Feature Tracking in Cardiac Magnetic Resonance Imaging
Abstract

Purpose: An accurate and practical method to measure parameters like strain in myocardial tissue is of great clinical value, since it has been shown that strain is a more sensitive and earlier marker for contractile dysfunction than ejection fraction (EF). Current CMR technologies for strain assessment are time consuming and difficult to implement in daily clinical practice. Feature tracking is a technology that can lead to more automatization and robustness of quantitative analysis of medical images with less time consumption than comparable methods.

Methods: Feature tracking (FT) technique was tested in computed-generated phantom models with varying image resolution, frame rate, and tissue dynamics. The phantom motion was known exactly and allowed a careful check of the intrinsic accuracy. The same technique was also tested in 7 patients undergoing clinical MRI: 2 normal, 1 non-ischemic, 2 ischemic cardiomyopathy, 2 hypertrophic cardiomyopathy. Analysis of clinical CMR involved calculation of ejection fraction (EF) as well as radial, circumferential and longitudinal strain.

Results: The method is well suited for tracking left ventricular myocardium in radial, circumferential as well as longitudinal direction. In addition, strain can be assessed at epicardial, endocardial and mid myocardial levels.
Conclusions: This new method offers a robust and time saving procedure to quantify myocardial tissue displacement, velocity and deformation parameters on regular SSFP sequences of CMR imaging. It therefore can be implemented easily into clinical practice.

Key words: feature tracking, strain, displacement, CMR

Introduction

Automatic detection of borders is a fundamental issue in image analysis. In cardiac imaging, the possibility of an automatic detection of the endocardial border in the imaging of the left ventricle would give objective measurement of the ventricular volumes, and myocardial deformation (strain). This was accomplished in echocardiography with speckle tracking technique. The development of reliable methods for the automatic border detection is a challenging task that has not received a generally reliable solution in cardiac magnetic resonance (CMR). In fact, in clinical practice, borders are either drawn manually by the operator or software detects interface between myocardium and cavity (1,2). In current article we introduce a different approach where the borders are not “detected”, rather they are “tracked”, i.e. followed in time, starting from one reliable existing instantaneous trace that is commonly -but not necessarily- manually drawn by the experienced operator over one single frame. The individual points composing such a first reliable trace are followed in time by searching the same features that are about one point in its neighborhood in the following frames. The tracked features can be the cavity-tissue boundary or anatomical elements that are different along the tissue. They are found by methods of maximum likelihood in two regions of interests between two frames.
The local frame-to-frame displacement is equivalent to evaluating the local velocity (ratio between displacement and time interval). The automatic evaluation of the velocity at a point is determined from comparison of the displacement of the image data about such point in two consecutive frames. Such methods have been used, in several different formulations, in many research fields. They fall in the general category known as *Optical Flow*, in advanced image analysis (3,4). They are commonly referred as *Speckle Tracking* in echographic imaging when such velocities are used to follow physiological motion (5,6) but also apply to any other image modality such as CMR where these methods are referred as feature tracking or border tracking.

**Material and Methods**

**Feature Tracking Method**

Endocardial or epicardial border of a 2D CMR cine is manually traced on one arbitrary frame (Fig. 1). Mid-myocardial features can be traced as well. Such border is then defined as a sequence of $N$ points, identified by their coordinate pairs $(x_i, y_i)$ with $i=1\ldots N$. The border tracking proceeds by tracking each single point, such a tracking is based on a hierarchical algorithm at multiple scales and by a combination of 1D tracking techniques, which guarantee higher accuracy, and 2D tracking, which is necessary to properly detect the 2D spatially extended features.

As a first step, in order to capture the large geometrical displacement of the border, the tracking is performed in the direction orthogonal to the border itself where the cavity-
tissue boundary is best recognizable. The tracking along this direction is performed by using the method of transmural cuts as follow (Fig. 2). A line crossing the wall, passing through the point and orthogonal to it is drawn. The pixels taken along the transmural line are placed in columns, each column corresponding to one frame of the sequence of images. In this way the evolution along a transmural cut can be represented for all instants at once in a two-dimensional representation where one axis is the distance along the line and the other axis is the time (Fig. 3). This representation is similar to what is referred to as an M-mode in Echocardiography, in CMR it corresponds more to the “scout” function. To improve the quality of the analysis, in the case of poor images with a low signal to noise ratio, the space time representation is built using a line for the transmural cut with a thickness of 5 pixels, this number is chosen arbitrarily as a minimum (odd) number that guarantees clearing noise of the size of 1 or 2 pixels. The border tracking is then performed along the space-time image.

In a second step, to account for the 2D displacement of the border, a standard 2D tracking (optical flow-based, see ref 4, or ref 6 for a straightforward application in echocardiography) is performed, for each point independently, on a MxM moving window that is always centered on the previously estimated border point. 2D tracking is performed in two steps, where half of the first estimation is employed to center the moving windows in the second tracking passage. The window size is then reduced from 32 to 16 pixels in the two sequential passages.
To improve the accuracy of the motion along the border, that is required to estimate rotation and torsion, the 1D tracking is performed along space-time images built from thick cuts "parallel" to the curved border (Fig. 3). At each point, independently, the pixels taken along the moving border, centered at the moving border points, are placed in columns, each column corresponding to one frame of the sequence of images. To improve the quality of the analysis, and to best capture the features at the border the line is extended of 5 pixels into the tissue (sub-endocardium) in coherence to what previously done for transmural cuts. The border tracking is then performed along the space-time image with the same procedure described above. To ensure the spatial coherence in the tracked border, a 3 point median filter and a 3 point Gaussian filter (of weights 0.25, 0.5, 0.25) is applied for the displacement computed at neighboring points at each step.

It must be stressed that the method is inherently two-dimensional, thus tracking is along all directions. However, it also employs sub-steps of 1D tracking because these facilitate the capture of large displacements (as found in the radial or base-to-apex, directions) and ensures a higher accuracy for sub-pixel estimates.

**Tracking Along the 2D Space-Time Image**

This section describes a procedure for following a border along one direction in a two-dimensional image (M-mode-like) starting from a known position at one instant.

Define $x$ as the horizontal direction and $y$ the vertical one. Columns are thus annotated $x_i$, $i=1…M$, where $M$ is the number of columns in the image. The tracking is given by determination of a discrete sequence of real numbers $y_i = y(x_i)$, starting from a known point $y_k$ corresponding to the columns $x_k$. 


The displacement from the known point $y_k$ to the point $y_{k+1}$ is estimated by evaluating the cross-correlation between the entire column at $x_k$ with the entire column at $x_{k+1}$. The cross-correlation function will present a maximum, the position of the maximum gives the value of the vertical displacement required to maximize the similarity between the two columns, therefore $y_{k+1}$ is estimated by adding such a displacement to $y_k$. This procedure is repeated between all pairs of nearby columns and the result is an estimate of the entire border $y_i$, $i=1...M$. The cross-correlation is here computed using a Fast Fourier Transform algorithm to reduce calculation time.

The first estimate $y_i$ is executed by cross-correlating columns of a relatively large span, 48 pixels, to ensure capturing large border displacements. This is then further refined iteratively. To accomplish this, a subset of the image is extracted by taking fewer points above and below the previous estimate $y_i$ and a new image whose center corresponds to the sequence $y_i$ is generated and used for the correction tracking. At every passage the vertical image size to 2/3 of its former size, and the refinement is repeated until no correction is found.

An improved and more natural result is eventually achieved by a final snake procedure, a technique to adapt a previously existing border to the content of a background image (7). This is achieved by imposing an attraction to feature elements in the background image, that in this case is the image brightness level corresponding to the fixed point $y_k$, and rules of elastic regularity to the border that makes increasingly difficult to develop high
curvatures. The entire process makes use of the heartbeat periodicity to ensure a periodic result and avoid the drift effect.

**Technical Limitation of Feature Tracking**

The border tracking technique, like any speckle tracking method, is based on quantification of changes on pixel brightness from one frame to the other. This gives a lower limit to velocity related to the need to see a speckle that is one pixel at one frame, moving to the neighboring pixel in the next frame. This limit is therefore

\[ \epsilon_v = k \frac{\Delta x}{\Delta t} \]

where \( \Delta x \) is the pixel size and \( \Delta t \) is the time interval between the two frames. The coefficient \( k \) depends on the quality of the tracking algorithm and on its ability to evaluate dynamic sub-pixel variations. This limit means that velocities that are well above this limit are estimated with great accuracy, such accuracy is reduced when velocity values approach and fall below such a limit.

This limitation also implies that an increase in the acquisition frame-rate (reduction of \( \Delta t \)) on one side allows an easier evaluation of large velocities and their rapid variations (like during the isovolumic phases). On the other side, it implies a reduced accuracy in the evaluation of lower velocities until it is not accompanied by a similar increase of spatial resolution (reduction of \( \Delta x \)).
**Phantom Image Preparation**

A series of artificial computer-generated loops has been prepared to allow testing of the image analysis procedure in simple and controlled conditions. For this, a phantom in a short axis projection of an ideal left ventricle was prepared as follows.

The endocardial and epicardial borders are represented by two concentric circles with radius \( R_0(t) \) and \( R_1(t) \), respectively. The image is prepared by making the annulus, which represents the tissue between the two borders, as uniformly colored gray on a black background. Then an 8x8 top-hat linear filter is applied to avoid unphysical discontinuities.

The epicardium movement is taken, in \([\text{mm}]\), as \( R_0(t)=10+5\cos(2\pi t/T) \) where \( T \) is the heartbeat period taken as \( T=1s \). The theoretical endocardial kinematics is constant along the border and depends on time only, velocity is only radial and given by \( V_0(t)=\frac{dR_0}{dt}=-\frac{\pi}{T}\sin(2\pi t/T) \), in \([\text{cm/s}]\). Percentage strain, computed relative to the length the border has at time zero, is \( St_0(t)=100\times\frac{(R_0(t)-R_0(0))}{R_0(0)}=100(\cos(2\pi t/T)-1)/3 \), and strain rate follows from (1) as \( SR_0(t)=10V_0/R_0 \) in \([\text{s}^{-1}]\). The epicardium is assumed either as moving accordingly to a constant thickness, \( R_1(t)=R_0(t)+5\text{mm} \), or as still \( R_1(t)=R_0(0)+5\text{mm} \).

Each image is square of size of \( 48\text{mm} \), centered on the tissue annulus, and has a resolution \( NxN \). Example images are shown in (Fig. 4), plates \( a \) and \( b \); the strain and strain rate time profiles are shown in (Fig. 4), plates \( c \) and \( d \). The loops are prepared by varying the resolution \( N \), the frame-rate \( FR \), and the epicardial type of motion.
The endocardial tracking method is applied to such images by taking on the first frame a number $N_p$ of points uniformly spaced along the circular endocardium.

**Clinical method**

7 patients underwent cardiac MRI for various indications (suspected Arrhytmogenic Right Ventricular Dysplasia, viability assessment in ischemic cardiomyopathy, hypertrophic cardiomyopathy, idiopathic cardiomyopathy) using 1.5 T clinical MRI scanner (Magnetom Avanto, Siemens, Malvern, PA) using a dedicated 8 channel phase array receiver coil. After localization of the heart, 6-13 contiguous short-axis slices (30 images per slice timed throughout the cardiac cycle) were performed to study the entire left ventricle from base to apex. Apical four and two chambers views were obtained. Cine images were acquired using a steady-state free precession pulse sequence. For left ventricular strain analysis, we selected mid ventricular short axis (for radial and circumferential strain) and apical four chamber view (for longitudinal strain). Left ventricular ejection fraction was calculated on MEDIS workstation.

**Results**

**Phantom Study**

The application of the image analysis method to the computer-generated phantom images is here analyzed. A global measure of the eventual error is computed by the root mean
square percentage difference. The root mean square, average and maximum errors in the endocardium strain are defined as

\[
\varepsilon_{St} = 100 \times \left[ \frac{\sum (St(t) - St0(t))^2}{\sum St0^2(t)} \right]^{1/2},
\]

\[
\varepsilon_{St}^{max} = 100 \times \left[ \frac{\max (St(t) - St0(t))^2}{\sum St0^2(t)} \right]^{1/2},
\]

where \( St0(t) \) is the exact value, \( St(t) \) is the value computed by the image analysis, and the summations extend over all frames \( NF = FR \times T \). The same definition is used for the radius, velocity, and strain rate. The tracking is about independent from the position along the endocardium, the differences between the different points is well below 1%.

Results for the root mean square errors are summarized in Table I for 15 phantoms with varying spatial resolution, frame-rate and the effect of varying the number of points used to track the endocardial border is also shown. The influence of the type of epicardial motion is considered for the two limiting cases when the epicardial border does not move (no motion) or moves with the endocardium (no thickening).

Errors are in all cases extremely small for the integral quantities (radius and strain) and slightly larger for the differential quantities (velocity and strain rate) that are related to the derivative of the former. This was expected because the derivative operator amplifies errors. The quality of results degrades when the resolution is reduced; in fact, the accuracy is related to the pixel size that represents (in a loose sense) the minimum displacement readable from one frame to the other. The time resolution does not affect the results significantly until the frame-rate is sufficient, at very high frame-rate results do not improve because frame by frame displacements becomes lower than the pixel size. This shows that an increase in the frame-rate is of little or no utility when it is not accompanied by an increase in the spatial resolution.
However, the simple sinusoidal motion here considered does not require an extreme time resolution. Endocardial results are not appreciably influenced by the type of motion that the epicardium undergoes. We also preliminarily verified that the results are not significantly affected by the adopted image filtering.

A visual presentation of results is given in (Fig. 4) where the computed endocardial border at two instants is reported over the phantom images (plates a and b). The strain and strain rate are reported in (plates c and d) for the case #1 and the small resolution case #8. The strain and strain rate in case #1 (squares) presents an excellent agreement with the theoretical value, the mean error being equal to 0.6% and 3%, respectively. The agreement is only a little worse in case #8, where image resolution is halved, with errors 0.9% and 4.5% for strain and strain rate, respectively.

**Clinical study**

Patient characteristics and strain data are presented in table 2. FT was performed successfully in all patients. Normal patients (no cardiac disorder identified) exhibited normal strain values and nearly (fig 5, panel A and B) synchronous contraction. As expected patient with idiopathic cardiomyopathy and left bundle branch block exhibited reduced strain values and marked dyssynchrony. Patients with ischemic cardiomyopathy exhibited reduced global strain values (lower with lower EF), segmental strain was markedly reduced (or flat) is segments with wall motion abnormalities (asynchrony). Patients with hypertrophic cardiomyopathy (with preserved or depressed systolic function) demonstrated globally reduced strain patterns.
Discussion

We describe a new method “feature tracking” to measure deformation and displacement parameters in myocardial tissue. This technology is here validated with an artificially prepared phantom images with the goal to check the correctness and accuracy of the method in an ideal situation. This option was preferred to the other option of artificially moving the myocardium of a MRI image to have a fully controlled situation, independent of the selected MRI image and of the method used to match the moved portion of image with the still background, and may lead to inconclusive results as a pure method testing. For the same reason the phantom images were perfectly clean from noise and addressed to test tracking in the radial direction only. In fact noise as well as inhomogeneities along the circumference must be described by a spatial distribution (characterized by a spatial spectrum, coherence, wavelength, assumed isotropic) and a time correlation (related to the spatial one) with a large arbitrariness. Eventually results can be insensible or very sensible to noise depending on such parameters. Afterward, when the phantom study demonstrated that method is technically correct and reliable, its applicability in the clinical setting is verified. Multi-directional strain analysis (radial, circumferential and longitudinal strain) was performed in 7 patients with wide range of EF and segmental wall motion abnormalities. The clinical test is aimed to verify the feasibility and meaningfulness of the FT method in examples of real MRI images, and results are not intended as a thorough validation of the diagnostic accuracy of this approach. However, in this clinical examples FT derived global strain values corresponded well with EF, segmental strain curves followed wall motion abnormalities.
This method was recently validated against HARP in assessment of peak global circumferential strain in a large population of patients with Duchenne Muscular Dystrophy (1). We found an excellent correlation between those two techniques, the analysis however was limited to global circumferential strain in mid ventricular short axis slice.

Feature tracking technology is relatively simple in usage and does not need any additional imaging sequences such as tagging, SENSE or DENSE. In addition to clinically validated assessment of peak global Ecc, regional strain data can be derived as well FT appears to be a robust technique able to estimate longitudinal and radial LV strain as well as longitudinal right ventricular strain.

More studies need to be done in order to exactly determine the limits of the technique regarding temporal and spatial resolution of the image. FT allows comparisons across other modalities such as echocardiography since the basic principle is the same.

Study limitation
This study did not analyze the effective reliability of the method in the clinical settings. Testing was limited to accuracy in a theoretical phantom, and to feasibility with CMR images. However in lieu of it we presented several patients with wide range of cardiac pathology.

Conclusion
Feature tracking has the potential to be integrated in clinical practice since it eliminates the need of time consuming analysis and additional acquisition procedure. FT can be used for assessment of left ventricular global and segmental strains and intraventricular dyssynchrony. As strain is a more sensitive method than EF in assessment of myocardial
contractility, we envision FT use in monitoring of cardiac toxicity in patients undergoing chemotherapy, differentiation of physiologic versus pathologic left ventricular hypertrophy, monitoring diseases progression and effect of various therapies etc. Further clinical studies are underway to assess usefulness of FT in clinical cardiology.

References


Table I. Phantom analysis of endocardial border tracking: root mean square percentual errors [%] are computed for the main quantities in correspondence of different phantom parameters. The parameters marked bold indicate the variations from the Phantom #1. The last phantom (*) is built without filtering the basic stepwise images with brightness changing abruptly in one pixel. Errors above 10% are marked bold.

<table>
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<tr>
<th>Phantom #</th>
<th>Frame Rate</th>
<th>Resolution</th>
<th>Epicardial motion</th>
<th>N</th>
<th>$\varepsilon_R$</th>
<th>$\varepsilon_V$</th>
<th>$\varepsilon_{SR}$</th>
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<td>with endo</td>
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Abbreviations:

$\varepsilon_R$ ... Root mean square percentual error Displacement [%]
$\varepsilon_V$ ... Root mean square percentual error Velocity [%]
$\varepsilon_{SR}$ ... Root mean square percentual error Strain [%]
$\varepsilon_{St}$ ... Root mean square percentual error Strain rate [%]
Resol ... Number of Pixel in X- and Y-direction
Framerate ... Number of frames per heartbeat
Np ... Number of points used to describe the border

**Table 2.** Resulted for global circumferential, radial and longitudinal strain analysis (%) for control patients as well as patients with know cardiac pathology.

<table>
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Abbreviations:

HCM: Hypertrophic Cardiomyopathy
DCM: Dilated Cardiomyopathy
LBBB: Left Bundle Branch Block
CMP: Cardiomyopathy
FIGURE LEGEND

**Figure 1.** CMR images of the left ventricle, in long axis view (left picture) and in short axis view (right picture), with a traced endocardial border drawn on top.

**Figure 2.** Space-time representation, where space is along a transmural cut, of the image sequence. The transmural cut is taken as for the starting point in figure 3. The time evolution of the starting point, tracked automatically, is reported.

**Figure 3.** Image of the left ventricle, in long axis view, with transmural cuts and cuts parallel to curved border.

**Figure 4.** Phantom study. Two images (cases #2) at maximum expansion (plate a) and contraction (plate b), the computed endocardial border points are overlapped. The strain (plate c) and the strain rate (plate d) computed with two different phantoms (cases #1 and #8) are shown in comparison to the effective values.

**Figure 5.** FT in patients undergoing clinical MRI studies: A and B: normal patients, C: idiopathic cardiomyopathy with LBBB, D and E ischemic cardiomyopathy, F and G: HCM,