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Regional assessment of LV wall in infarcted heart using tagged MRI and cardiac modelling

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Abstract

A segmental two-parameter empirical deformable model is proposed for evaluating regional motion abnormality of the left ventricle. Short-axis tagged MRI scans were acquired from 10 healthy subjects and 10 postinfarct patients. Two motion parameters, contraction and rotation, were quantified for each cardiac segment by fitting the proposed model using a non-rigid registration algorithm. The accuracy in motion estimation was compared to a global model approach. Motion parameters extracted from patients were correlated to infarct transmurality assessed with delayed-contrast-enhanced MRI. The proposed segmental model allows markedly improved accuracy in regional motion analysis as compared to the global model for both subject groups (1.22–1.40 mm versus 2.31–2.55 mm error). By end-systole, all healthy segments experienced radial displacement by ~25–35% of the epicardial radius, whereas the 3 short-axis planes rotated differently (basal:
3.3°; mid: −1° and apical: −4.6°) to create a twisting motion. While systolic contraction showed clear correspondence to infarct transmurality, rotation was nonspecific to either infarct location or transmurality but could indicate the presence of functional abnormality. Regional contraction and rotation derived using this model could potentially aid in the assessment of severity of regional dysfunction of infarcted myocardium.

Keywords: tagged MRI, cardiac, regional motion, non-rigid image registration, myocardial infarction, deformable model

(Some figures may appear in colour only in the online journal)

1. Introduction

Over the cardiac cycle, the left ventricle (LV) in a short-axis view undergoes a complex motion, involving radial and rotational components (Buckberg et al. 2008). Non-uniform motion between different cardiac segments exists both in the healthy heart and the diseased heart (Bogaert and Rademakers 2001), leading to a need for regional analysis of cardiac motion. Detailed analysis of regional cardiac wall motion could provide valuable information pertaining to functional abnormality when global functions such as stroke volume and ejection fraction fail to elucidate these disease conditions (Shehata et al. 2009). Various imaging techniques have been used clinically to quantify regional cardiac wall motion, including nuclear techniques (Gilland et al. 2008), tissue Doppler (Ho and Solomon 2006) and speckle tracking echocardiography (Kachenoura et al. 2010) as well as magnetic resonance imaging (MRI) (Li and Denney 2006). However, clinical assessment of regional cardiac wall motion remains subjective. Wall motion within each cardiac segment, as defined in accordance with the American Heart Association (AHA) 17-segment model (Cerqueira et al. 2002), is categorized qualitatively through visual inspection by a clinical specialist (Castillo et al. 2003) as having normal, hypokinetic, akinetic or dyskinetic motion. Such manual assessment is greatly dependent upon expertise, and thus subject to high inter-observer variability and low reproducibility (Ledesma-Carbayo et al. 2006).

Among different imaging modalities, tagged MRI (Zerhouni et al. 1988) has become the gold standard in assessing regional deformation of the LV wall as it does not involve ionizing radiation and reduces the operator dependency associated with echocardiography (Shehata et al. 2009). Although various methods have been developed to quantify regional myocardial displacement using tagged MRI, none of these methods have been used in routine clinical assessment due to their complexity. The harmonic phase imaging (HARP) technique has been the mostly widely used and validated method for strain quantification (Shehata et al. 2009, Bilgen 2010). Yet, due to phase wrapping artifacts, HARP is limited to tracking small deformations (Osman et al. 2000, Chiang et al. 2013) and requires high quality images (Loganathan et al. 2007, Smal et al. 2012). In contrast, feature-based image analysis methods, such as the deformable models, provide deformation analysis using extracted tag lines and the myocardial contours. These methods depend heavily upon the accuracy of control points detection based on tag extraction methods (Wang and Amini 2012), and thus may fail with images of a low quality. Comparing four conceptually different approaches, including optical flow, HARP, active contour-based tag line registration, and tagged image registration methods, (Smal et al. 2012) concluded that the image registration approach outperformed other methods in terms of tracking accuracy, regardless of image quality. However, this approach is computationally intensive due to the requirement to optimize a large number of parameters, and the algorithm’s accuracy may be compromised by local minima in the registration optimization function (Wang and Amini 2012).
Recently, new mathematical models were developed to assess motion abnormalities in the LV from grid tagging (Shi et al. 2012). With a 2-parameter model, both time-dependent contraction and rotation at different short-axis planes, known to be the two major components occurring in the LV wall during both contraction and relaxation phases, were estimated (Alrefae et al. 2008, Shi et al. 2012). The contraction term incorporates both circumferential shortening and radial wall thickening. The experimental results demonstrated the robustness of the model in estimating global motion in normal subjects. Since this method has only been applied to a limited number (five) of healthy subjects, it is difficult to evaluate accurately the ability of such an algorithm in assessing the regional motion abnormalities that occur due to localized necrosis of myocardium after a myocardial infarction (MI) event.

In the present study, we have extended upon the global method proposed by (Shi et al. 2012) so as to improve the quantification of regional motion at different cardiac phases by performing non-rigid registration to fit the empirical 2-parameter model at each cardiac segment defined by the AHA model. We have shown, for the first time, this simple model can be used to efficiently compute two motion parameters, consisting of contraction and rotation, for each LV segment in both healthy subjects and postinfarct patients.

2. Methods and materials

2.1. MRI data acquisition

We recruited 10 healthy volunteers and 10 patients with recent post-ST elevation myocardial infarction (post-STEMI) confirmed by delayed contrast enhanced magnetic resonance imaging (DE-MRI) and electrocardiography (ECG) signal (i.e. less than 6 months after admission). Inclusion criteria for control subjects were no prior history, symptoms or medication for cardiovascular disease, in addition to normal cardiac function indicated in echocardiographic examination. Exclusion criteria for patients were unstable angina, atrial fibrillation, tachycardia and moderate to severe valvular regurgitation or stenosis. Demographics of all subjects are provided in table 1. The study was conducted in the University of Malaya Medical Centre, and the imaging protocol was approved by the Institutional Ethics Committee (Ref: 989.75). All subjects provided written, informed consent.

All MRI scans were acquired using a 1.5T MRI system (Signa HDxt 1.5T, GE Healthcare, WI, U.S.A) with a dedicated cardiac phase-array receiver coil and ECG gating. For each healthy subject and patient, multiple short-axis tagged MRI images (13–15 slices) were acquired covering the LV from base to apex during contiguous end-expiration breath-holds. Imaging parameters were: TE/TR = 3.2/6.8ms, flip angle = 12°, field of view (FOV) = 350 × 350mm², output image size = 256 × 256pixels, in-plane resolution = 1.37 × 1.37mm, slice thickness = 8mm, gap between slices = 0mm, breath-hold time = 15s. At each short-axis plane, tagged MRI images were acquired at 20 cardiac phases/cycle with a grid spacing of 5mm. MR tagging was produced using a spatial modulation of magnetization (SPAMM) sequence using fast gradient-echo-sequence (FGRE) for signal generation. For patients with infarct, collocated short-axis DE-MRI images were obtained ~10–20min after intravenous injection of a gadolinium-based contrast agent to enable assessment of the location and extent of infarction.

2.2. Two-parameter deformable model of left ventricle

Contraction and twisting are the two main components of LV deformation that are evident in tagged MRI images during the systolic phase (Buckberg et al. 2008). In this study, we
performed non-rigid registration to fit an empirical two-parameter model proposed by (Shi et al 2012) to the tagged MRI data. The motion field of the myocardium for each cardiac segment, as defined by the contraction and rotation parameters, was derived from this registration process. The method is summarized below.

At the first cardiac phase, the LV appears at its dilated state and the grid tag experiences no deformation. The image from this cardiac phase was treated as the reference image for the subsequent image registration, with time denoted as $t = 0$. At $t > 0$, myocardial tissue undergoes deformation, as indicated by changes in the grid tag, in the form of radial displacement and rotation. Consistent with the Lagrangian description of the motion (Bower 2011), vectors $r = (x, y, t = 0)$ and $r' = (x', y', t)$ were used to denote Cartesian coordinates of the myocardial tissue before and after the deformation. The difference between these two vectors, i.e. $r' - r$, denotes tissue displacement over a period of time $t$. Magnitudes of $r$, i.e. $r = \sqrt{x^2 + y^2}$, and $r'$, i.e. $|r'| = \sqrt{x'^2 + y'^2}$, denote distances to the origin before and after the tissue deformation, respectively.

In-plane short axis radial displacement and rotation from coordinate $r = (x, y, t = 0)$ to $r' = (x', y', t)$ were related by a radial transformation matrix, $T$, and a rotation matrix, $R$, as follows:

$$\begin{bmatrix} x' \\ y' \end{bmatrix} = RT \begin{bmatrix} x \\ y \end{bmatrix}$$

where $R = \begin{bmatrix} \cos \theta & \sin \theta \\ -\sin \theta & \cos \theta \end{bmatrix}$ and $T = \begin{bmatrix} 1 - d & 0 \\ 0 & 1 - d \end{bmatrix}$. The origin is defined as being located at the centre of the LV cavity. The relative radial displacement, denoted by $d$, was formulated as $d = (|r| - |r'|)|r| = \alpha(t)r^2$, whereby $\alpha(t)$ is a time-dependent parameter. Larger $\alpha$ indicates larger radial tissue displacement. To indicate contraction, $d$ was subtracted from unity. As radial displacement is more significant in the endocardium than in the epicardium, the relative radial tissue displacement $d$ was formulated to be inversely proportional to $|r|^2$, allowing radial dependency. Consequently, our model formulation not only takes into account circumferential shortening, but also radial wall thickening.

In addition to contraction, the myocardium goes through a higher degree of rotation near the endocardium than in the epicardium (Akagawa et al 2007). The rotation angle $\theta$ in matrix $R$ was therefore formulated to be radial- and time-dependent, denoted as $\theta = \beta(t)r$. In this notation, positive (negative) $\beta$ leads to clockwise (counterclockwise) rotation when the tagged MR image is viewed in the direction from apex to base. An increase in the magnitude of $\beta$ implies larger rotation angle, whereas $\beta = 0$ indicates no rotation as in the case of motion at the midventricular level. The effect of applying these geometric transformations globally on an annular mesh of a simulated grid and a segmented short-axis tagged MR image of the LV wall is illustrated in figure 1.
To be consistent with (Shi et al 2012), and to allow for a fair comparison across subjects irrespective of differences in their LV sizes, both $\alpha$ and $\beta$ values estimated from the registration process (to be described in the next section) were normalized by the individual’s maximum epicardial radius ($r_{epi}$) at the corresponding short-axis plane and time frame. The normalized values, $\alpha/r_{epi}$ and $\beta/r_{epi}$, are reported.

2.3. Non-rigid registration for motion quantification

To perform motion analysis, the LV myocardium was initially segmented from the image at the first cardiac time frame by manually delineating the endocardial and epicardial borders. Regions outside and inside of the myocardium were then masked, leaving only the myocardium with superimposed grid tags to create an LV myocardial mask as shown in figure 1(b). Tracking the changes of $\alpha$ and $\beta$ values over time was performed automatically by non-rigidly transforming the mask using equation (1) to match with the deformed myocardium in tagged MR images from subsequent cardiac time frames.

Instead of using the annular mesh of a simulated grid (figure 1(a)) as a reference image as described by (Shi et al 2012), we extracted the actual LV myocardial mask (figure 1(b)) and maintained the pixel grey levels within it for registration. This is because the short-axis plane of the LV wall is known to be non-perfectly circular, especially in postinfarct patients (Konermann et al 1997, Bogaert and Rademakers 2001). Moreover, a better match to the deformed tagged MR images from subsequent time frames was found to be achieved (result not shown) by using the original pixel grey levels of the LV myocardial mask as compared to the binary grey levels of a simulated grid. In this regard, the signal processing and motion estimation approaches presented are applicable to more realistic cardiac abnormalities and thus denote significant improvements over the previous work in (Alrefae et al 2008, Shi et al 2012), which dealt with modelling normal heart. The similarity between the transformed mask and the deformed myocardium was evaluated using normalized cross-correlation, which is a registration similarity measure which accounts for linear intensity fluctuation across the image sequence (Makela et al 2002). We computed the similarity values only within the mask region, excluding pixels outside of it. During the registration process, the optimal motion values that yielded the best match (i.e. the maximum correlation value) were estimated using an exhaustive search to within a range of 0–25 mm for $\alpha$ and -12–12° for $\beta/r_{epi}$. We have empirically chosen these ranges based on the observation of the minimum and maximum values of both types of motion in our data sets, as well as from those reported in the literature (Akagawa et al 2007, Helle-Valle et al 2009, Shi et al 2012).

As the wall motion is asymmetric and non-uniform across different cardiac segments, especially in cases of MI, the optimal $\alpha$ and $\beta$ values were computed separately for each segment.
Specifically, the registration was performed by first dividing the mask into different segments of 60° according to the AHA model (Cerqueira et al. 2002). This involves identifying the midline of the septum between the two attachment points of the RV-LV wall with reference to the LV centroid, followed by registration of each of these segments to the tagged MR images obtained from subsequent time frames. Overall, three different short-axis LV locations were chosen for analysis in each subject, consisting of the basal, midventricular and apical slices. A total of 16 cardiac segments were assessed, whereby segment 17 at the apex was excluded as it was only assessable from the long-axis scan. To reduce computation time and to correct for the shift in the LV centroids at different time frames, all images were cropped, prior to registration, to encompass predominantly the LV region, and with the image origin positioned at the centroid coordinate of the LV myocardium. As tags faded gradually throughout the cardiac phases due to relaxation of the tissue magnetization process (Prince and McVeigh 1992), we restricted our analysis to the systolic window, whereby the contrast-to-noise ratio of the tags was still considered sufficient for reliable registration.

The motion analysis framework was implemented using MATLAB (vR2012a, Mathworks, Natick, MA) on an Intel Xeon CPU E5-2620 @2.00 GHz computer. We have performed analysis on all 10 healthy subjects and 10 postinfarct patients.

2.4. Accuracy of motion quantification

To evaluate the improvement from the proposed segmental model fitting, we performed registration on the entire LV myocardial cross-section using both the global model approach proposed by (Shi et al. 2012) and our proposed modified algorithm.

To assess the registration accuracy, an expert clinician manually validated the position of grid intersections within the myocardium region by landmarking them at both mid-systolic and end-systolic phases. The root mean square (RMS) distances, i.e. the positional errors, between the intersections of the transformed grid using the two different types of registration models and the actual grid intersections landmarked by the expert at both cardiac phases were computed. This accuracy analysis was performed on all 20 data sets from both patients and healthy volunteers (involving approximately 1300 point-to-point comparisons) and the mean RMS distances in mm were reported as the registration error.

We additionally computed the localized cross-correlation map of the transformed and the systolic tagged MR images, with a sliding window of $8 \times 8$ pixels, to allow for both quantitative and visual assessment of the registration accuracy. The values obtained range from $-1$ to $1$. A value of zero indicated there is no correlation and negative values indicate an inverse correlation. At locations where the grid patterns from the two images match well, a high positive cross-correlation value was obtained.

2.5. Effect of infarct transmurality on motion parameters

The presence of contrast enhancement in each of the 16 LV segments was identified from the DE-MRI images of the postinfarct patients and scored by an expert clinician. Transmurality of necrosis in each segment was classified into 3 categories: (i) 0% (non-infarcted), (ii) 1–50%, and (iii) >50%. The effect of the infarct transmurality on $\alpha_{r\text{epi}}$ and $\beta_{r\text{epi}}$ values at different wall segments was tabulated and reported. As the sample size is small and the data is not normally distributed, significant statistical differences among different categories of transmurality were analyzed using the non-parametric Mann–Whitney U test provided in the SPSS statistical software (SPSS, v22.0, SPSS Inc., Chicago, Illinois). The significance level for $p$ was set to 0.05.
3. Results

3.1. Accuracy of motion assessment

Figure 2 shows the estimation accuracy of the cardiac wall motion in terms of RMS positional error between the model-estimated and manually landmarked tag intersections in both healthy subjects and patients. Positional error (both median and interquartile range) from our proposed segmental model fitting was much lower than that produced by the global model fitting. The median (interquartile range) error of the former was 1.22–1.40 mm (1.14–1.90 mm) versus 2.31–2.55 mm (2.97–3.17 mm) for the latter. The positional error resulting from our proposed model was found to be comparable to that reported by (Atalar and McVeigh 1994) using the conventional least-squares-error tag centre detection method for tag thicknesses of 1 pixel. Comparing both subject groups, the overall positional errors were slightly larger in patients regardless of the model fitting approaches, probably due to the higher variability and complexity in the regional motion of diseased heart tissue.

Figures 3 and 4 illustrate the improvement in the model fitting results using the segmental model as compared to the global model at the basal, midventricular and apical slices in a healthy subject and a postinfarct patient, respectively. Columns 2 and 4 in these figures depict the superimposition of the transformed grid (blue) on the end-systolic grid (red) after registration using the global and segmental models, respectively. Columns 3 and 5 in these figures are the normalized cross-correlation maps which show the goodness of match between the transformed and the end-systolic tagged MRI images. The more red the myocardium appears in these maps (i.e. the higher the coefficient value), the better is the match between both images. Apparent improvement was shown in the goodness of match in both healthy subjects and patients with the use of the segmental model approach as compared against the global model in all three short-axis planes. It can be seen that the global model failed to estimate the correct motion at certain regions, as depicted by the mismatch.
between the blue and red lines as well as the low coefficient values (cyan regions) in the correlation map.

3.2. Contraction and rotation in healthy subjects

Time series analysis of the tagged MR images shows a highly complex motion of the human left ventricle. Figure 5 shows the mean contraction \((\alpha/r_{epi})\) and rotation \((\beta/r_{epi})\) values for the healthy subjects throughout systole, estimated using the segmental model approach. Consistent with the current knowledge of heart physiology, the estimated contraction increased in magnitude towards the end systole in all three short-axis levels (i.e. base, midventricle and apex). By end-systole, all cardiac segments have contracted by approximately 0.25–0.35 in scale of \(\alpha/r_{epi}\). In other words, all points on the epicardium were contracting by 6.25–12.25% from the epicardial wall, where the percentage value was derived using the relative displacement equation \((\alpha/r_{epi})^2 = 1 - |r_{epi}|\). Slight inhomogeneities in contraction across segments were also observed.

The rotation varied for all three planes as estimated using our proposed segmental model. At end-systole, the base was clearly rotating in a clockwise direction \((\beta/r_{epi} \text{ was positive})\), the midventricle remained nearly un-rotated \((\beta/r_{epi} \text{ was close to 0})\), whereas the apex was rotating in the counterclockwise direction \((\beta/r_{epi} \text{ was negative})\), when viewed from the apex to the base. Rotation at the midventricular level was small, while the opposite motions of the base and the apex created torsional motion.
3.3. Effect of infarct on cardiac motion

Figure 6 shows the motion field obtained by the segmental model at base, midventricle and apex of a healthy subject (figure 6(a)) and a postinfarct patient (figure 6(b)). As shown by the motion vectors in the healthy subject, while the heart contracts during systole, all segments in the base were rotating clockwise whereas the apex was rotating counterclockwise. This resulted in an overall torsional movement. The midventricle underwent predominantly contraction with negligible rotational movement at systole. Compared to the healthy subject, LV contraction in the postinfarct patient was markedly reduced, especially within the infarcted segments. Additionally, infarction was also found to disrupt the expected direction of rotation within the affected segment at each plane. In the example shown below, this disruption was evident at the infarcted segments at basal and the midventricular regions.

The bulls-eye diagram in figure 7 shows the relationship between the extent of infarction and the two motion parameters analyzed in this study. We illustrate the relationship in two patients with left anterior descending (LAD) related MI. The analysis consisted of a segment-to-segment comparison of the two parameters at end-systole with respect to the normal range for the healthy subjects (figure 5). The abnormal deviation of the contraction $\Delta(a/r_{epi})$ from the normal range figure 7(b) was observed to occur primarily within the infarct segments as well as at the adjacent segments in both patients. The affected segments (blue segments) displayed less contraction as indicated by the negative deviation.
values. In contrast, some of the remote segments (yellow and orange segments) displayed a higher degree of contraction when compared against normal heart, as indicated by the positive deviation values.

Rotation, in contrast to systolic contraction, was found to deviate from the parameter range in normal heart at places nonspecific to the infarct segments. In patient 1, for example, we observed large absolute deviations of rotation |∆(β/\text{r}_{\text{epi}})| in the range of 5–10° with reference to the mean and standard deviation values. These occurred primarily in healthy segments surrounding the infarcted segment at the base. Many of these healthy segments rotated in the opposite direction (as indicated by asterisks), i.e. counterclockwise, as opposed to the expected clockwise direction seen in healthy subjects. In patient 2, the non-specific deviation in the rotation angle was also observed in other planes. Though nonspecific, motion abnormality was clearly identified in all the short-axis planes with regional infarctions.

The correlation between the contraction and infarct transmurality for all patients is plotted in figure 8 in comparison to healthy subjects (control). Results of our analysis for healthy subjects showed a median contraction (α/\text{r}_{\text{epi}}) ranging from 0.28–0.30, with an increase in the amount of variation from base to apex. Although we observed a general trend of reduction in the contraction with increasing infarct transmurality, only segments with >50% infarction at the base and midventricular levels showed statistically significant reduction as compared to the non-infarcted segments (at the p-value = 0.05 level), as indicated by symbol ξ.

Figure 9 shows the relationship between myocardial rotation and infarct transmurality. For healthy subjects, our analysis showed that the base, midventricle and apex were rotating with median values of 3.3°, -1° and -4.6°, respectively. No clear correlation was observed between myocardial rotation and the degree of necrosis at the base and midventricle, though the interquartile range of the rotation angle at the base was reduced at higher transmurality level. However, at the apex, the non-infarcted and segments with 1–50% infarction generally rotated with larger angles in the counterclockwise direction as compared to controls. The segments with >50% infarction, by contrast, experienced a significant reduction in rotation as compared to control and non-infarcted segments. Directionally opposite rotation was also observed in several cases.
Circumferential strain (shortening) and torsion (a measure of the difference in rotation between base and apex) have been recognized as markers of impaired myocardial function (Helle-Valle et al. 2009, Shehata et al. 2009). Various analysis methods for tagged MRI, including HARP (Osman et al. 2000), feature-based image analysis approaches (Deng and Denney 2004), as well as optical-flow based methods (Florack and van Assen 2010) have been demonstrated in detecting abnormalities in myocardial strain due to ischemic heart diseases (Smal et al. 2012), dilated (Young 2001) and hypertrophic cardiomyopathy (Ennis et al. 2003). Despite extensive efforts, they have not been adopted in routine clinical assessment, primarily due to their dependency on image quality as well as the need for user interaction (Castillo et al. 2003, Smal et al. 2012, Wang and Amini 2012). In view of this, non-rigid registration approaches (Ledesma-Carbayo et al. 2008, Smal et al. 2012), proven to yield better tracking accuracy in the presence of noisy images (Smal et al. 2012), have been proposed. Nevertheless, these methods often involve optimization of a large number of parameters (Ibrahim 2011), making them computationally intensive and thus impractical clinically. Furthermore, to date, most algorithms (Rougon et al. 2005, Ledesma-Carbayo et al. 2008, Smal et al. 2012) have only focused on registration accuracy, without demonstrating any potential clinical application.

The segmental 2-parameter model approach proposed in the present study was able to capture abnormalities in both contraction and rotation of each cardiac segment in postinfarct

Figure 6. Example of motion field at end-systole, relative to end-diastole, for a healthy volunteer (a) and a postinfarct patient with occlusion in all three main arteries (b), at basal, midventricular and apical slices. The motion field was overlaid on the short-axis tagged MR image and myocardium contours at end-diastole. Yellow regions in (b) indicate infarct regions identified from the DE-MRI image. Length of motion vector indicates the magnitude of displacement. Scale bar represents a distance of 7.5 mm.

4. Discussion

Circumferential strain (shortening) and torsion (a measure of the difference in rotation between base and apex) have been recognized as markers of impaired myocardial function (Helle-Valle et al. 2009, Shehata et al. 2009). Various analysis methods for tagged MRI, including HARP (Osman et al. 2000), feature-based image analysis approaches (Deng and Denney 2004), as well as optical-flow based methods (Florack and van Assen 2010) have been demonstrated in detecting abnormalities in myocardial strain due to ischemic heart diseases (Smal et al. 2012), dilated (Young 2001) and hypertrophic cardiomyopathy (Ennis et al. 2003). Despite extensive efforts, they have not been adopted in routine clinical assessment, primarily due to their dependency on image quality as well as the need for user interaction (Castillo et al. 2003, Smal et al. 2012, Wang and Amini 2012). In view of this, non-rigid registration approaches (Ledesma-Carbayo et al. 2008, Smal et al. 2012), proven to yield better tracking accuracy in the presence of noisy images (Smal et al. 2012), have been proposed. Nevertheless, these methods often involve optimization of a large number of parameters (Ibrahim 2011), making them computationally intensive and thus impractical clinically. Furthermore, to date, most algorithms (Rougon et al. 2005, Ledesma-Carbayo et al. 2008, Smal et al. 2012) have only focused on registration accuracy, without demonstrating any potential clinical application.

The segmental 2-parameter model approach proposed in the present study was able to capture abnormalities in both contraction and rotation of each cardiac segment in postinfarct
Figure 7. Relationship between infarct location and contractions as well as rotation for two patients with LAD-related MI. (a) Transmurality of necrosis was classified into three categories: (i) 0% (non-infarcted), (ii) 1–50% and (iii) >50%. Difference in contraction and absolute difference in rotation angle with respect to the mean of the control group at the end-systolic phase are shown in (b) and (c), respectively. Segments with differences lower than the standard deviations were set to zero and displayed in green. Asterisks (*) in some segments of the bulls-eye diagram in (c) indicate opposite rotation angles with respect to the control group.

Figure 8. Comparison of contraction ($a/r_{epi}$) among base, midventricle and apex of healthy subjects (control) and patients at the end-systolic phase, estimated using the segmental two-parameter model. Transmurality of myocardial necrosis in patients was classified into three categories: (i) 0% (non-infarcted), (ii) 1–50% and (iii) >50%. Box–whisker plot indicates the median, interquartile range, minimum and maximum values. Number within each box plot indicates the median value. Symbol * denotes significant difference as compared to control, whereas $\xi$ denotes significant difference as compared to 0% (non-infarcted) with threshold of $p < 0.05$. 
patients. In contrast to the global approach, the proposed segmental method does not assume symmetrical motion of the LV wall and has shown an improved ability to more accurately estimate inhomogeneities in cardiac motion across the entire myocardium in short-axis planes (figures 2–4). This improvement was observed in both healthy subjects and patients, and the difference in segmental motion becomes more significant after regional infarction (Sellke et al 2010).

The trends of contraction and rotation over the systolic period (figure 5) estimated from our segmental model were in agreement with the findings of other studies using echocardiography (Akagawa et al 2007, Helle-Valle et al 2009) and MRI (Rougon et al 2005, Florack and van Assen 2010). Motion parameters recovered are consistent with the generally accepted cardiac physiology whereby healthy myocardium contracts and twists synchronously during systole causing the base to rotate clockwise and apex to rotate anticlockwise, ejecting blood from the chambers (Buckberg et al 2008). The advantage of the algorithm implemented in this study includes its simplicity, as it involves fitting only two motion-related parameters on the LV wall at the short-axis plane. This analytical model can be used to derive other aspects of LV wall mechanics including radial and circumferential strains (Shi et al 2012), which were typically estimated in the literature by fitting a larger number of parameters (Deng and Denney 2004), rendering the process more time consuming and complex.

Our study has also shown that the segmental model was able to identify the abnormal reduction in contraction in the infarcted segments of postinfarct patients, as compared to the healthy subjects (figure 8). This is consistent with published findings, which reveal a significant reduction in circumferential shortening in patients with LV systolic dysfunction (Sengupta et al 2008, Helle-Valle et al 2009, Shehata et al 2009). While the infarcted regions have reduced or zero contractility (hypokinesia or akinesia), some of the adjacently or oppositely located

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**Figure 9.** Comparison of rotation ($\beta_{rer}$) among base, midventricle and apex of both healthy subjects (control) and patients at the end-systolic phase as estimated using the segmental two-parameter model. Transmurality of myocardial necrosis in patients was classified into three categories: (i) 0% (non-infarcted), (ii) 1–50% and (iii) >50%. Box–whisker plot indicates the median, interquartile range, minimum and maximum values. Number within each box plot indicated the median value. Symbol * denotes significant difference as compared to control, whereas $\xi$ denotes significant difference as compared to 0% (non-infarcted) with threshold of $p < 0.05$. 

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healthy segments were found to display hypercontraction (figure 7). Hypercontraction or hyperkinesia is seen as a compensation mechanism to counteract the loss of wall function due to infarction, in which the undamaged segments work harder in order to maintain the systolic function close to normal (Oubel et al 2012). This ultimately leads to the enlargement of the heart. Overall, our results showed a general trend of reduction in contraction with an increase in infarct extent (figure 8), consistent with (Helle-Valle et al 2009). In parallel with findings by (Chan et al 2006), while >50% infarct segments showed a significant reduction in contraction as compared to the non-infarcted segments, no statistically significant difference was found between 1–50% infarcted (subendocardial) segments and non-infarcted segments.

Regional rotation, in contrast, was found to be dissociated from contraction and was non-specific in indicating the location of infarction (figures 7 and 9). Such dissociation has previously been reported by (Helle-Valle et al 2009) in their study of apical rotation using speckle tracking echocardiography. We found no clear correlation of rotation angle with infarct transmurality at base and midventricle. Infarction to the apex, however, was shown (figure 9) to cause the apex to rotate more in the intended counterclockwise direction in non-infarcted segments and segments with 1–50% infarction. Our findings are in agreement with previously published literature (Sengupta et al 2008, Nakatani 2011) which suggested that in the presence of subendocardial dysfunction, contraction of the subepicardium became more significant, thus resulting in an increase in myocardial rotation. Transmural infarct segments, on the other hand, produced a substantial reduction in apical systolic rotation due to a reduction in both subendocardial and subepicardial contraction (Kroeker et al 1995, Nagel et al 2000). Consistent with (Helle-Valle et al 2009), we have also observed duality in rotation direction specifically in the adjacent regions of the infarct segments in some patients. An example is given in figure 7 Patient 2, whereby at base level, the rotation of infarcted anteroseptal segment remains normal, but this segment was being pulled by two adjacent segments in opposite directions. Specifically, the anterior segment rotated in the clockwise direction, whereas the inferior-septal segment rotated in the counterclockwise direction. This abnormality in rotation was not restricted to the regions adjacent to the infarct zone, but could manifest over the entire LV cross-section. This could be explained by the disruption of the complex LV fiber arrangement in the presence of infarcted tissue (Lorenz et al 2000).

Limitations were noted with this preliminary study, including the small number of subjects that were assessed. By including a larger population in the study with both subject groups matched with regards to age and sex, a better estimate of the contraction and rotation statistics could have been achieved to assist in determining a good cut-off point for the diagnosis of segmental abnormality (as given by figure 7) in patients. Though generalized clinical conclusion cannot be drawn, we have shown encouraging results with the use of this method. Future study can also consider dividing the myocardium to smaller segments than that recommended by AHA to provide finer approximation of the motion alteration due to regional MI.

The current implementation of the algorithm requires user interaction to delineate the myocardium prior to model fitting. The level of interaction has the potential to be reduced given a reliable and automated segmentation algorithm integrated into the existing motion analysis framework. The implementation of the algorithm was coded without speed optimization. The time for motion estimation over the systolic period of a patient for all segments was approximately 28 min. Improvements in computation time may be achieved with the inclusion of an appropriate optimizer in the algorithm, such as gradient descent optimizer (Klein et al 2005, Oubel et al 2012) or through the use of parallel computing (Sharma and Martin 2009). Sudden changes in the radial motion and rotation due to the presence of regional infarction may cause...
discontinuity of the tag lines between the adjacent segments. Constraints may be incorporated into the algorithm to ensure such continuity between the segments in future studies.

We also note that motion analysis in this study was restricted within systole as tag fading using the SPAMM imaging sequence precludes the accurate estimation beyond end-systolic phase. The use of the CSPAMM (Fischer et al. 1993) sequence has been shown to reduce the tag fading, and the use of our proposed method for CSPAMM tagged images can potentially be extended to the entire cardiac cycle. Alternatively, data acquisition at 3T MR scanner would produce less fading of tag lines because of the increased T1 recovery time.

5. Conclusions

We have proposed and evaluated a segmental two-parameter model for regional motion analysis using tagged MRI acquired from both healthy subjects and postinfarct patients. Our study has shown that contraction reflects the transmural extent, with comparable contraction levels in the healthy and non-infarcted segments as compared to segments with hypokinesia or dyskinesia in myocardium with transmural infarction. Myocardial rotation, however, is not a specific measure to identify the location and extent of infarction but can be used to indicate functional abnormality. Both contraction and rotation may be combined to assist in the assessment of myocardial dysfunction in conjunction with DE-MRI. Potential applications include differentiation of non-viable scar tissue from dysfunctional but potentially salvageable viable myocardium to help in the selection of patients who would benefit most from revascularization (Turkbey and Dombroski 2009). Other potential application includes examining progression of heart remodelling after infarct or myocardial hypertrophy (Ennis et al. 2003) as well as evaluating the effect of fibrotic tissue deposition in the diabetic heart (Loganathan et al. 2007).

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