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An Optimization of Ventricular Fiber Orientation by Cardiac Anatomy

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***Abstract*—in this amazing world human facing a lot of problems in medical field especially in cardiac function. This concept is an enhanced technique for analyzing the myocardial fiber defects. Atlas is refer to by the use of high resolution ex-vivo magnetic resonance (MR) and diffusion tensor (DT) MR image of a normal human heart is captured. In this particularly analyzing the ventricular chamber of the captured image through in-vivo computed tomography (CT) image for estimating the fiber orientation and utility in simulation of cardiac electrophysiology. The map of the estimated fiber orientation is generated by the electrocardiogram (ECG) with this approach the state of the art technique is the method for estimate of cardiac fiber orientation.**

***Index Terms*— Digital image processing, myocardial, electrophysiology, magnetic resonance imaging (MRI).**

I. INTRODUCTION

Radiology plays a vital role in advancement of biological science and also it is a science that uses medical imaging. The various techniques such as MRI, CT that are used to analyze the images. In different biological systems, heart is the important organ and also it is the advanced example of virtual organ. The phenomena such as normal ventricular propagation, arrhythmia, defibrillation, electromechanical coupling and cardiac resynchronization are studied by using state-of-art whole-heart models of electrophysiology and electro mechanics [2]. In recent years, there is a massive improvement in imaging, includes ex vivo structural and diffusion tensor (DT) magnetic resonance imaging (MRI), this facilitates the whole cardiac structure through in vivo computed tomography image referred to us Atlas. Due to this consequence of converting ex vivo image to in vivo image, it helps to detect the function of individual fiber in the chambers of heart and also helps to understand the myocardium fiber orientations for patient specific model of cardiac function and assessment of therapy.

For computational studies of cardiac electrophysiology and mechanics the data of myocardial fiber orientation is necessary that have been predominantly obtained using histological methods, for examples the canine [9] and swine [10] ventricular models developed at Auckland University, and the rabbit ventricular model developed at the University of California, San Diego [11].This method requires more time this leads to tissue deformation errors occurs because of cutting, confinement to 2-D and not suit in vivo. Recently, for histological DTMRI has transpired as energetic alternative and in many species this data of DTMRI cardiac fiber have been captured [3], [4], [12]. Fiber orientation of acquired DTMRI have been shown to correlate to match with histological data [8] and have in process to be used in whole heart computational electrophysiology and electrophysiology [3], [5]-[7], [14],[15]. The faster data acquisition, 3-D data and nondestructive imaging are offered by DTMRI and DTMRI technology has several disadvantages, there is a presence of bulk motion of the heart interfaces with diffusion measurements those that reduce its application in the vivo setting. The conversion of DTMRI technology to in vivo setting are first made by Edelman et al. [16], but it was later known that myocardial strain was interacted by their diffusion coding so, to correct the strain effects a method as proposed [17], [18] but this correction was accomplished to increase in noise acquisition and complexity.

The strain correction method has been used to perform the 3-D tractography of ventricular fiber orientations in combination with interpolation [19], for interpolation the data prior was very inadequate, scan time was upto 40 min and breathholding difficult task for patients which leads to causes inter-slice misalignments. To examine fiber orientation in the ventricles there have been a lot of methods but compared to all methods, rule based methods have earned some esteem; the inclination angles alter sleekly in the transmural direction by the ventricular walls which are assigned by the fiber orientations [4], [5]. For showing that estimated fiber orientations relate mutually with DTMRI data better than the fiber orientations, they tested the approach using normal and failing canine hearts integrated using

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a rule based method. To calibrate estimation error in each heart and the denouement of this error on simulation of cardiac electrophysiology in the peculiar heart; but simulating sectional electrical activation maps and also in pseudo Electrocardiograms (ECGs).

II. EXISTING SYSTEM

In the presented methods the multi resolution wavelet analysis method is used to demarcate the characteristic features of ECG. In the QT segment of digitized electrocardiograph recordings, the feature extraction technique that based on a discrete wavelet transform (DWT) was proposed. At first, by using DWT technique the signal is denoised and noise components parallel coefficients are disposed. For the detection of R peak, here the multi resolution technique is used. Then, Q,S peak, QRS onset and offset points are recognized.

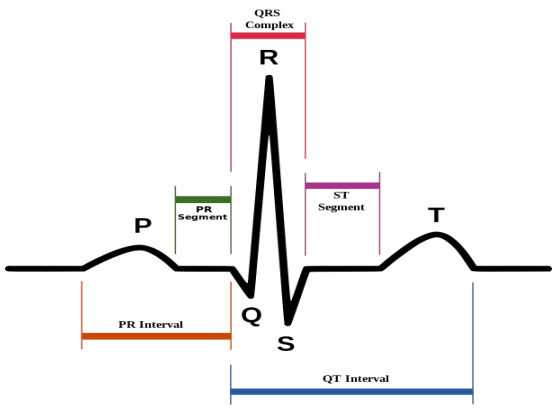

Figure 1: ECG beat with characteristics points

For make out many cardiac diseases, Electrocardiogram (ECG) is widely used one. This heart disease is one of the crucial causes of morality all over the world. The source of ECG is the heart muscles electrical activation that causing sequence of repolarization and depolarization in its membrane. Due this electrical activation there is an occurrence of some electrical pulses that are propagated along the cell fiber and passed to adjoining cells.

Figure 1 shows that the result is generated electrical pulses travels through the cardiac surface. By using the surface electrodes, the electrical impulse can be amplified that is detected and displayed as the ECG. For ECG recording, 12-lead electrode systems are used. ECG waveform contains five various component waves, they are P, Q, R, S and T these waves preceded by conditional U wave. For identification of patients with cognitive heart failure, a wavelet based soft decision methods is used. By using discrete wavelet transform, a method of analysis of myocardial infarction is explored.

Here, the wavelet transform (WT) provides a description of a signal, decaying it at various time frequency resolutions. WT is best tool for analysis of non-stationary signals such as ECG. ECGs various wave components having separate frequencies, that are visible under multi resolution analysis.

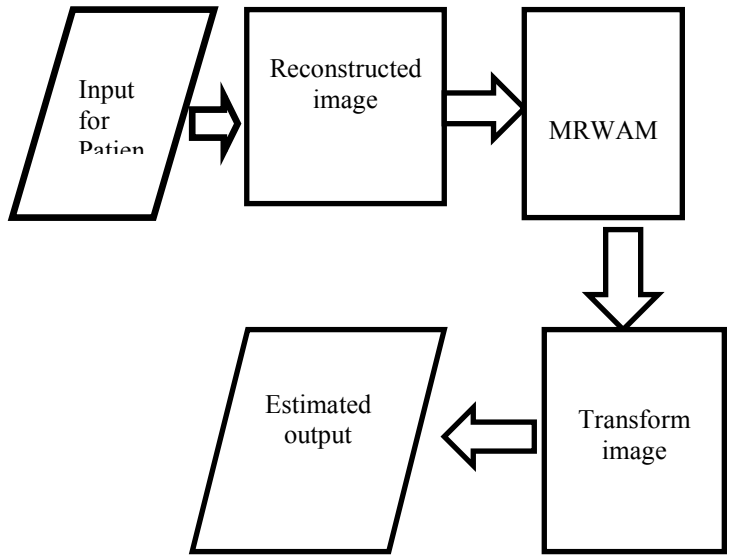

Figure 2: Block Diagram of Existing method

A. Multi resolution wavelet analysis method:

In figure 2 the Multi resolution wavelet analysis used for obtaining a feature withdrawal time localization of spectral components and it gives the representation of time frequency of the signal.

Between many time-frequency representations, the DWT is the most efficient because it has unique properties and ability to resolve a diverse. The function of various scale and translation allows one to estimate various frequency and signals time

Localization. For fine analysis, DWT becomes computationally rigorous. It is less efficient and natural and it take some energy to invest in wavelets to become able to select the best one for particular function and to implement it correctly. To overcome the drawbacks of DWT, we proposed two algorithms, affine and LDDMM transform

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III. PROPOSED SYSTEM

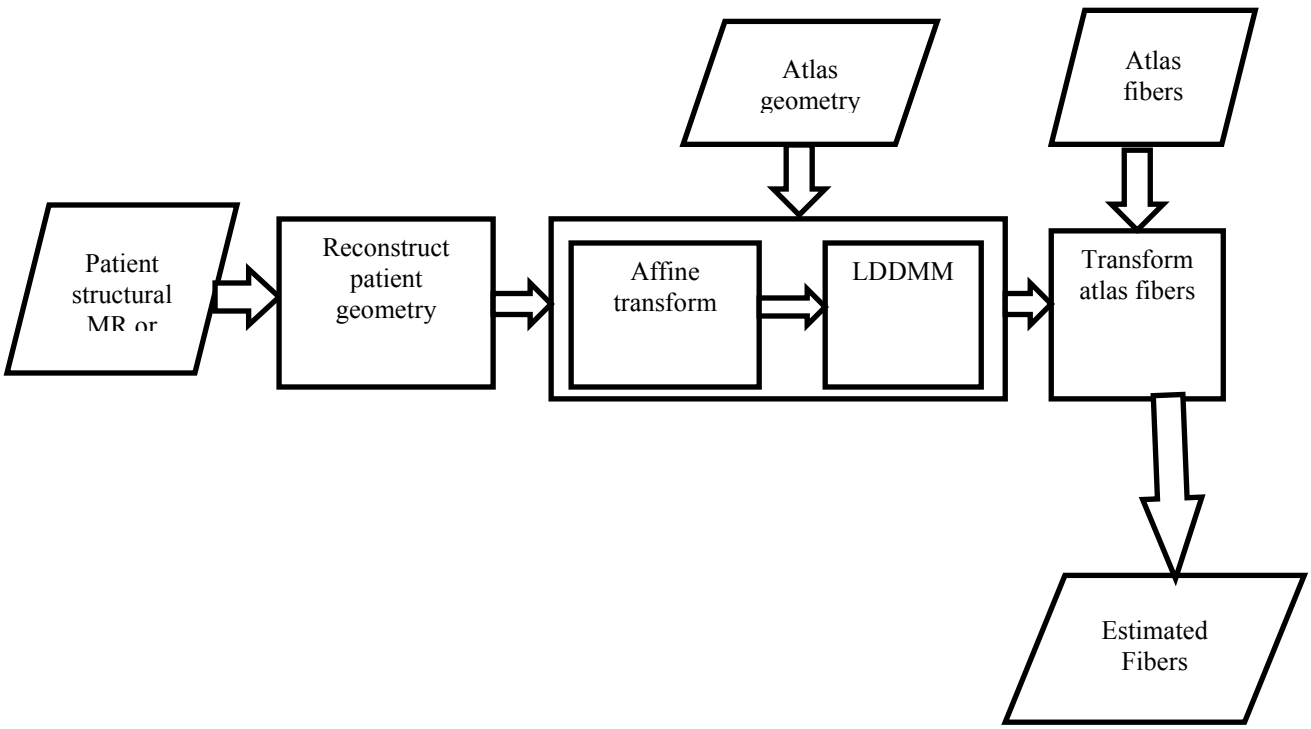

Figure 3: Block diagram of proposed system

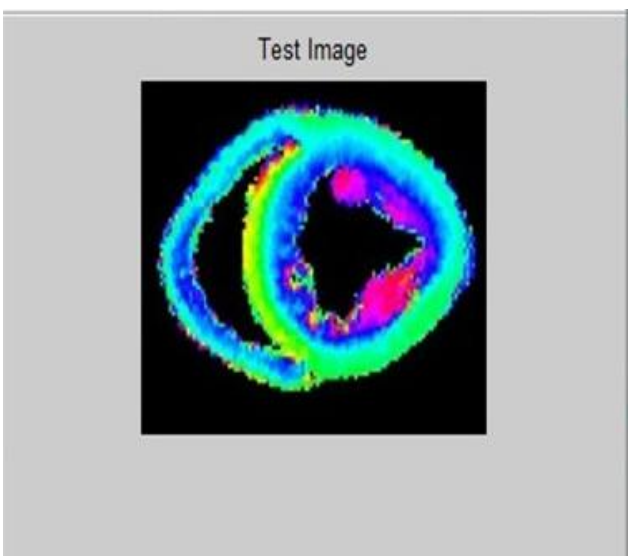

Figure4*: Test* image of the patient ventricular model

A. Estimation of fiber orientation

The vital impression of our fiber assessment practice is to achievement matches in fiber alignment, virtual to geometry, among dissimilar hearts in suggestion to near the fiber assembly of a heart for which only the geometry information is presented. Figure 3 shows our dealing pipeline of our practice.

In the resulting subsets, it define the few component of the pipeline by representing how the assessment is implements for an example patient who is scan by using in vivo computed tomography (CT)that is shown in figure 4. The number some blocks is referred by the following subsection below atlas ventricles; the fibers forming a left-handed helix on the epicardium are clearly visible.

C. Reconstruction for patient geometry:

Ventricular geometry of the patient heart in diastole is reassembled from an *in vivo* CT image with a subdivision method that is parallel to the one used for the atlas. The patient image is resampled prior to reestablishment such that the in plane resolution was 0.4297 ×0.4297mm as in the atlas. Likewise, the amount of parts for which significant are physically select, and the interval of out-of-plane

interpolation is adjusted so that the segmented patient heart image have a slice thickness of 1mm.

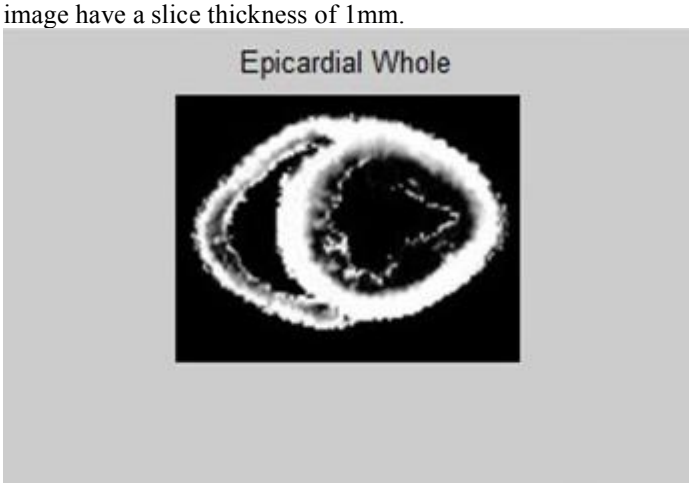

Figure 6: Epicardial whole of the patient heart

B. Atlas Geometry: The previous stage of restoration of the patient ventricles, the ventricular image of the atlas is distorted with the tool of computational anatomy it will be match for the patient geometry image. The main two steps of deformation are performing. The first step is affine transform.

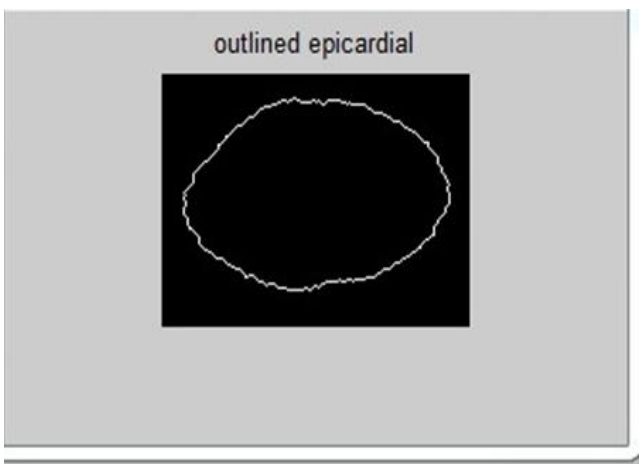

Figure7: Outlined epicardial

The second step is LDDMM (Large Deformation Diffeomorphic Metric Mapping). The affine transform is fully based on a set of thirteen standard points: the left ventricular(LV)Base, the two right ventricular (RV) placing points at the base, the two RV placing points between base and apex, and four sets of two points that evenly divide RV and LV epicardial at base, the above Figure 6 shows the affine transformation of the atlas to match the patient ventricles. In the second step of atlas distortion, the affine-transformed atlas ventricles were further distorted to match the patient geometry, using large deformation diffeomorphic metric mapping (LDDMM).

LDDMM is an image registration process that computes Diffeomorphic (invertible, smooth, and with a smooth inverse) Transformation between images.

D. Deformation field for atlas fiber orientations:

The ventricular fiber orientation of the patient image is obtain to estimate by the atlas and the transformation matrix of the affine matching and the deformation field of LDDMM transformation is applied in sequence of DTMRI image. The morphing of the atlas DTMRI image consisted of spatial repositioning of image voxels in accordance with the spatial transformation of the geometry images, and reorientation of the DTs. Further orientation of the DTs, it considered two methods, namely the preservation of principal directions (PPD) and finite strain (FS).

The estimate of the patient fiber orientations is obtained from the morphed atlas DTMRI image by computing the primary Eigen vector of the DTs[13]. Figure 7 shows a streamlined visualization of the estimated fiber orientations in the patient ventricles. The fiber orientations vary smoothly across the myocardial wall, form a left-handed helix on the epicardium, and appear as a right-handed helix on the endocardium.

E. Measurement of Estimation Error and Its Effects

Examinations of efficiency of the proposed practice are perform on hearts due to the absence of human hearts. Ventricles segmented from a total of six normal and three failing canine hearts, the fiber orientations of all of which were acquired in diastole with *ex vivo* DTMRI at a resolution of 312.5×312.5× 800m are used. The datasets have been employed in previous studies [3],[12],where the reader can find details of the attainment.

IV. RESULT AND CONCLUSION

Figure 8(A)-(C) demonstrates smooth visualization of estimated and also DTMRI derived fiber orientations in normal and failing hearts. Qualitative assessment shows that fiber orientation estimated align well with DTMRI derived ones. Figure 8(D) point up, superimposed on the heart 1 geometry, the error distribution in normal hearts inclination angles, averaged across all five estimates. Figure 8 (E) shows the mean error distribution in failing heart inclination angles, covered the geometry of heart 1. Figure 8(F) and (G)presents sections of tissue from the distribution the error calculated in the each of the three failing hearts of ventricles were mapped onto the heart 1 geometry in the basis of point-to-point correspondences acquired from using affine and LDDMM transformation; the error mapped were averaged. It emphasizes transmural difference of error. The histograms of errors in Fig. 4(H) indicate that there are small error values of most myocardial voxels in normal and failing ventricles have less error than 20 degree respectively. In all estimated datasets, and all image voxels that belongs to the myocardial were found to be 14.4° and 16.9 ° of mean error in the ventricles of normal and failing hearts.

The combined cases of normal and failing heart, the mean error in the entire myocardium were 15.4°. The mean 3-D acute angle between estimated and acquired fiber direction in normal and canine heart were 17.5° and 18.8°, respectively. The 3-D angles are similar to the estimation errors, it shows that the data loss in the recounting fiber orientations by means of insignificant inclination angles. By indicating that the difference in estimation quality from one atlas to another was small by the standard deviation of errors across the five different orientation of fiber of heart 1was just 1.9°

In our technique, the processing pipeline is customized by restoring the PPD method with FS method in the affine transformation steps gives the endocardial region that is shown in figure 9. To compare the performance of the PPD and FS reorientation approach, the endocardial region is modified by replacing PPD method with FS method in affine transformation process. This concludes in predicting the fiber orientation in the failing ventricles, the PDD method outperforms the FS method.

Here, it express the findings about the outcomes of estimation error on the electrophysiological simulations results. The activation maps produced by the estimated fiber orientations are equal to that of acquired orientations; the previous epicardial activations arise at the same position and it matches with the direction of propagation.

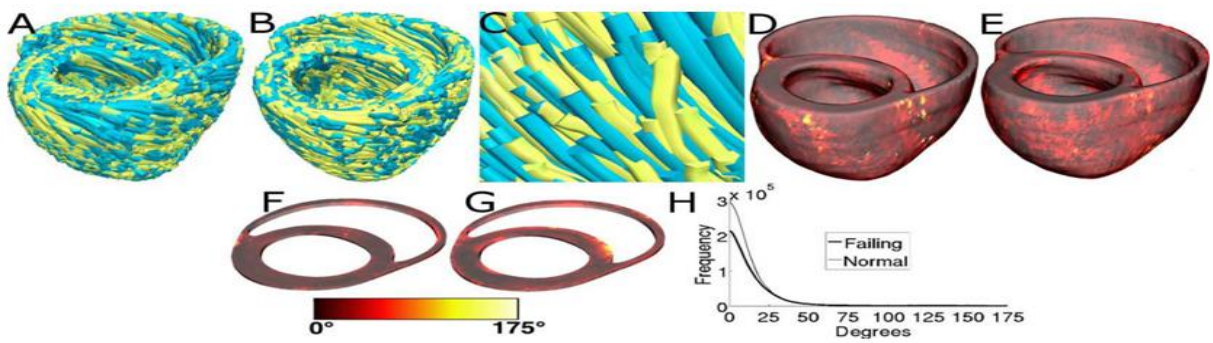

Figure 8: Comparing Estimated fiber orientations with DTMRI derived orientations

Between the normal ventricles, the activation time is151 ms, while the mean total activation time is 154 ms. In normal ventricular models among the acquired and estimated fiber orientation cases the overall mean difference is 5.7 ms. That regions with huge variations in local activation times in the normal ventricles are close to the base, particularly near the RV outflow tract, where the wall is very thinner. Here consider that this thinness of the wall leads imperfect registration and large estimation errors, which in turn in local activation times it cause large difference.

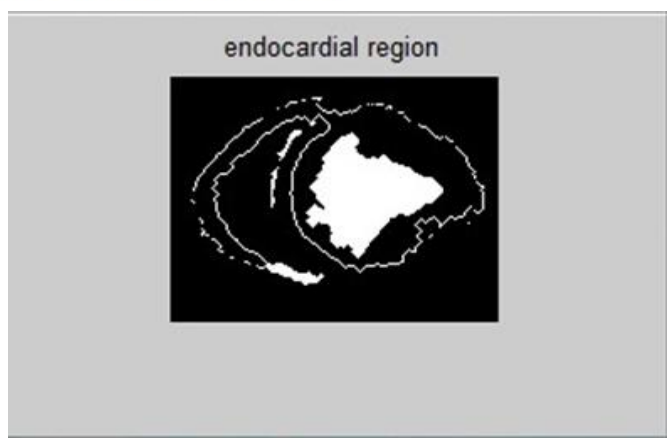

Figure 9: Endocardial region

A significant conduction slowdown near the RV outflow tract has caused due to the large estimation errors. These errors, though, stayed local, and in activation times it did not have a significant posture on the global variations. Figure 10 shows that for sinus rhythm simulation, pseudo-ECGs obtained with models 1and 3 have identical morphologies. 4.4% was the MAD score between these two waveforms. By average, 10.9% is the MAD score among sinus rhythm pseudo-ECG with each of it.

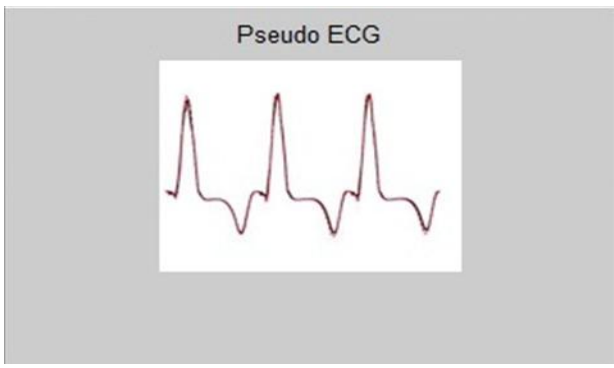

Figure 10:Results from simulation one beat of sinus rhythm

Among this heart models with acquired and estimated fiber orientations, the mean difference in total activation times is only 5.2 ms (3.1%), as the same time the mean MAD score is 1.25%.These results shows that the effect of simulation of ventricular activation in sinus rhythm in normal heart.

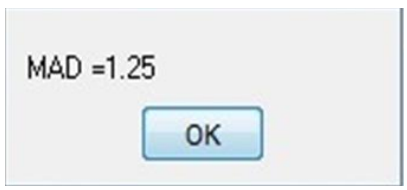

Figure 11 : Mean Absolute Deviation value for sample

The ECG morphologies equivalent to estimated and acquired fiber orientations in hearts. 9.3% was the mean MAD score. These outcomes shows that canine heart failure models with estimated fiber orientations can nearly imitate outcomes of VT simulations execute using acquired fiber orientations.

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