SPECIAL ARTICLE

Definitions for a Common Standard for 2D Speckle Tracking Echocardiography: Consensus Document of the EACVI/ASE/Industry Task Force to Standardize Deformation Imaging

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Recognizing the critical need for standardization in strain imaging, in 2010, the European Association of Echocardiography (now the European Association of Cardiovascular Imaging, EACVI) and the American Society of Echocardiography (ASE) invited technical representatives from all interested vendors to participate in a concerted effort to reduce intervendor variability of strain measurement. As an initial product of the work of the EACVI/ASE/Industry initiative to standardize deformation imaging, we prepared this technical document which is intended to provide definitions, names, abbreviations, formulas, and procedures for calculation of physical quantities derived from speckle tracking echocardiography and thus create a common standard. (J Am Soc Echocardiogr 2015;28:183-93.)

Keywords: Echocardiography, Two-dimensional, Deformation imaging, Strain, Strain rate, Speckle tracking, Left ventricle, Myocardial, Standard, Definitions

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INTRODUCTION

This document represents a consensus statement from the EACVI/ASE/Industry Task Force to standardize deformation imaging ('the Task Force') to communicate standard physical and mathematical definitions of various parameters commonly reported in myocardial deformation imaging. It is aimed primarily at the technical engineering community and also interested clinicians. The document is not intended to explore the wide range of clinical applications of deformation imaging.

There is a growing body of evidence showing that the assessment of myocardial deformation by Doppler or speckle tracking techniques provides incremental information in the clinical setting.[1] Deformation imaging has been shown to provide unique information on regional and global ventricular function with some studies showing reduced inter- and intraobserver variability in assessing regional left ventricular (LV) function.[2] The main areas of application of these techniques have been assessment of myocardial mechanics, ischaemic heart disease, cardiomyopathies, LV diastolic dysfunction, and in detecting subclinical myocardial dysfunction in patients undergoing chemotherapy for cancer or in those affected by heart valve diseases.[3] Over the years, a number of software packages and algorithms have entered the market, but a practical limitation to the use of these techniques in routine clinical practice has been the significant variability that exists among vendors. Such a variability relates to several factors: differences in the terminology describing myocardial mechanics; the type of stored data which is used for quantitative analysis (e.g. proprietary formats vs. standard DICOM format); the modality of measuring basic parameters (tissue Doppler vs. speckle tracking); the definition of parameters (many vendors use proprietary speckle tracking algorithms or define different tracking regions for the same parameter); and the results output.[1,4-8]

Recognizing the critical need for standardization in strain imaging,[9] in 2010, the European Association of Echocardiography (now the European Association of Cardiovascular Imaging, EACVI) and the American Society of Echocardiography (ASE) invited technical representatives from all interested vendors to participate in a concerted effort to reduce intervendor variability of strain measurement.[10,11]

As an initial product of the work of the EACVI/ASE/Industry initiative to standardize deformation imaging, we prepared this document which is intended to provide definitions, names, abbreviations, formulas, and procedures for calculation of physical quantities derived from speckle tracking echocardiography and thus create a common standard. This document is purely technical and provides technical information only. Therefore,

- It is not intended to provide information about the clinical relevance of different measurements.
- It is not intended to suggest which parameters a product should preferably include.
- It is not intended to favour speckle tracking over other approaches for the echocardiographic quantification of myocardial function, such as tissue Doppler, which can provide comparable parameters of comparable relevance.

By providing clear definitions of the standard quantities that any software solution should report, the differences among different products should be limited to:

- *Technical:* accuracy and reproducibility of the proprietary approach to speckle tracking;
- *Marketing:* choices about how and what different products report;
- *Innovations:* further parameters or representations beyond what is reported in this document.

Readers interested in a more in-depth description of the mathematics and physics are referred to the structural mechanics literature.[12]

GEOMETRY DEFINITIONS

Region of Interest

The complete myocardial region of interest (ROI, Figure 1) is defined at end-diastole by:

- *Endocardial border:* the inner contour of the myocardium;
- *Epicardial border:* the outer contour of the myocardium;
- *Myocardial midline:* the middle ROI axis defined in the middle between inner and outer ROI contours.

Each of these contours can be either user-defined or generated automatically. In any case, where they are generated automatically, the user should be allowed to check them and, if needed, edit them manually. Extreme care should be taken in the definition of ROI, as inclusion of pericardium will result in reduction of measured strain. Different generations of the software appear to have different ROI defaults, and lack of user interaction will contribute to measurement variation.

Endocardial measurements pertain to the behaviour of the endocardial border and are represented there. Midline measurements, when available, refer to the behaviour of the middle ROI axis and are represented there. Epicardial measurements pertain to the behaviour of the epicardial border and are represented there. In case of tracking results that represent the average of measurements obtained over the full myocardial thickness, these are typically represented with the mid-wall line specifying this full wall reference.

When the epicardial border is not drawn, then the measurements typically refer to the single endocardial border and are presented there.

Task force recommendation: The key requirement for any software solution is that it explicitly states what is being measured and the spatial extent (in pixels or millimeters) over which the data is sampled for a given ROI. Measurement definitions can be: endocardial, midline, epicardial, or full wall.

Segment Definitions

Segments are the anatomical units of myocardium for which the results of the various strain analysis will be reported.

Apical Views. Topographic definitions of the myocardial ROI in apical views are shown in Figure 1, where:

- 'Left/right base': end points of the endocardial border.
- 'Midbase': midpoint between two basal end points of the endocardial border.
- 'Apex': the most distant from 'midbase' or a manually defined endocardial point.
- 'Left/right ROIs': ROI from the left/right base to apex.

The segments on the left and the right sides of the ROI are then defined such as to have the same end-diastolic length (the precise

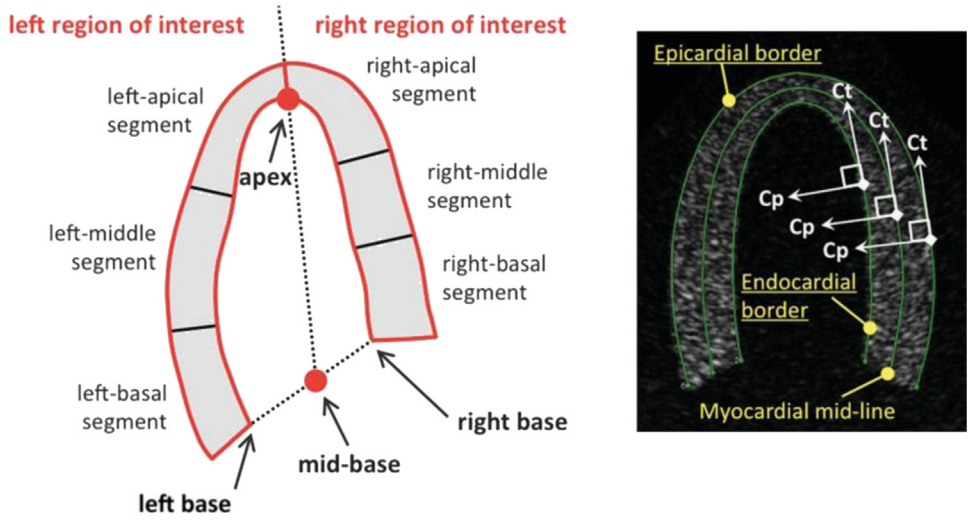

Figure 1 Right panel: Speckle tracking-derived parameters are reported with reference to the endo–epicardial or mid-myocardial line or to the full wall independent from the way they are achieved. The longitudinal or circumferential component of any parameter is then directed tangentially to the respective line (C_t), while the radial component is directed perpendicular to it (C_p). Left panel: Segmentation of the Region of Interest (ROI) in apical views as applicable for a 16 or 18 segment model. See text for details.

definition of end-diastole will be discussed below). Then, individual segments follow the underlying tissue and change their lengths during the various instants of the cardiac cycle. Thus, segmentation is performed as follows:

- Take the border at the user-defined or automatically selected frame,
- Define left and right ROIs,
- Divide each ROI into segments of equal length at the time point of end-diastole.

In the standard six-segment model (employed for global LV 16- or 18-segment models), the length of the three left segments is equal to the (left ROI length)/3, and the length of the three right segments is equal to (right ROI length)/3.

In case of a 17-segment model (which is not recommended for functional imaging, since the apical cap does not contract), the basal, mid, and apical segments have the length of 2/7th of the right and left ROI length, respectively, while the apical cap is composed of 1/7th of the right plus 1/7th of the left ROI.

Note that when segmental lengths are different, this fact must be taken into account when computing averages from segmental values.

Since the segments are presented with anatomical names corresponding to the LV wall the image refers to, it is necessary that the system recognizes or allows selection of the specific view under analysis. The system should also recognize or allow selection of whether an image is recorded as flipped left/right or inverted up/down.

Short-axis Views. Topographic definitions of the myocardial segments in short-axis views are shown in Figure 2. These views are approached differently from apical views: segments should be defined by measuring the angle relative to a centre of cavity, and imposing equality of angle coverage instead of tissue length. Alternatively, segments may be defined as having an equal border length at the end-diastolic frame—in similarity to apical views. Depending on the segmentation model used, the apical short-axis ROI is subdivided into six or four segments (Figure 3). The anterior insertion of the RV free wall is used as an anatomical reference.[13]

Segmentation Models. Segmentation models are built to reflect coronary perfusion territories, to result in segments with comparable myocardial mass, and to allow comparison within echocardiography and with other imaging modalities. Accordingly, a 17-segment model is commonly used (Figure 3, central panel).[13] The 16-segment model (Figure 3, left panel) divides the entire apex into four segments (septal, inferior, lateral, and anterior). The 18-segment model (Figure 3, right panel) divides the apex into six segments similar to the basal and mid-ventricular level. The last of these, the 18 segment model is simple and well suited to describe myocardial mechanics from two-dimensional (2D) data, but results in an overweighting of the apical-region (distal) myocardium in the overall score.

Task force recommendations: Segment definitions refer to the anatomy at the end-diastolic frame. If the segmentation is automatically proposed by the analysis software, a manual correction to modify the anatomy relative to the segments must be allowed to adjust for varying anatomy. Furthermore, the selection of a specific view, image inversion, or the possible left/right flip must be possible.

MEASUREMENTS

Velocity

Velocity is a vectorial quantity with a direction and amplitude. Velocities are commonly reported just as measured, but sometimes they are reported after subtracting the average velocity of the overall LV. While in some cases this subtraction may correct for overall LV translation, it may also mask or diminish segmental motion differences in others. For example, the difference between the fast inward motion of one LV wall and the slower motion of opposite wall will become smaller when the overall LV velocity is used for compensation.

In the apical views, the velocity vector is projected in two components:

- V_r—the radial component, which is perpendicular to the endocardial border (or any other reference border) and which is assumed to be positive when directed towards the cavity (contraction).
- V_l—the longitudinal component, which is tangential to the endocardial border (or any other reference border) and which is assumed to be positive when directed from the base towards the apex (see definition in Figure 1).

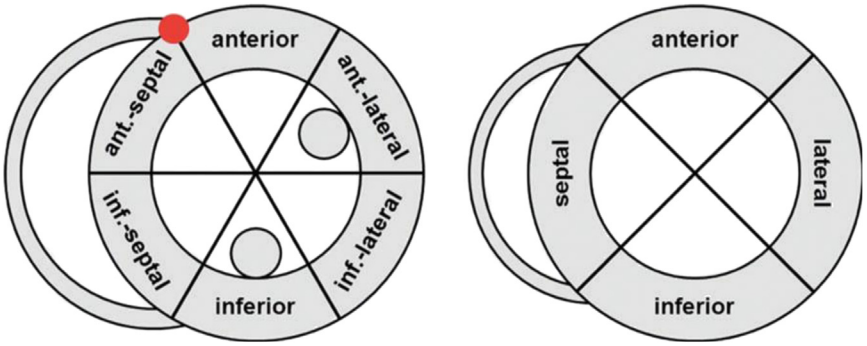

Figure 2 Segmentation of the ROI in the short-axis view. Left panel: Six segments are used for basal and mid-levels in the 16-segment model as well as for the apical level in the 18-segment model (608 segments). Right panel: Four segments for the apical level in the 16-segment model (908 segments). The *red dot* marks the anterior insertion of the right ventricular free wall, which defines the border between the (antero-)septal and the anterior segment.

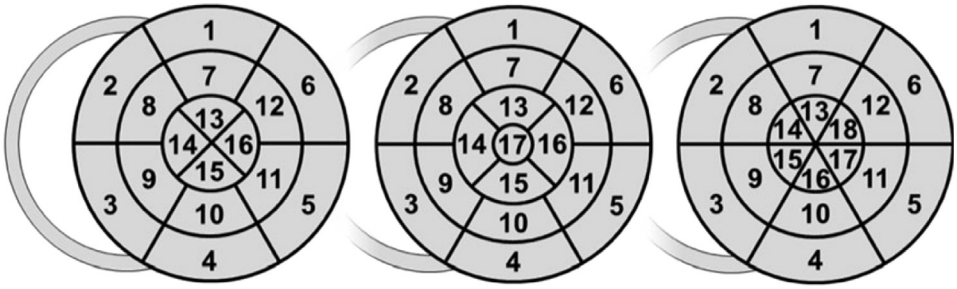

Figure 3 Schematic diagram of the different LV segmentation models. Left panel: 16-segment model. Central panel: 17-segment model. Right panel: 18-segment model. In all diagrams, the outer circle represents the basal segments, the mid one the segments at the mid-papillary muscle level, and the inner circle the apical level. In the 17-segment model, an additional segment (apical cap) is added in the centre of the Bull's eye. The anterior insertion of the right ventricular wall into the left ventricle defines the border between (antero-)septal and anterior segments (see Figure 2). Starting from there, the ROI is subdivided into six equal segments of 608. In case, the circle is subdivided into four segments, (as used for the apical level of the 16- and 17-segment models), the ROI is divided into four equal segments of 908, while the mid of the anterior segment in the four-segment and six-segment-segmentation have to coincide.

In the transversal/short-axis views, the velocity vector is projected in two components as well:

- V_r—the radial component, which is perpendicular to the endocardial border (or any other reference border) and which is assumed to be positive when directed towards the cavity.
- V_c—the circumferential component, which is tangential to the endocardial border (or any other reference border). The tangential (rotational) component is assumed to be positive when counterclockwise in a conventional short-axis view (i.e. probe on top of the image, as if looking from the apex to the base). The circumferential velocity may be reported as angular velocity (rotation rate). For this, velocity is normalized (divided) by the distance from the centre of the cavity and it is reported in radians per second or degrees per second. The instantaneous centre of the cavity is calculated as the 'centre of gravity' or 'centroid' with respect to the endocardium or any other reference border and can move during the cardiac cycle.

Task force recommendations: Myocardial velocities should be reported perpendicular or tangential to the defined border. Other ways of reporting need explicit indications. Likewise, the presence or absence of compensation for the LV translation must be explicitly indicated. The ability to switch the compensation for LV translation on or off is desirable.

Displacement

Displacement $X(t)$ is defined as the time integral of the corresponding velocity:

$$X(t) = \int_{ED}^{t} V(t')dt'. \quad (1)$$

Therefore, longitudinal, circumferential, and radial displacements are given by integration formula (1) using the longitudinal,

circumferential, and radial velocities, as described above, respectively.

Strain and Strain Rate

Strain (*S*) describes the deformation of an object normalized to its original shape and size. Strain rate (SR) describes the rate of deformation (i.e. how fast the deformation occurs).

A hypothetical one-dimensional object (a line) can only deform in one direction (it shortens or lengthens). Two common approaches to describe this length change are to use Lagrangian and natural strain.

For Lagrangian strain, a single reference length (L_0) is defined, against which all subsequent deformation will be measured. Lagrangian strain can therefore be calculated as follows:

$$S_L(t) = \frac{L(t)-L_0}{L_0}, \quad (2)$$

where $L(t)$ is the length at a given point in time and L_0 is the reference length at the reference time t_0, usually taken at end-diastole. Strain is a dimensionless entity, reported as a fraction or percent.

The Lagrangian strain rate is simply the derivative of Lagrangian strain:

$$SR_L(t) = \frac{dS_L}{dt} = \frac{1dL}{dt}. \quad (3)$$

Natural strain, on the other hand, employs a reference length that changes as the object deforms. It therefore describes the instantaneous length change. It provides an instantaneous absolute definition of natural strain rate that is independent of reference times. Natural strain rate is thus the temporal derivative of natural strain and describes the instantaneous rate of length change:

$$SR_N(t) = \frac{dS_N(t)}{dt} = \frac{1}{L(t)}\frac{dL}{dt}. \quad (4)$$

This equation differs from Eq. (3) in having a denominator that varies continuously. Natural strain can then be calculated by integrating Eq. (4):

$$S_N(t) = \int_{t0}^{t} SR_N(t)dt = \int_{t0}^{t} \frac{1dL(t)}{L(t)dt}dt = \ln\frac{'L(t)'}{L_0}. \quad (5)$$

Note that the reference length $L(t)$ is constantly changing in contrast to Lagrangian strain [Eq. (2)], which always refers to L_0 (Figure 4).[14]

The above-mentioned concepts apply in principle to all three one-dimensional (longitudinal, circumferential, and radial) displacement and strain components.

Rotational Mechanics

The rotational deformation of the LV around its long axis is described by two parameters.[15] The difference in the systolic rotation of the myocardium in an apical and basal short-axis plane is commonly referred to as twist and reported in degrees. If normalized to the distance between the respective image planes, it is referred to as torsion and consequently reported in degrees/cm. Although the latter is physically more precise, it is impossible to measure it with confidence using two-dimensional echocardiography. Twist can be obtained, but is imperfectly defined since the exact position of the image planes relative to the heart and relative to each other is unknown. The temporal derivative of twist is referred to twisting and untwisting rate and given in degrees/s.

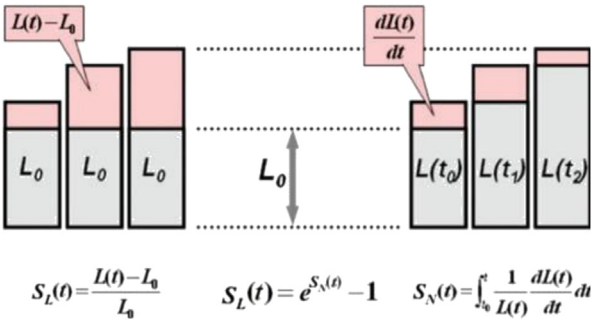

Figure 4 Graphical representation of the difference between strains. Left panel: Lagrangian strain S_L relates the actual length always to the baseline length of the object. Right panel: Natural strain S_N relates the instantaneous length changes to the variable instantaneous length. Modified from Voigt [14].

Task force recommendations: Twist and torsion describe the rotational deformation of the LV around its long axis. Both parameters are poorly defined in 2D echocardiography and caution is urged in their use.

Baseline Drift. The calculation of both displacement and strain from either tissue Doppler or speckle tracking data is influenced by small measurement errors, which result in a baseline drift (Figure 5). A correction of this unwanted drift can be done in many possible ways and may be included in the analysis software to ensure that the displacement or strain returns to zero after one cardiac cycle.

It is also true that the displacement might not be zero at the end of the cardiac cycle due to any LV global translational motion, which is not synchronized with heart cycle (e.g. breathing).

Task force recommendation: Since intensive drift correction may mask poor tracking, applied drift compensation should be indicated to the user and options for turning it off or on should be available.

Tracking Quality. Speckle tracking works in general better along the ultrasound beam than across beams. Furthermore, due to the beam divergence with increasing depth in a sector image, tracking across beams works better in regions close to the transducer than in the far field of the image.

Tracking quality may be suboptimal if regions of the myocardium are poorly visualized, if stationary image artefacts (reverberations) compromise speckle recognition or if spatial or temporal resolution of the image acquisition is insufficient.

Task force recommendations: Analysis software should offer an automated measure of tracking quality. Furthermore, the user should be always offered a display, where he/she is able to visually check tracking quality by comparing the underlying image loop with the superimposed tracking results, along with the actual curves derived from that tracking.

Regularization. Several vendors use models of normal cardiac deformation, spline-functions, or other types of spatial and temporal smoothing for the regularization of tracking results. Excessive regularization, however, may reduce the resolution of the tracking results or may even compromise the validity of data. Proprietary filters may contribute to the variation between vendors. Besides that, user-defined regularization settings have become an important source of variation using the same machine.

Task force recommendations: Analysis software should inform the user about measures, which are applied for regularization. Regularization should be limited to the necessary minimum. Options to control

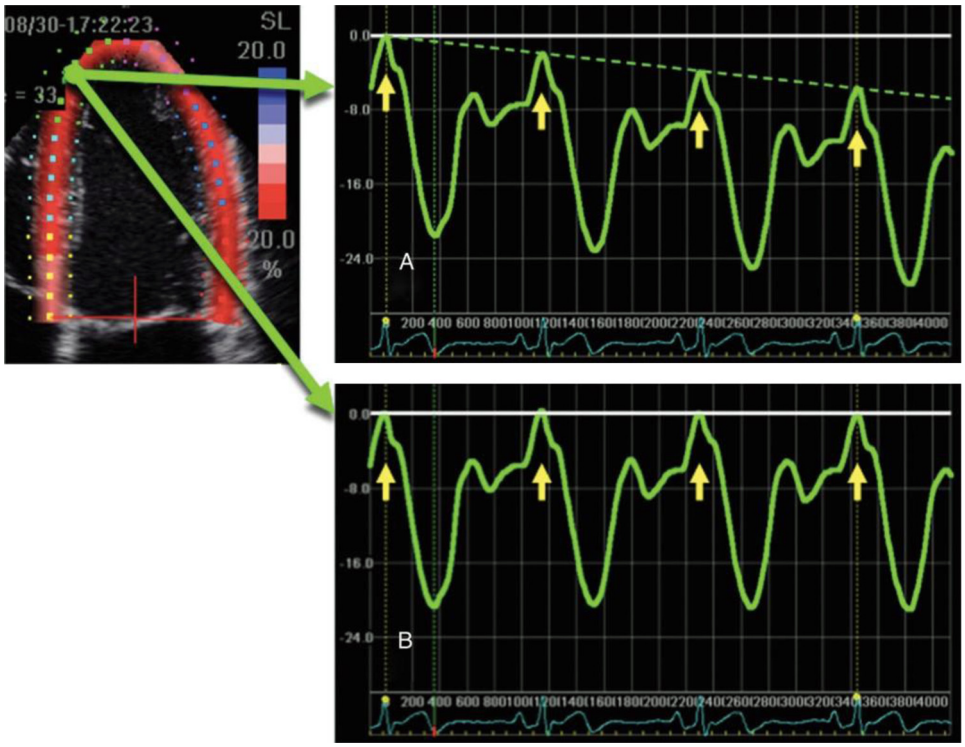

Figure 5 **(A)** Tracking-derived strain curve with drift (*dotted green line*). **(B)** Drift is compensated by subtracting the averaged drift component from the curve. Since the ECG trigger is often used as time reference, the curve returns to zero at each QRS (*yellow arrows*).

regularization settings should be available to the user. A record of the processing settings is prudent in longitudinal studies.

Multidimensional Deformation. Until now, all the concepts have been exposed assuming that the deforming object has only one dimension. However, if the deforming object is two-dimensional, then the deformation is not limited to shortening and lengthening only. A 2D object can deform perpendicular to the borders (Figure 6A and B). Furthermore, an object can deform parallel to a border (Figure 6C and D). This type of deformation is called 'shear strain'.

To define the deformation of a 2D object in a comprehensive way, all four strain components are written in a single matrix, which is referred to as the strain tensor:

$$\begin{pmatrix} S_{XX} & S_{XY} \\ S_{YX} & S_{YY} \end{pmatrix} \tag{6}$$

where the diagonal elements (*xx* and *yy*) reflect linear (or normal) strain and the off-diagonal elements are shear strain.

Three-dimensional echocardiography investigates the heart as a three-dimensional object. Applying the concepts of normal and shear strain, nine different strain components can be distinguished, namely three linear strains (*xx*, *yy*, and *zz*) and six shear strains (*xy*, *xz*, *yx*, *yz*, *zx*, and *zy*).

For regional function analysis, the coordinates *x*, *y*, and *z* can be replaced by the longitudinal, radial, and circumferential ones of the heart. From this, it follows that the torsion of LV is reflected and can be described by the longitudinal-circumferential shear (lc-shear) of the LV myocardial wall.[16]

Furthermore, from the principle of conservation of volume for incompressible material, it can be concluded that if two linear strain components are known, the third can be calculated.

Specific Aspects of Echocardiographic Strain and Strain Rate Measurements

Natural vs. Lagrangian Strain. There are circumstances where it is more appropriate to use Lagrangian strain than natural strain and others where the opposite is true. A natural strain rate calculation is better suited for use with tissue Doppler imaging, since the reference length is different at each interrogation time point (each colour tissue Doppler frame) and so will not be the same as at the reference time point. On the other hand, speckle tracking will lend itself more readily to the calculation of Lagrangian strain, since the baseline length is always known and can easily be used as a reference. Fortunately, natural and Lagrangian strains are related so that one can be converted into the other:

$$S_L(t) = e^{S_N(t)} - 1, \tag{7}$$

$$S_L(t) = \ln(s_L(t) + 1), \tag{8}$$

$$SR_L(t) = \frac{1}{e^{(-S_N(t))}} SR_N(t), \tag{9}$$

$$SR_N(t) = \frac{1}{SL(t) + 1} SR_L(t). \tag{10}$$

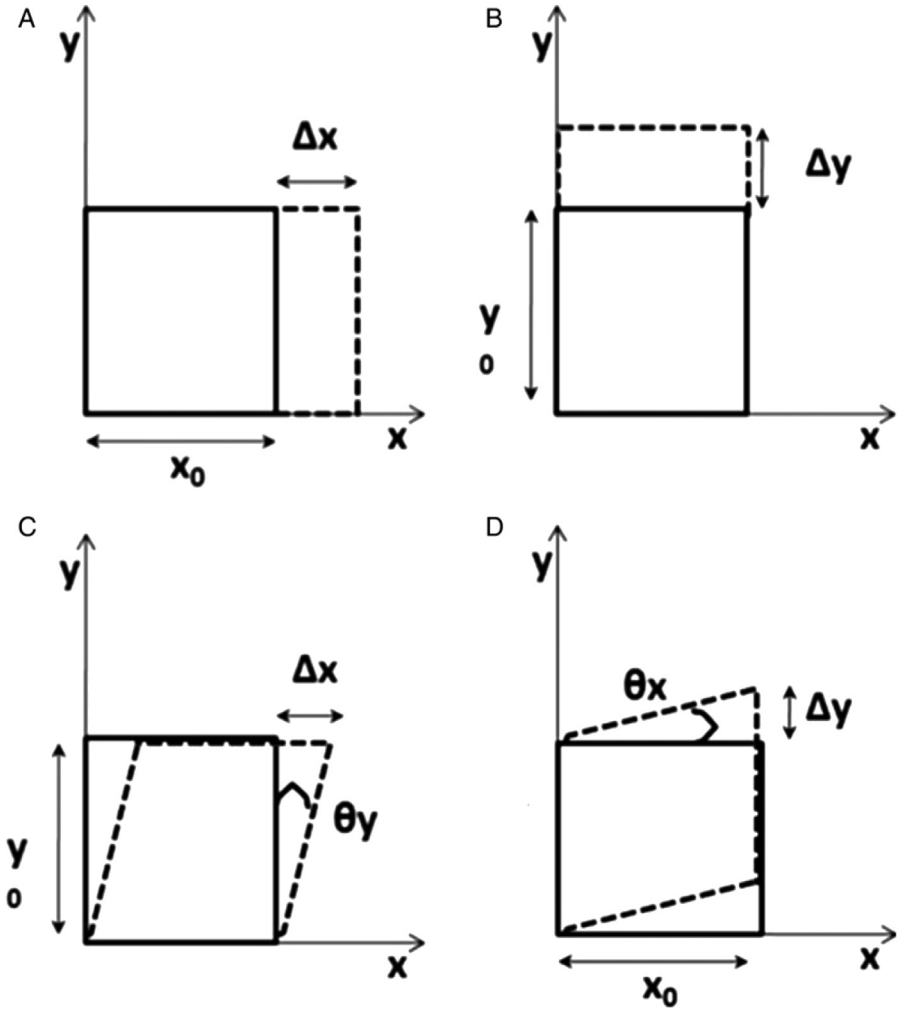

Figure 6 The deformation of a 2D object can be described by four strain components: two normal strain components (**A** and **B**) and two shear strain components. The shear strain components are characterized by the angles θx and θy. Modified from D'hooge *et al*. [15].

If the extent of deformation is small (~5–10%), Lagrangian and natural *S* values are close. However, for the large myocardial deformations which may occur during rapid filling and ventricular ejection, the differences become significant (Figure 7, left panel).[15] Strain rate shows even greater discrepancies (Figure 7, right panel).

Task force recommendations: Speckle tracking software packages should commonly report Lagrangian strain (*S*). Natural strain rate (SR_N) is commonly reported when using tissue Doppler, but can also be derived from speckle tracking by conversion from the Lagrangian strain rate. The reported type of strain or strain rate (i.e. Lagrangian vs. natural) must be indicated by any software package.

Timing of Mechanical Events

End-diastole. Since cardiac function is a cyclic process, the selection of a reference point in time ('beginning of the cardiac cycle') is arbitrary. In order to report displacement or deformation, however, a time point must be defined, at which the reference position (displacement) or reference length (strain) can be measured. End-diastole is conventionally used for this purpose.

End-diastole is commonly characterized by the closure of the mitral valve (i.e. the frame before mitral valve completely closes is called end-diastole). Other events which are time-related to mitral valve closure may be used as a surrogate, such as the beginning of the QRS complex in the ECG, ECG R-peak, the largest diameter or volume of the LV, or the peak of the longitudinal global strain curve. All surrogate time markers may be suboptimal under certain circumstances. Mitral valve closure and ECG parameters may dissociate in patients with conduction delays. Similarly, diameter- or strain-based parameters may fail in regional dysfunction. Volumetric measurements require at least two or three apical image acquisitions.

Task force recommendation: As a compromise between feasibility and accuracy, analysis software commonly uses the peak of the QRS complex to define end-diastole, but it should also offer the user the option to over-rule this definition if deemed inappropriate in a certain pathology or when analysing other cardiac structures than ventricles (e.g. atria). In any case, the user must be informed about the time reference which is used.

End-systole. End-systole coincides with aortic valve closure, which can be visualized in the parasternal or apical long-axis view or by detecting the closure click on the spectral tracing of the pulsed-wave Doppler of aortic valve flow. Potential surrogate parameters are the nadir of a global strain or volume curve.

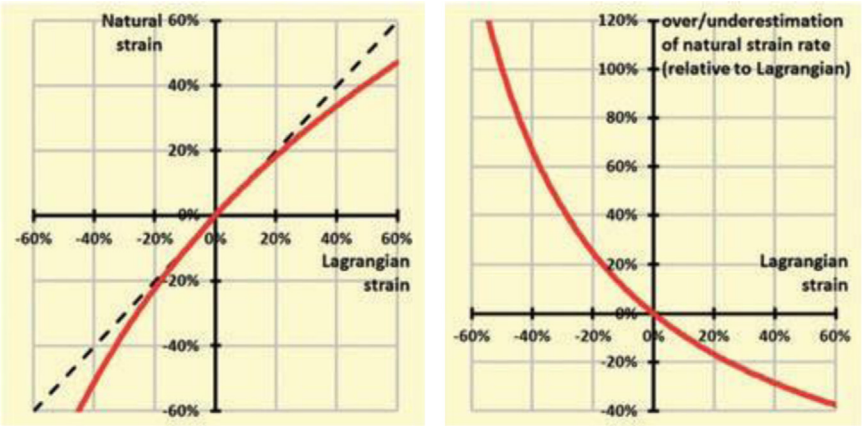

Figure 7 Comparison between Lagrangian and natural strain. Left panel: Plot showing the relationship between Lagrangian and natural strain (*solid red line*). Beyond about ±15%, the divergence from the line of identity (*black dotted line*) becomes relevant. Right panel: Plot showing the degree by which natural strain rate will over or underestimate Lagrangian strain rate, depending on the instantaneous Lagrangian deformation. For example, if the Lagrangian strain is −20% and the Lagrangian strain rate is +10%/s, then the natural strain rate will be 25% greater than this and show a value of +12.5%/s. If instead the Lagrangian strain is +20% (with the same strain rate), then the natural strain rate will be 16.7% less and show a value of +8.33%.

Task force recommendation: The user must be informed about the time reference, which is used to define end-systole and be offered the opportunity to over-rule this definition if deemed necessary according to the pathophysiological situation.

Measurement Points. Clinically relevant strain values along strain curves are, but are not limited to:

- End-systolic strain: the value at end-systole (the way end-systole is defined should be specified);
- Peak systolic strain: the peak value during systole;
- Positive peak systolic strain: a local myocardial stretching, sometimes occurring to a minor extent in early systole, or as a relevant deformation in regional dysfunction;
- Peak strain: the peak value during the entire heart cycle. The peak strain may coincide with the systolic or end-systolic peak, or may appear after aortic valve closure. In the latter case, it may be described as 'post-systolic strain'.

The mentioned values are shown in Figure 8. Others may be introduced when relevant for specific clinical pathophysiological situations.

Task force recommendations: End-systolic strain (ESS) should be reported as a default parameter for the description of myocardial deformation. Other parameters may be reported in addition. Reported parameters need to be labelled in a way that the definition of the parameter is clear to the user.

Global and Segmental Values

The previous definitions of strain and strain rate permitted to define the value at every point along the selected myocardial line, at every instant during the cardiac cycle. Of special interest in cardiology are strain and strain rate segmental and global values.

The segmental strain or strain rate is defined as the average value in the segment. This definition applies to any strain or strain rate component.

The global strain or strain rate is calculated by using the entire myocardial line length while computing the deformation. Alternatively, global strain can also be computed by averaging the values computed in a number of points within the myocardial line, or by averaging the values computed at the segmental level from the same frame. These last two methods are mathematically equivalent to the former when the points/segments used for the averaging are equi-spaced at the reference frame. Alternatively, a weight proportional to the pertinent length of every segment at end-diastole must be employed. Calculations which average peak values obtained at different points in time are not compatible with the aforementioned definition.

If global parameters are calculated by segmental averaging, the badly tracked segments can be excluded; in this case, a reproducible way to properly differentiate good and bad tracking results would be desirable. However, when global strain or strain rate values are calculated by segmental averaging and some badly tracked segments are excluded (no more than 1 per view), results will differ depending on which segment will be excluded, as apical segments usually show greater strain values than basal segments.

Longitudinal strain may be calculated as an endocardial strain, midline strain, epicardial strain, or averaged over the entire cardiac wall. There is currently insufficient evidence to favour one way over another. An analysis software should clearly declare which type of strain is reported.

Table 1 summarizes a summary of commonly used parameters to describe myocardial motion and deformation.

Note that names and abbreviations of circumferential and longitudinal parameters should include, as a subscript, information to state the myocardial layer they refer to or if they refer to the average over the full myocardium. Global parameters should include, as an index, information if they refer to a single image plane or the entire ventricle.

Task force recommendations: The global strain or strain rate should be calculated by using the entire myocardial line length or using alternative methods (i.e. averaging the values computed in a number of points within the myocardial line, or by averaging the values computed at the segmental level), which are mathematically equivalent. When global strain is computed in a manner not equivalent to using the entire myocardial line length, this must be explicitly stated to permit comparability. The location where global strain values were measured (i.e. measured at the endocardium, midline, or averaged over the entire cardiac wall) must be explicitly reported by the software.

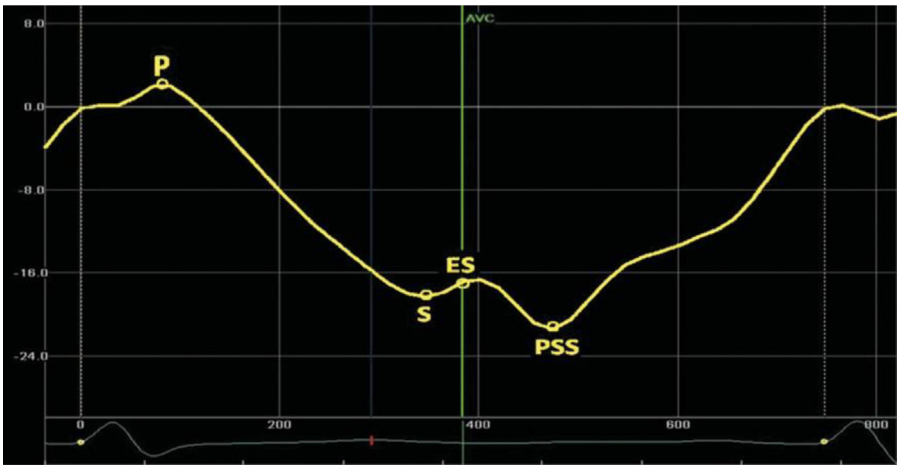

Figure 8 Longitudinal strain curve with a selection of strain values at clinically relevant timings. P, peak positive strain; S, peak systolic strain; ES, end-systolic strain; PSS, post-systolic strain. The *yellow dashed line* indicates begin of QRS; the *green dashed line* aortic valve closure (AVC).

Serial and Cross-sectional Comparison of Strain Values

The generally negative sign of longitudinal and circumferential strain can lead to confusion when comparing patients or discussing serial values, since deterioration in LV function results in a counterintuitive increase in the arithmetic value of strain. Accordingly, the task force feels that when comparing strain values, one should implicitly consider the absolute value of strain.

Task force recommendations: We recommend that all references to strain changes actually consider the absolute value of the number, so that increases in GLS mean that the number is becoming more negative, and decreases are observed when LV function deteriorates and GLS becomes less negative. Any exception to this convention should be explicitly stated. The notation of strain values as numbers should always include the sign.

DISCUSSION

The members of the EACVI/ASE/Industry Task Force to standardize deformation imaging have reached a consensus about the content of the present document to provide standardization of the definitions and the calculation of the various quantities usually reported in myocardial deformation imaging. The main purpose of this document is to provide the theoretical basis to explain physiological significance and mathematical computation of the various parameters to clinicians interested in using deformation imaging for both research and clinical work and to ensure a common background for the different realizations of this echocardiographic technique.

However, in addition to physiological variation in beating hearts, differences between similar calculations performed by different imaging equipment and different software are still possible for numerous known technical reasons.

Imaging Systems

Different image systems present technical differences in terms of spatial (number of beams per image sector, pixel size) and temporal (acquisition frame rate) resolution of the acquired images. Accuracy of the tracking technique is inversely proportional to the pixel size. In addition, for any given tracking technique and heart rate, there is an optimal frame rate for best tracking and this data should be known by the users. Finally, accuracy is reduced if the increase in frame rate is obtained at the cost of decreased spatial resolution (decreased number of beams per image sector).

Frame rate determines how short mechanical events may be still resolved and correctly displayed. Clinical literature reports that acquisition frame rates ranging from 40 to 80 Hz have been widely used to measure motion and deformation at normal heart rates.[1] As mechanical events become shorter with an increasing heart rate, the frame rate needed to resolve a particular physiological event becomes higher. Therefore, the frame rate should be increased with the increase in heart rate especially for paediatric studies, exercise, and pharmacological stress exams with agents (e.g. dobutamine), which increase heart rate. There is no sufficient evidence to recommend a certain number, but as an extrapolation, the standard frame rate for resting heart rate should be increased proportionally with the expected heart rate. While motion and deformation are less demanding, time-dependent parameters, such as velocity and strain rate, requires high frame rates (>100 fps) even at rest to resolve all relevant events. Since it might be very challenging with current imaging systems to adapt this further to higher heart rates, the user should make use of these parameters at higher heart rates only when their validity has been checked.

In addition to image resolution, overall image quality greatly influences the quality of tracking. The method works best when all the walls are visible in all the frames included in the clip, while a degradation of results is expected when the myocardium is temporarily not visible in some segment.

Images of varying quality and different spatial and temporal resolution produce a potential variability in the results of deformation imaging.

Software Application

The methods of calculation described above allow computation of various metrics once the geometry of the tissue is known in all frames after the tracking procedure has been performed. Different software applications, however, employ different tracking techniques, which detect tissue motion with different accuracy and reliability. Thus, the output depends on the performance of the specific tracking

Table 1 Recommended names, abbreviations, and units for 2D speckle tracking-derived parameters

Parameter	Definition	View for data acquisition	Abbreviation	Unit
Longitudinal velocity	Respective motion or deformation component parallel to the reference contour, viewed in from base to the apex	All three apical views (recommended) and parasternal long-axis view (not recommended in routine clinical practice)	V_l	cm/s
Longitudinal displacement			D_l	mm
Longitudinal strain rate			SR_l	1/s
Longitudinal strain			S_l	%
Radial velocity	Respective motion or deformation component perpendicular to the reference contour, viewed from the contour towards the LV cavity	All three apical views and parasternal short-axis view (recommend) and parasternal long-axis view (not recommended)	V_r	cm/s
Radial displacement			D_r	mm
Radial strain rate			SR_r	1/s
Radial strain			S_r	%
Circumferential velocity	Respective motion or deformation component tangential to the reference contour, perpendicular to the LV long axis, with counterclockwise orientation when viewed from the apex. Angular components refer to the centre of gravity of the LV within the image plane	Short-axis views only	V_c	cm/s
Rotation rate			RotR	8/s
Circumferential displacement			D_c	mm
Rotation			Rot	8
Circumferential strain rate			SR_c	1/s
Circumferential strain			S_c	%

algorithms, on the level of spatial or temporal smoothing involved, and whether they are optimized for certain conditions, e.g. particular image acquisition settings. Furthermore, user-defined settings may influence the comparability of strain and—in particular—strain rate data.[17]

The main activities of the EACVI/ASE/Industry Task Force to standardize deformation imaging are focused to promote concerted initiatives to reduce intervendor variability of strain measurement. At present, intervendor agreement has been tested both on synthetic data sets and during live testing on humans.[10,11]

Limitations of the Technique

There are a number of intrinsic limitations to strain imaging. First, a basic assumption underlying 2D speckle tracking is that in-plane displacements of tissue correspond to the displacements of local patterns in the gray scale distribution of a 2D echocardiographic clip. However, it should be appreciated that this may not always be the case. For example, through-plane displacement of a tapering, helically structured or otherwise obliquely angulated form could be misinterpreted, both visually and by speckle tracking, as in-plane deformation or displacement in a 2D sequence of images. This off-plane limitation issue is known to be more critical in short axis than in apical views.[18,19] When suspicion of artefacts due to through-plane motion arise, 3D imaging could be used, if available, to verify this and avoid potential misinterpretation. The user should take into account that 3D speckle tracking has the same intervendor variability limitations that affect 2D speckle tracking[20] and has lower temporal and spatial resolution than 2D imaging.

The local frame-by-frame tracking is based on the search of a maximum likelihood between two local speckle patterns in two consecutive frames. All kinds of ultrasound noise reduce the tracking quality. Good image quality enhances the clarity of speckle patterns and improves accuracy and robustness of their detection. It is therefore important to note that the acquisition of standardized image planes in optimized quality is essential for reducing inter- and intra observer variability of tracking data.

The most critical limitation in the tracking techniques is the temporal stability of tracking patterns. The ultrasound speckle patterns are generated by the interference of the ultrasound waves reflected from tissue structures. Speckle patterns are not stable temporally not only due to through-plane motion, but also due to physiological changes of living tissue structures and changes of interrogation angles between moving tissue and ultrasonic beam. The accumulation of small random errors in detection of speckled patterns along the tracking process can lead to inaccurate tracking results.

CONCLUSIONS AND NEW BEGINNINGS

This strain standardization Task Force was initiated by EAE (presently EACVI) and ASE to develop an academia–industry consortium for achieving consensus on a list of standard definitions and nomenclature for the clinical parameters evaluated with 2D speckle tracking

technology. This marks one of the first steps in reducing intervendor differences and ambiguities in the strain algorithms. We strongly encourage clinicians and researchers to remain aware of the potential variations in techniques before considering a given quantitative difference as clinically meaningful. The task force recognizes that the progress in research and technical development may require reconsiderations; however, the definitions provided herein are expected to provide a valid basis that allows a better comparison between vendors and the development of more meaningful clinical applications.

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