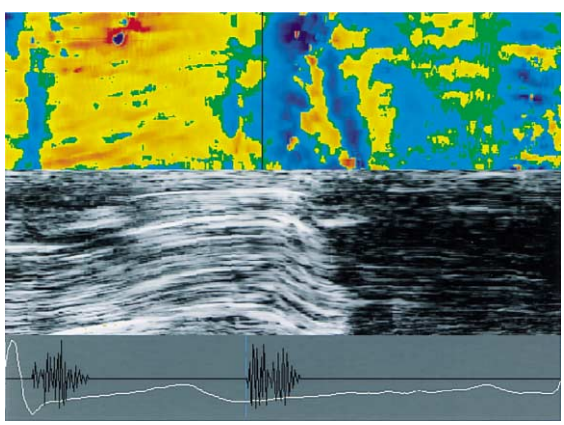

Fig. 4. Strain rate imaging M-mode of the IVS for one heart cycle in one of the healthy subjects in a similar way as in Fig. 3, but with the corresponding tissue M-mode in the middle and standard phonocardiogram in the bottom image. The vertical line marks the mitral valve opening while the blue vertical line in the phonocardiogram marks the aortic valve closure.

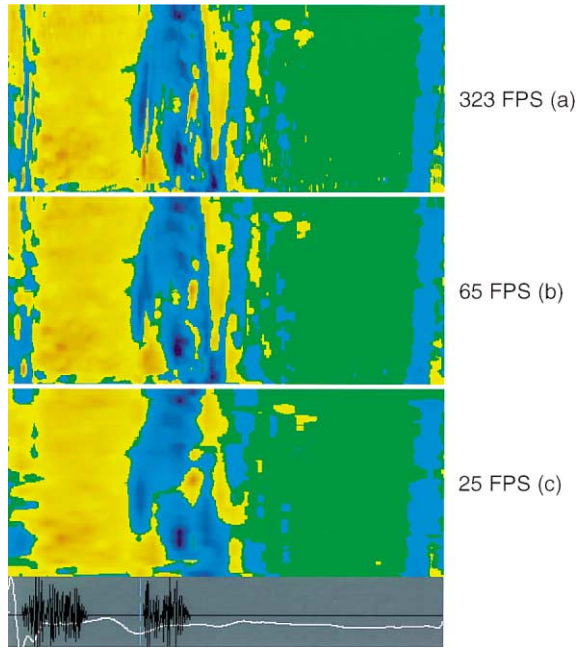

Fig. 5. (a) A strain rate imaging M-mode calculated from data with a frame rate of 323 FPS. (b) A strain rate imaging M-mode from the data in (a) but decimated down to 65 FPS. (c) A strain rate imaging M-mode from the data in (a) but decimated down to 25 FPS. Linear interpolation is performed to get an equal number of data points in all the M-modes. Green colour in these strain rate M-modes corresponds to strain rates values around zero (see the colour scheme in Fig. 3).

This study shows the feasibility of high frame rate strain rate imaging, with a possibility of studying events and temporal-spatial inequalities of down to 3 ms duration. It is important to realise that strain rate imaging shows only deformation. Shortening may either be contraction or elastic recoil after stretching. In systole, the isovolumic contraction started almost simultaneously judged by a shortening followed by a short simultaneous elongation which we believe is a recoil

before the ejection phase. As the ventricle shortens during the isovolumic contraction period, this is in accordance with the ventricle assuming a more spherical shape. The shape and geometry of the ventricle, arrangement of the fiber bundles in the wall and the nonuniform stress distribution at rest are all major determinants of the ventricular contraction–relaxation pattern. These structural features result in nonuniform configurational changes during systole, when the left ventricle changes shape (Brutsaert, 1987). In addition, the mechanical performance of the myocardium in different layers of the ventricular wall is not uniform. The nonuniform changes in the IVS are not so easily seen with the strain rate imaging method in the beginning of systole. Earlier studies have shown that the nonuniformity is more pronounced at higher loading conditions, being most marked during relaxation (Housmans et al., 1984).

The exact onset of diastole is controversial and varies according to the definitions described by Wiggers (1921) which includes isovolumic relaxation time in systole, by Brutsaert et al. (1984), Brutsaert and Sys (1989) which includes rapid filling phase in systole according to the triple control of relaxation, or using a clinical definition. Using the clinical definition, diastole is divided into two phases with an isovolumic relaxation period from aortic valve closure to mitral valve opening and an auxotonic period from mitral valve opening to mitral valve closure. These phases are demonstrated in Fig. 4 and this study shows that elongation starts even before the isovolumic relaxation period. In the auxotonic period this method demonstrates clearly the three events with rapid filling phase, slow filling phase (diastasis) and the atrial filling phase. The slow filling phase is earlier described to account for only 5% of the total filling (Opie, 1998). In the slow filling phase the peak strain rate fluctuated around zero and this indicates hardly any changes in length in this phase of diastole.

The new strain rate imaging method with very high frame rates show individual differences in the healthy subjects with regards to the contraction–relaxation patterns. A possible explanation is the nonuniform configuration of the myocardium, or the fact that we can only measure one of the strain components. In healthy subjects no fundamental differences in the myocardial structure or performance is expected. Strain rates in the directions transverse to the ultrasound beam as well as shear strain rates are not currently accessible in simultaneous recordings, but we still believe that strain rate imaging can be an important tool in physiologic research.

Mean peak systolic strain rate and mean peak diastolic strain rate of the rapid filling phase were somewhat higher than earlier published values obtained with lower frame rates (Stoylen et al. 2000; Voigt et al. 2000). The results were in accordance with other studies that mean peak diastolic strain rate of the rapid filling phase was higher than mean peak systolic strain rate (Stoylen et al. 2001). The reason for this is probably that the acceleration is higher in early diastolic filling phase due to less resistance to flow. The reason for somewhat higher strain rate values might be that it is easier to get peak strain rates with this new method if there are peaks of very short duration. It may also be that the lower noise level leads to recognition of peak strain rate that has previously been classified as noise. However, in this study, the centre of the sector is oriented along the IVS, while the previous studies utilised standard apical planes with the sector centre in mid-chamber. This may account for a larger angle and hence, lower measured values. High frame rate strain rate imaging therefore seems to have the potential of providing additional physiological information compared to standard strain rate imaging and of course compared to standard echocardiography. Its main advantages seem to be the high temporal resolution with respect to the location of the start and the propagation of events in the myocardium. This may be especially promising in the field of arrythmias and conduction abnormalities. With around 3 ms between the strain rate images, one should be able to track the mechanical events following the depolarisation wave.

A limitation of this method is the sector angle of 20–30°. This was sufficient to cover the IVS from an apical view, but makes it impossible to study more than one wall of the left ventricle at a

time. Increasing the sector width reduces either the spatial resolution or the frame rate.

In conclusion, high frame rate strain rate imaging makes it possible to study rapid deformation patterns in the heart walls.

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