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Stress Distribution in a Rotationally Symmetric and a Measurement Based Left Ventricular Shape Model

Yasuhiro TAKADA*, Akira AMANO*, Jianyin LU‡, Takao SHIMAYOSHI§,
Satoshi ISHIKAWA¶, Tetsuya MATSUDA*, Hidetoshi KOTERA†
*Graduate School of Informatics, †Graduate School of Engineering, Sakyo-ku, Kyoto, Japan 606-8501
‡Leading Project for Biosimulation, Kyoto University, Shimogyo-ku, Kyoto, Japan 600-8815
§ASTEM RI/Kyoto, Kyoto, Shimogyo-ku, Kyoto, Japan 600-8813
¶Mechanical Design and Analysis co, Tokyo, Japan 182-0024
Email: {ytakada, amano, tetsu}@sys.i.kyoto-u.ac.jp, lu@biosim.med.kyoto-u.ac.jp,
kotera_hide@mech.kyoto-u.ac.jp, simayosi@astem.or.jp

***Abstract*— We evaluated the stress distribution in a geometrical shape model and a shape model obtained from human heart using two different fiber orientations. For both orientation models, the results showed large differences of the stress distributions between the mathematical shape model and the maesurement based shape model. These results suggest that stress distribution is highly dependent on the model geometry and the usage of a maesurement based shape model is important for the evaluation of the left ventricular (LV) wall stress distribution. This fact may have some influences on the reported homogeneity of stress distribution with anatomical fiber orientation model that uses mathematical shape model.**

I. INTRODUCTION

Stress and strain distribution over the cardiac wall is a significant determinants for blood flow, oxygen consumption and tissue remodeling. However, experimental measurement is a difficult task, since strain can only be measured accurately at limited sites in the epicardial wall[1]. The reliability of wall stress measurements is limited, since tissue is damaged by insertion of the force transducer. For these reasons, the distribution of stress and strain is assessed by conducting simulation experiments with finite element (FE) models.

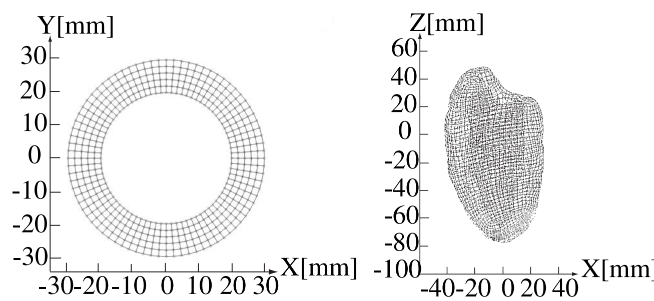
Arts and their group showed that the transmural stress distribution became increasingly homogeneous with anatomical distributions of the fiber orientation. The shape of their model was rotationally symmetric[2]-[6]. Guccione et al. reported that the transmural distribution became inhomogeneous in their shape model. The long axis cross sectional shape of their model was obtained from the real heart, however, the short axis cross sectional shape was symmetric[7].

These studies suggest that a model of mathematically ideal geometry may lead to an even stress distribution. For our investigations presented in this paper, we prepared both a mathematical model of rotationally symmetric shape and a three dimensional left ventricular FE model obtained from human left ventricular MRI data at end diastole. With these models we evaluated the relation between left ventricular shape geometry and stress distribution.

II. METHODOLOGY

A. Finite element model of LV

1) Geometry: In this study, two different shape models were considered for the LV model: A cylindrical shape model (cylinder model) with 80 elements in circumferential direction and 5 layers in transmural direction(Fig.1(a)) and a maesurement based model (real model) constructed from a set of human MRI data. The model consisted of 23418 elements, the details of which are described in [8](Fig.1(b)). The long axis of both models were aligned along the z-axis.

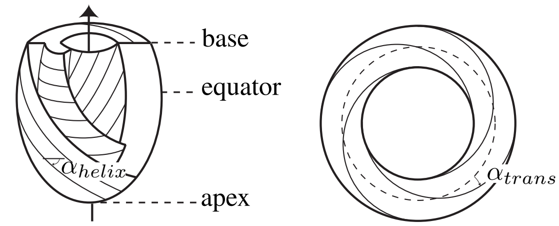


(a) cylinder model (b) real model

Fig. 1. LV model

2) Fiber orientation: The fiber orientation of a model is represented by two angles: α_{helix} and α_{trans}[9](Fig.2). We used a non-physiological model that referred to as simple orientation model and a physiological model that referred to as Huyghe model[10]. The simple orientation model has constant α_{helix} and α_{trans}. The simple orientation model whose α_{helix} equals d degree is called d degree simple orientation model. The Huyghe model is based on an anatomical model assessed by Streeter et al.[9], and is optimized to produce homogeneous mechanical stress distribution. In this study, we used a 30 degree simple orientation model that shows similar ejection fraction and motion to Huyghe model.

3) Myocardial cell model: For the myocardial electrophysiological cell model, the "Kyoto Model"[11] was used. Although there are several other myocardial cell models,



(a) α_{helix} (b) α_{trans}

Fig. 2. Two angles are used to describe fiber orientation.

this model has already incorporated the contraction model. The contraction model used in Kyoto model is the Negroni and Lascano model (NL model)[12] which produces accurate contraction force.

4) Material property: The active contraction force generated by the myocardial cell is dealt as the external force of each node of the FE model. The direction of the force is determined by the fiber orientation of each element.

The NL model uses $F_p = K_p(L_0 - L)^5$ to simulate the parallel elastic component which can be recognized as the passive tissue material property in the FE model. This extension force is nonlinearly related to the half sarcomere length $L[\mu\text{m}]$, where $L_0(= 0.97[\mu\text{m}])$ is the resting half sarcomere length and K_p is a constant.

For simplicity, we used a linear elastic component as material property. This may lead to errors in the resulting stress distribution in the LV tissue. When the sarcomere length is in the physiological range, however, the deviation of the the linear elastic equation from the equation above is neglectable.

B. Simulation system

As the electrophysiological model of myocardial cell and the FE model should be calculated simultaneously, we need a coupling simulation system. We used a strong coupling system for the LV motion simulation in a distributed simulation environment[13].

C. Quantification of mechanical stress

Regional mechanical stress was quantified by the equivalent Von-Mises stress at end systole. The transmural position of LV wall was normalized with respect to percent depth, where 0% depth corresponds to the endocardium and 100% depth corresponds to the epicardium. The circumferential position is given by the rotational angle with respect to the x-axis. In additions, we define the curved line of constant depth in short axis cross sections as short axis iso-depth line. In the same way, we also define one in long axis cross sections as long axis iso-depth line.

III. RESULTS

A. Simulation results obtained with the cylinder model

Transmural stress distribution of the 30 degree simple orientation model and the Huyghe model are presented in Fig.3. The regional stress of the simple orientation model is monotonously decreasing from endocardium to epicardium. In contrast, the regional stress of Huyghe model is lowest at 80% depth and increases toward endocardium and epicardium.

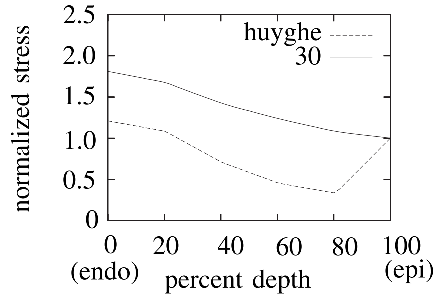
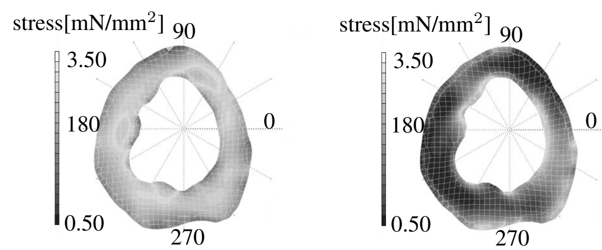


Fig. 3. Transmural stress distribution of the cylinder model. The regional stress of the simple orientation model is monotonously decreasing from epicardium to endocardium. The regional stress of the Huyghe model is lowest at 80% depth and increases toward endocardium and epicardium.



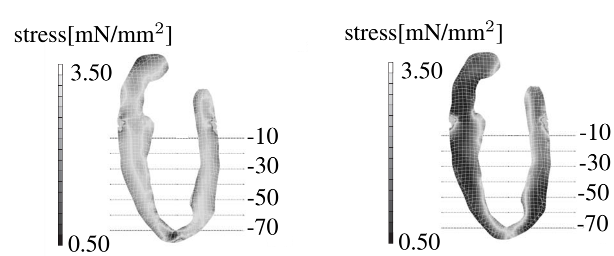
(a) simple orientation model (30 degree) (b) Huyghe model

Fig. 4. Circumferential stress distribution of the real model. The results show inhomogeneity and dependency on the fiber orientation model.

B. Simulation results on the real model

Short axis and long axis cross sectional view of the regional stress distributions in the 30 degree simple orientation model and the Huyghe model are presented in Fig.4 and Fig.5, respectively. The short axis view corresponds to the xy plane at $z = -20$. The long axis view corresponds to the xz plane ($\theta = 0$, $\theta = 180$). In contrast to the cylinder model, the stress distribution of the real model shows inhomogeneous in circumferential and longitudinal direction for both fiber directions. Generally, the stress of the simple orientation model is higher than that of the Huyghe model.

In short axis cross sections at $z = -20, -40, -60$, we calculated the average and the standard deviation (SD) of the stress along short axis iso-depth line from $\theta = 0$ to



(a) simple orientation model (30 degree) (b) Huyghe model

Fig. 5. Longitudinal stress distribution of the real model. The results show inhomogeneity and dependency on the fiber orientation model.

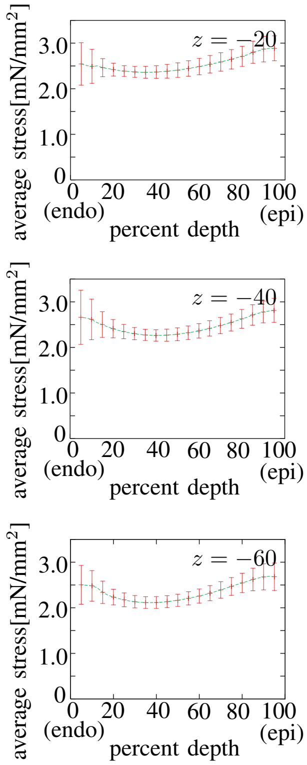


Fig. 6. Circumferential stress average and variance of the simple orientation model. (30 degree)

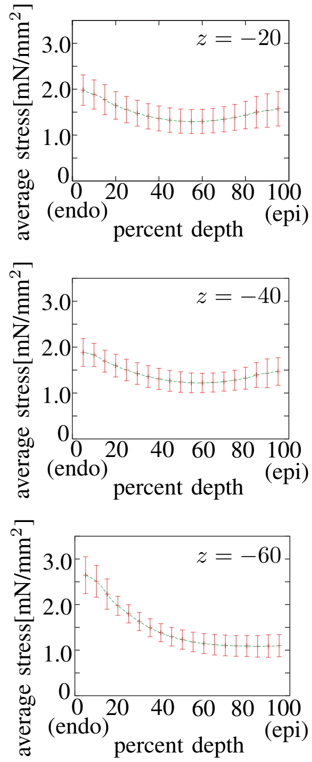
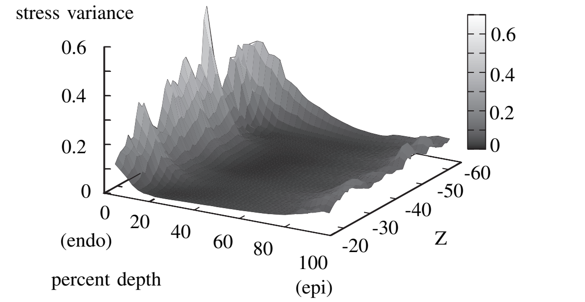
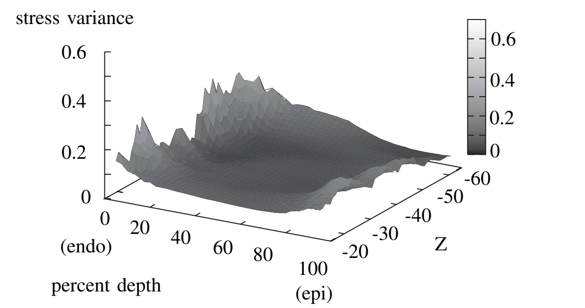


Fig. 7. Circumferential stress average and variance of Huyghe model.



(a) simple orientation model (30 degree)



(b) Huyghe model

Fig. 8. Circumferential stress distribution for transmural and longitudinal direction. For both models, the variance increases at the endocardium. However, the variance of the Huyghe model is relatively even compared to the simple model.

$\theta = 360$ (Fig.6, 7). The error bar represents the SD. These results demonstrate that for the simple orientation model, the average stress is almost even in each z coordinate which is similar in the Huyghe model at $z = -20$ and $z = -40$, while it rapidly decreases from endocardium to epicardium at $z = -60$. In addition, the average stress in the simple orientation model is generally larger than that in the Huyghe model. To evaluate the circumferential stress variance from $z = -15$ to $z = -65$, we calculated it along short axis iso-depth line from $\theta = 0$ to $\theta = 360$ in each z coordinate (Fig.8). The result shows that the stress variance in short axis cross sections increases towards endocardium for both models. However, the stress variance of the Huyghe model is relatively even compared to that of the simple orientation model.

In long axis cross sections at $\theta = 30, 150, 270$, we calculated the average and the SD of the stress along long axis iso-depth line from $z = -15$ to $z = -65$ (Fig.9, 10). The error bar represents the SD. These results demonstrate that for both models, the average and the SD stress differ when θ position changes. The average stress is relatively even at $\theta = 30$ and 150 in the simple orientation model and at $\theta = 270$ in the Huyghe model, while it changes from endocardium to epicardium at $\theta = 270$ in the simple orientation model and at $\theta = 30$ and 150 in the Huyghe model. To evaluate the longitudinal stress variance from $\theta = 0$ to $\theta = 360$, we calculated it along long axis iso-depth line from $z = -15$ to $z = -65$ in each θ position (Fig.11). For both models, stress variance varies along θ position.

IV. DISCUSSION

In this study, we evaluated and compared the stress distribution of different fiber orientation models by using a mathematical shape model and a measurement based model of the real human heart.

For the cylinder model, the stress of the 30 degree simple orientation model was monotonously decreasing from endocardium to epicardium, while that of the Huyghe model showed V-shaped. Previous work of Huyghe [14] reported that transmural stress distribution was homogeneous when the paramaters of Huyghe model was optimized. However, our study yielded different results. Such difference might occur, since we did not consider inner cavity pressure and anisotropic material property.

For the Huyghe model, the V-shaped change for depth of wall that was seen in cylinder model was not demonstrated in the real model. For the simple orientation model, the stress distribution of the cylinder model decreased monotonously from endocardium to epicardium, while the stress distribution of the real model demonstrated only a little change. Our simulation results showed that the transmural stress distribution is highly dependent on model geometry.

When a mesurement based LV geometry model is used, transmural distribution may be different from the reported ones that use mathematical shape model. Therefore a measurement based shape model is important for the evaluation of the LV wall stress distribution.

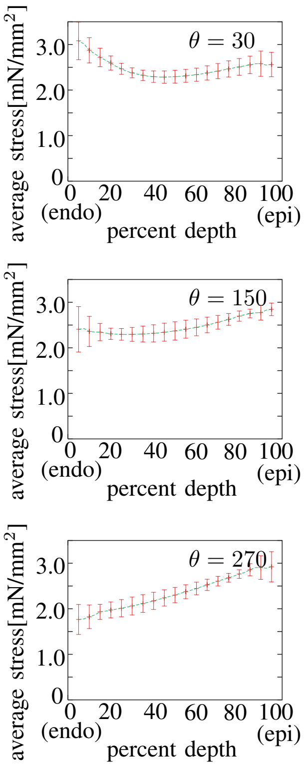


Fig. 9. Longitudinal stress average and variance of simple orientation model. (30 degree)

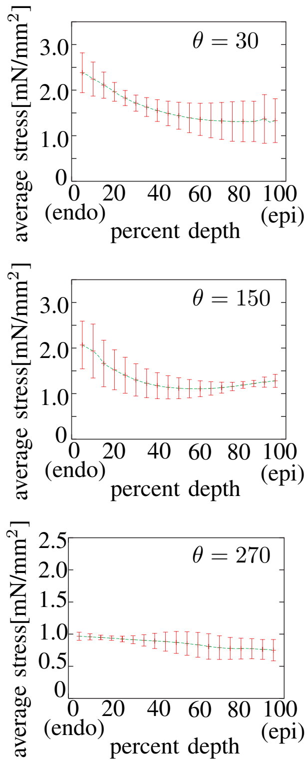
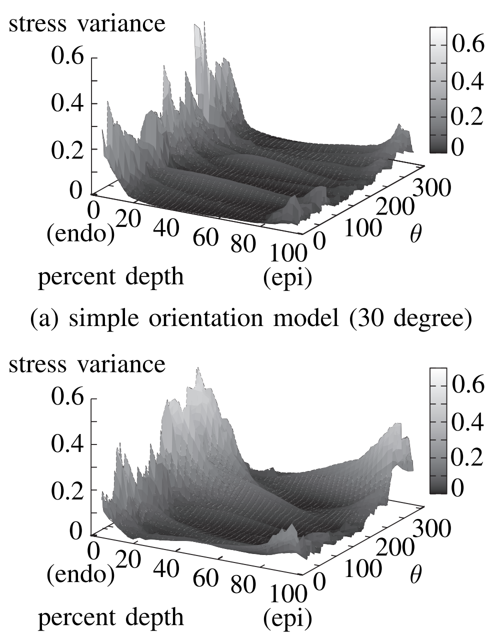


Fig. 10. Longitudinal stress average and variance of Huyghe model.



(a) simple orientation model (30 degree)

(b) Huyghe model

Fig. 11. Longitudinal stress distribution along transmural and circumferential position. The regions of high stress are different within the models.

V. CONCLUSION

In this study, we evaluated the stress distribution by using a mathematical model of rotationally symmetric shape and a mesurement based model of the real human heart with different fiber orientation models. Essential differences between the stress distribution of mathematical shape model and the measurement based shape model was demonstrated. The results suggest that a measurement based shape model is important for the evaluation of the LV wall stress distribution.

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