Electronics and Communications in Japan, Part 2, Vol. 90, No. 12, 2007
Translated from Denshi Joho Tsushin Gakkai Ronbunshi, Vol. J88-D-II, No. 5, May 2005, pp. 943–953

Model Generation Interface for Simulation of Left Ventricular Motion

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SUMMARY

In the field of life science, a tremendous amount of quantitative data related to biological systems continues to accumulate. There are still many aspects of biological phenomena which are not fully analyzed, due to interaction between various phenomena and mechanisms. The simulation technique provides useful approaches to the analysis of such phenomena, in which the model describing the phenomenon and the model parameters are adjusted. Usually, however, simulation models for biological functions are very complex, and a tremendous amount of time is required in comprehensively adjusting the model and its parameters and then evaluating the simulation results. This study aims at construction of a model for the left ventricular motion, and considers the left ventricular shape model, the cell orientation model, and the coronary artery model as its components. An interface is constructed by which the above models and the model parameters can be adjusted. Using this interface, it is possible to evaluate the simulation model efficiently. © 2007 Wiley Periodicals, Inc. Electron Comm Jpn Pt 2, 90(12): 87–98, 2007; Published online in Wiley InterScience (www.interscience.wiley.com). DOI 10.1002/ecjb.20425

Key words: biological function simulation; left ventricular motion simulation; myocardial cell model.

1. Introduction

With the progress of research in the field of life science, a tremendous amount of quantitative data related to biological systems, such as the human genome sequence, continues to accumulate. However, these data are mostly concerned with particular phenomena or mechanisms of microscopic structure. It is understood that biological phenomena are realized through a complex interaction between these phenomena and mechanisms; however, there are still many points which have not been fully analyzed in the interactive aspects.

Consequently, it is becoming important to understand biological phenomena by integrating various processes. The Physiome Project [1] is intended to promote research in this direction. One of the strongest tools for analysis in such an approach is simulators for biological functions.

In research related to the Physiome Project, the most advanced biological model is the electrophysiological model of cells, especially myocardial cells. Using such a precise cell model and components, there have been efforts to construct precise models for tissues and organs. There have been especially intensive studies of tissue and organ models for the heart [2, 3]. It has been revealed that in addition to the precise cell model, various models such as cell orientation models and models of the mechanical properties of tissue should be combined in order to construct a precise overall model.

In early studies, the mechanical deformation of the heart has been considered as a deformation problem for a model with the properties of linear elastic material, in which the two-dimensional short-axis cross section or a spheroid is used as the shape model. With an increasing volume of physiological findings, research has advanced to the use of the model with more precise material properties, shape, and cell orientations.

The cell orientation in the heart is indispensable information in the discussion of the anisotropic material properties of heart tissue. An example of an investigation of the effect of anisotropy on the left ventricular motion is the study by Yettram and colleagues [4]. Their paper considers the anisotropic material properties, in contrast to the conventional material property model of the heart muscle in which an isotropic linear elastic material is considered. Deformation simulation of the two-dimensional short-axis cross section has shown that the stress distribution is made more uniform by anisotropy.

The nonlinear material property of the myocardial tissue has been measured precisely by Lin and Yin [5]. In this paper, the material property of both resting and the force generating myocardial tissue are modeled.

Investigations concerning the three-dimensional cell orientation model include the study by Bovendeerd and colleagues [8], in which it is shown that the stress distribution is made more uniform by optimizing the cell orientation model. Specifically, the results are as follows. In the model in which the orientation of the cells on the short-axis cross section changes linearly from –60° to 60°, from the endocardium to the epicardium, the stress along the fiber direction ranges from 30 kPa to 100 kPa. When the cell orientation is optimized, in contrast, the stress ranges from 52 kPa to 55 kPa in the model with a midwall angle of –15°. They also performed a simulation for ischemia [9].

Investigations based on precise cell orientation models obtained by actual measurement include study by Guccione and colleagues [6, 7], which uses a cell orientation model based on real heart measurements in which the cell orientation changes from the endocardium to the epicardium [17, 18]. The effect of the rate of change of the cell orientation on the epicardium stress distribution at end systole is evaluated by simulation. In this study, the spheroidal model is used as the shape model. It is shown that a spheroidal coordinate system is more effective in calculations by the finite-element method. The nonlinear property derived from their measurements is used as the material property of the myocardial tissue, but the cell model is not used in the contraction force generation.

For simulations of the ischemic state, it is considered useful to construct a model of the coronary artery and to evaluate the effect of changes in the myocardial infarction region on cardiac function. As regards the construction of models for the coronary artery, there is a study by Smith and colleagues in which the results of actual measurements are presented and a simulation of the coronary artery is constructed [10, 11]. However, this study gives no evaluation of the relation between the myocardial infarction region and cardiac function.

There are many studies of the propagation of membrane excitation in the whole heart [12]. A simulation of ECG generation has been performed utilizing the results of such a study [13]. In most of these studies, a particular phenomenon, such as the stress distribution in the normal heart or the reproduction of the ECG, is considered. Then the simulation is performed using a simulation model which consists of the minimum model or structure that can reproduce the phenomenon under consideration. Usually, no account is taken of how known elements not considered in the simulation can affect the results. An example is as follows. In a study in which the relation between the cell orientation and the stress distribution in the normal heart is evaluated, the effect of the cell orientation or the nonlinear material properties of the tissue on the stress distribution in the ischemic heart is not examined.

From the viewpoint of analysis of biological phenomena and the examination of the validity of models, it is very important to construct various element models and to evaluate simulation results efficiently by varying the kind of model and its parameters. Such a model often has a complex structure and contains many parameters. Usually, a long time is required to construct the model.

Examples of past studies of the interface or tools to construct or modify biological simulation models include generation tools for the gene pathway model [14, 15]. However, these models consider only the metabolic system, and there has been no study of tools to construct or modify composite models built up from models concerned with various different phenomena. As regards the simulation of heart motion, there is a tool that sets the boundary conditions or performs computation by the finite-element method [2], but there is no report of tools that construct or modify models covering multiple phenomena.

In this context, this paper considers the construction of a simulation model for left ventricular motion, and proposes an interface for constructing a simulation model in which the kinds of element model or the parameters can be easily adjusted. A basic simulator is constructed, which calculates the pulsation of the heart period based on the constructed simulation model, and it is demonstrated that the effects of the model and the parameters on the motion can be efficiently evaluated.

2. Model of Left Ventricular Motion

2.1. Components of model for left ventricular motion

In constructing the simulation model for left ventricular motion, the following element models must be considered as components of the whole model.

- Three-dimensional shape model for left ventricle

This is the model that represents the three-dimensional shape of the left ventricle.

- Cell orientation model

This is the model that represents the orientation of the myocardial cells in the left ventricular wall.

- Coronary artery model

This is a model representing the coronary artery, which provides oxygen and nourishment to the myocardial tissue by means of the blood.

- Conducting system model

This is a model that represents the three-dimensional arrangement of the conducting system, which is the conduction path for the periodic excitation generated in the pacemaker cells at the sinuatrial node near the right atrium to the whole heart, and also its electrophysiological properties.

- Membrane excitation propagation model

This is the model for the process in which the excitation of the myocardial cell generated from the conducting system propagates through the myocardial tissue.

- Whole-body circulation model

This is a fluid model that represents the hemodynamics, such as blood pressure, blood flow, and vessel resistance.

- Fluidic properties of blood

These are the fluid properties of the blood, which are necessary in calculating the flow and pressure of the blood in the ventricle and the atrium.

- Electrophysiological model of myocardial cells

This is a model that represents the activity of myocardial cells, composed of functional elements of the cell such as the ion channel and the contraction force generation mechanism.

- Microcirculation model

This is a model that represents substance transfer from the capillary vessels, which is an important mechanism in maintaining cell metabolism.

- Material properties

This is a model that represents the mechanical properties of the left ventricular tissue, such as elasticity and viscosity.

The motion of the left ventricle is considered as the result of the complex interaction of these models. At present, the accuracy of these models is not sufficient. It is thus important to modify the kinds of proposed element models or the parameters of the element models so that the effect of these models and parameters on the left ventricular motion can be evaluated.

The simulator for the left ventricular motion which was constructed in this study can handle the three-dimensional shape model, the cell orientation model, the coronary artery model, the cell physiological model, and the material property model, which have become relatively precise as a result of physiological experiment or measurement. In the modification of the parameters of the element models, existing tools like CAD systems can sometimes be utilized, as in the cases of the three-dimensional shape model and the material property model. An interface that can modify the parameters of the element models, for which existing tools cannot handle, and can also modify the combination of element models, is constructed. Specifically, the modification interfaces are constructed as follows for the three-dimensional shape model of the left ventricle, the cell orientation model, and the coronary artery model.

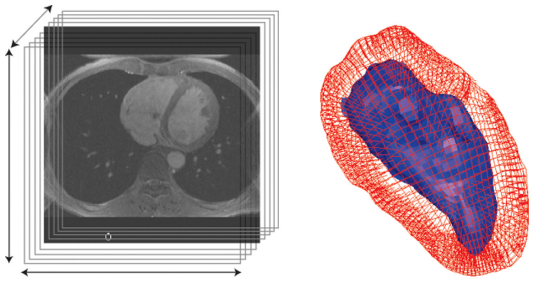
The simulator implemented at the present stage can handle cell physiological phenomena and deformations by structural mechanics, but cannot handle fluid dynamics or electric conduction phenomena. Consequently, the conducting system model, the whole-body circulation model, the blood fluid dynamics model, and the microcirculation model have not yet been implemented. Thus, the cell physiological phenomena and the mechanical deformations of the actual heart are not precisely reproduced in the simulation results; however, it is possible to evaluate on a relative basis how the model affects a particular phenomenon, such as the shape model and rotation around the long axis at end systole. We plan to continue the implementation of the models which have not yet been implemented.

2.2. Three-dimensional model of left ventricle

In the finite-element calculation of the deformation of the beating left ventricle, the mesh model is used as the three-dimensional shape model for the left ventricle. The human left ventricle is considered in this study.

From the set of human chest MR images [Fig. 1(a)] at systole, the region corresponding to the left ventricular wall was extracted, and three-dimensional image data for the left ventricle were constructed. The MR images used in the experiment were a set of cross-sectional images of the human chest and upper abdomen. Each image was composed of 256×256 pixels with pixel size of 1 mm $\times$ 1 mm. The set of images consisted of 65 slice images taken at 2-mm intervals along the body axis.

The isosurface was generated for the three-dimensional left ventricular image, and a hexahedral mesh was generated by a mesh generation method based on the cognitive model [16]. Finally, the long axis of the left ventricle was determined manually and the shape was rotated so that the long axis was the Z axis of the three-dimensional shape model. Figure 1(b) shows the generated mesh data. In the figure, the endocardium of the left ventricle is displayed as



(a) The set of chest MR images (b) Hexahedral mesh data

Fig. 1. Generation of LV 3D shape model.

the surface. The hexahedral mesh corresponding to the ventricular wall is displayed as a wire frame.

2.3. Cell orientation model

It is known that the orientation of the myocardial cells on the left ventricular wall depends on the position of the heart. The apex and base of the left ventricle are denoted as "apex" and "base," respectively. The endocardium, that is, the inner face of the left ventricular wall, and the epicardium, that is, the outer face of the left ventricular wall, are denoted as "endo" and "epi," respectively.

The axis directed from the apex to the base is called the left ventricular long axis. The cross section with the maximum radius, among the cross sections perpendicular to the left ventricular long axis, is called the equator. The inclination of the cells from the plane perpendicular to the left ventricular long axis is defined as α_{helix} [Fig. 2(a)], and the inclination of the cells from the tangent of the circle on the plane perpendicular to the left ventricular long axis is defined as α_{trans} [Fig. 2(b)]. The orientation of the cells is represented by these angles.

In this study, two models are provided in order to evaluate the effect of the cell orientation on the wall motion. One is the "simple cell orientation model," in which α_{helix} and α_{trans} are kept constant in the heart wall. The other is the model proposed by Huyghe and colleagues (called the Huyghe model), which is based on the measurements of cell orientations in the real heart.

2.3.1. Coordinate system

The three-dimensional shape model is represented by orthogonal coordinate system *XYZ*. The Huyghe model is represented in the prolate spheroidal coordinate system ΞΘΦ. When the same point is represented by the coordinates (ξ, θ, ϕ) in the prolate spheroidal coordinate system and the coordinates (*x*, *y*, *z*) in the orthogonal coordinate system, the coordinates are related as follows:

$$x = C \sinh\xi \sin\theta \cos\phi \tag{1}$$

$$y = C \sinh\xi \sin\theta \sin\phi \tag{2}$$

$$z = C \cosh\xi \cos\theta + \delta_z \tag{3}$$

where *C* is the focal length and δ_z is the parallel translation component in the *Z* direction.

Figure 3 shows the axes of the prolate spheroidal coordinate system in the space of the orthogonal coordinate system. As can be seen from Fig. 3(c), the curve for constant θ in the prolate spheroidal coordinate system gives a hyperbola, which is always orthogonal to the curve for constant ξ on the cross section for any ϕ. In the prolate spheroidal coordinate system, ξ = 0 gives a straight line, which is the *Z* axis in the orthogonal coordinate system.

2.3.2. Simple cell orientation model

The simple cell orientation model is a model in which α_{trans} is 0° and α_{helix} remains constant at any point on the heart wall (Fig. 4). At the point (*x*, *y*, *z*) represented by the orthogonal coordinate system, the cell orientation (v_x, v_y, v_z) is expressed as follows:

$$v_x = \cos\phi \cos\alpha_{helix} \tag{4}$$

$$v_y = \sin\phi \cos\alpha_{helix} \tag{5}$$

$$v_z = \sin\alpha_{helix} \tag{6}$$

$$\phi = \tan^{-1}(y/x) \tag{7}$$

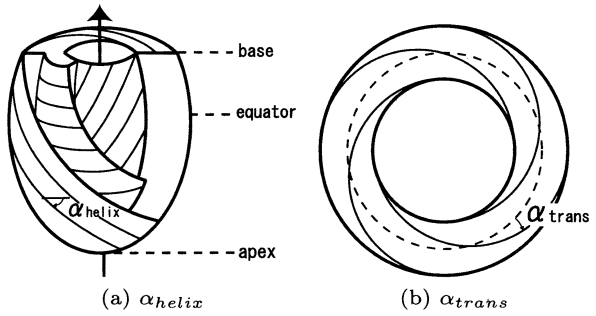


Fig. 2. Two angles which denote fiber orientation.

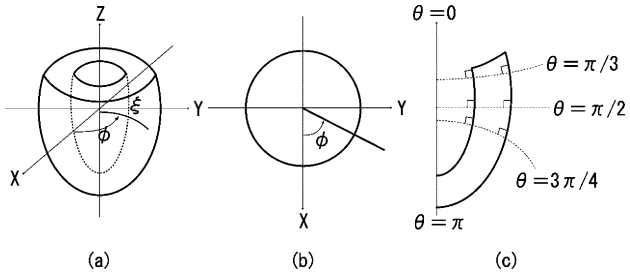


Fig. 3. Prolate spheroidal coordinate system.

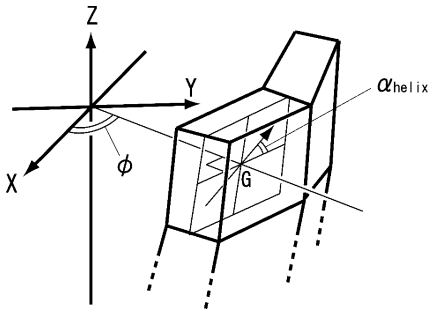


Fig. 4. Simple cell orientation model.

In the following, a simple cell orientation model with α_{helix} kept as d deg is called the simple $d°$ cell orientation model.

2.3.3. Huyghe model

Streeter and colleagues [17] actually measured α_{helix} in dog heart, and based on the result, proposed the following mathematical model for α_{helix} (Fig. 5):

$$\alpha_{helix}(\bar{\xi}) = \alpha_{h0} + \alpha_{h1}\bar{\xi} + \begin{cases} 4\alpha_{h2,en}(\bar{\xi}+0.5)^2 & -1.0 \le \bar{\xi} < -0.5 \\ 0 & -0.5 \le \bar{\xi} \le +0.5 \\ 4\alpha_{h2,ep}(\bar{\xi}-0.5)^2 & +0.5 < \bar{\xi} \le +1.0 \end{cases} \quad (8)$$

where $\alpha_{h0} = 15°$, $\alpha_{h1} = -60°$, $\alpha_{h2,en} = -15°$, and $\alpha_{h2,ep} = -15°$.

For α_{trans}, Huyghe and colleagues [18] proposed the following mathematical model, based on the measurements of dog heart:

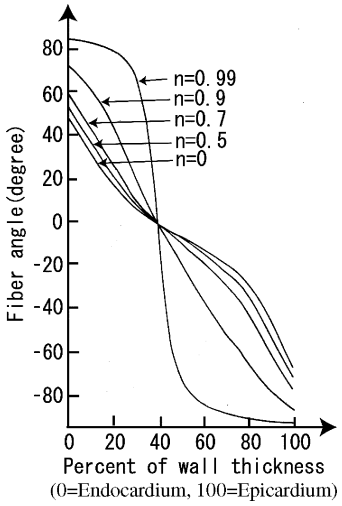


Fig. 5. Transmural distribution of α_{helix} in Streeter model.

$$\alpha_{trans}(\bar{\xi},\bar{\theta}) = \begin{cases} 4\alpha(1-\bar{\xi}^2)\bar{\theta}^2 & \bar{\theta} > 0 \\ -\alpha(1-\bar{\xi}^2)\bar{\theta}^2 & \bar{\theta} \le 0 \end{cases} \quad (9)$$

Here, $\bar{\xi}$ and $\bar{\theta}$ are the ξ and θ coordinates in the prolate spheroidal coordinate system described in Section 2.3.1, normalized to the left ventricular shape. $\bar{\xi}$ is the value of ξ normalized so that it is 0 on the endocardium and 1 on the epicardium. $\bar{\theta}$ is the value of θ normalized so that it is –1 at the apex and 0.5 on the base. α is a constant, which is set as $\alpha = 13.5°$ in Ref. 18.

Figure 6 shows the behavior of α_{trans} from the endocardium to epicardium in this model. The solid line shows the behavior at the base. The dot-dashed line shows the behavior on the equator ($\theta = \pi/2$). The dotted line shows the behavior at the apex. Figure 7 shows the behavior of α_{trans} from the apex to the base. In this study, the above two models are referred to together as the Huyghe model.

2.4. Coronary artery model

The coronary artery is the artery which provides nourishment and oxygen to the heart muscle cells. In diseases, such as myocardial infarction, where the blood flow partially stops, the contractile function of the myocardial cells is locally decreased and the blood flow pumped to the whole body is decreased. In the simulation of left ventricular motion, it is important to introduce the coronary artery model to reproduce and analyze pathological states such as myocardial infarction.

In order to construct the shape model for the coronary artery based on actual data, the coronary artery region was extracted from the set of MR images described in Section 2.2. The resolution of the MR images was 1 mm, and it was difficult to extract arteries with a smaller diameter. Consequently, the artery was added manually based on the positions of the extracted arteries.

In evaluation of pathology such as myocardial infarction, it is important to determine the coronary artery territories which represent the correspondence between the point of the coronary artery where the blood flow is stopped and the area of the heart muscle which loses the contractive function when the blood flow at that point in the coronary artery is stopped. It is understood in general that normal

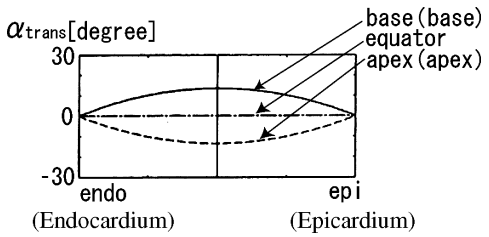


Fig. 6. Transmural distribution of α_{trans}.

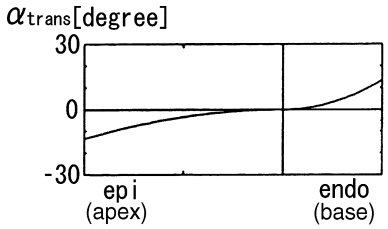


Fig. 7. Distribution of α_{trans} from apex to base.

human heart muscle tissue receives oxygen and nourishment from only one coronary artery [19]. From that viewpoint, it is assumed that the coronary artery providing oxygen and nourishment to each mesh in the three-dimensional left ventricular shape model is the vessel composing the coronary artery which is the closest to the mesh.

2.5. Cell physiological model

In this study, the "Kyoto Model" [20], proposed by Noma and colleagues, is used as the electrophysiological model for the myocardial cell. The myocardial cell is a long, thin cell with an approximate size of 10 μm × 7 μm × 100 μm. The cell generates a specific action potential in response to electrical stimuli from the pacemaker cells. When stimulated, the calcium concentration increases rapidly in the cell, and a contracting force is generated in the long axis direction. These activities are produced by functional elements, such as the ion channel on the cell membrane, and also intracellular mechanisms such as the contraction generation mechanism in the cell. The Kyoto Model models these functional elements precisely and can reproduce the action potential and the contracting force very accurately. In this study, the contracting force generated by each element of the three-dimensional left ventricular model is calculated by the Kyoto Model.

2.6. Material property model

It is known that the material property of the myocardial tissue, which represents the deformation characteristics of the object in response to a given external force, has strong nonlinearity. It is also known that the property changes depending on the physiological state of the cell [5]. Hitherto, however, no reliable measurements have been obtained, since precise measurement is very difficult.

Consequently, linear elastic elements are used as for the material property available at the present stage. Young's modulus and Poisson's ratio are specified as the parameters of the material property model.

3. Interface for Simulation Model Construction

The existing interfaces for three-dimensional shape model generation and modification, such as the three-dimensional CAD system, cannot handle some of the element models. For these element models, an interface is constructed which can interactively modify the model or the model parameters (Fig. 8).

3.1. Interface for left ventricular region extraction

The left ventricular region is extracted from a set of MR images, taken as a large number of two-dimensional cross-sectional images of the three-dimensional chest region. The interface for two-dimensional image region extraction is constructed for this process (Fig. 9). The operator observes three cross sections, namely, the transversal, sagittal, and coronary cross sections, and specifies the region to be extracted by using a mouse on the two-dimensional image. A three-dimensional smoothing filter is applied to the generated three-dimensional image data, and the result is defined as the three-dimensional left ventricular image.

The coordinates of each cell orientation model are defined with the long axis of the left ventricle as the reference. The operator of this interface defines the long axis of the left ventricle by specifying two points on the axis, while observing the images in three directions.

3.2. Interface for generation and modification of cell orientation model

3.2.1. Determination of *C* in prolate spheroidal coordinate system

The Huyghe model described in Section 2.3.3 is defined in the prolate spheroidal coordinate system repre-

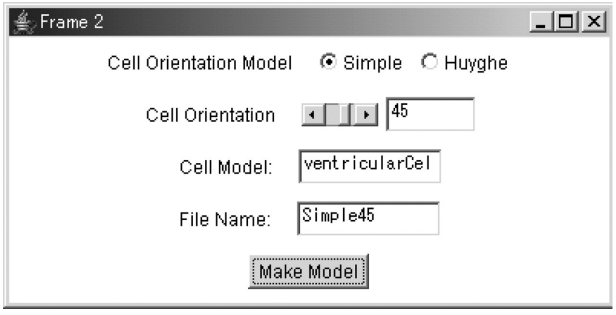


Fig. 8. One simulation model generation interface panel of the left ventricle motion simulation.

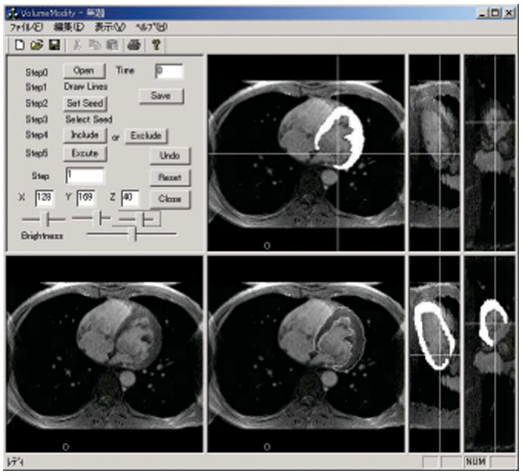


Fig. 9. Interface for extracting LV area.

sented by Eq. (3). In order to establish the correspondence between this coordinate system and the orthogonal coordinate system, the focal length C must be determined. It is desirable in the Huyghe model that the endocardium and the epicardium form constant ξ surfaces in the prolate spheroidal coordinate system. For this purpose, the coefficient C and the magnitude of the parallel translation δ_z in the Z direction must be determined.

Let the mean and the standard deviation of the ξ coordinate of the vertices composing the endocardium and the epicardium be ξ_{mean} and ξ_{dev}, respectively. Then, C and δ_z are determined by the following minimization:

$$\min_{C,\delta_z} \frac{\xi_{dev}}{\xi_{mean}} \tag{10}$$

In the prolate spheroidal coordinate system, ξ cannot be defined at point $(0, 0, z)$ on the Z axis such that $-C < z < C$. Consequently, the above optimization should be performed so that the absolute value of the Z coordinate of any points composing the endocardium is greater than C.

3.2.2. Three-dimensional heart shape model and prolate spheroidal coordinate system

The Huyghe model is represented by $\overline{\xi}$ and $\overline{\theta}$ obtained by normalizing the ξ and θ coordinates in the prolate spheroidal coordinate system. The $\overline{\xi}$ coordinate of the point in the myocardial tissue is calculated from the ξ coordinates of the intersections of the constant θ curve with the endocardium and the epicardium, respectively, as shown in Fig. 10. The $\overline{\theta}$ coordinates are calculated from the θ coordinates at the base and the apex.

Consider a point $g = (\xi_g, \theta_g, \phi_g)$ in the prolate spheroidal coordinate system. Let the values of ξ at the intersections of the curve $\theta = \theta_g$ and $\phi = \phi_g$ with the endocardium

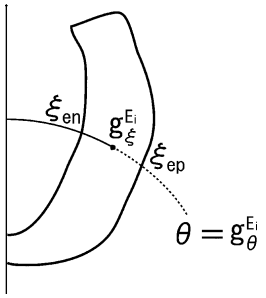


Fig. 10. Relations between ξ_{en}, ξ_{ep} and endocardium, epicardium.

and the epicardium be ξ_{en} and ξ_{ep}, respectively. Let the value of θ at the base be θ_{base}. Then, the values α^g_{helix} and α^g_{trans} of α_{helix} and α_{trans}, which represent the cell orientation at that point, are calculated as follows, using Eqs. (9) and (8):

$$\overline{\theta}_g = -1.5 \times \frac{\theta_g - \theta_{base}}{\pi - \theta_{base}} + 0.5 \tag{11}$$

$$\overline{\xi}_g = \frac{\xi_g - \xi_{en}}{\xi_{ep} - \xi_{en}} \tag{12}$$

$$\alpha^g_{helix} = \alpha_{helix}(\overline{\xi}_g) \tag{13}$$

$$\alpha^g_{trans} = \alpha_{trans}(\overline{\xi}_g, \overline{\theta}_g) \tag{14}$$

3.3. Interface for coronary artery blood flow modification

When the blood flow stops at a particular point in the coronary artery, oxygen and nourishment are not provided. Thus, there is an area in which the contracting force is not generated. The relation between the point of blood flow obstruction and the area not generating contraction force is determined by the procedure described in Section 2.4. In order to simulate the motion of the left ventricle when myocardial infarction occurs, an interface was prepared in this study to construct the following simulation model (Fig. 11). An arbitrary point on the coronary artery is specified. Then the contracting force is set as 0 in the corresponding area.

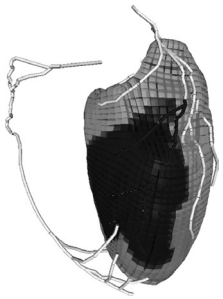


Fig. 11. Interface for myocardial infarction model.

The left ventricle and the coronary artery are displayed on the interface in three dimensions. By specifying arbitrary position on the coronary artery, the area of myocardial infarction in the left ventricle is indicated by changes of color.

4. Simulation of Left Ventricular Motion

In order to verify the effectiveness of the interface for constructing the model of left ventricular motion, a preliminary simulator was implemented. The implemented simulator uses a three-dimensional hexahedral mesh shape model as the component of the model for the left ventricular motion. The contracting force calculated by the cell physiological model is assumed to occur in the mesh in the direction determined by the cell orientation model.

If the area of myocardial infarction is specified in the coronary artery model, the contracting force is not produced in the specified mesh. The change of shape due to the contracting force produced in the mesh is calculated by the finite-element method. For the material property used in the finite-element method, the parameters specified by the material property model are used.

The finite-element method and the cell physiological model are coupled by performing the calculations alternatively with small intervals. In the simulation model at the present stage, it is assumed that the same contracting force is produced in all elements. Thus, the contracting force calculated from a cell model is used in all elements.

5. Experiment

5.1. Experimental conditions

The three-dimensional shape model used in the simulation for the left ventricular motion is a hexahedral mesh model composed of 7638 nodes and 5612 elements. As the cell orientation models, simple orientation models with α_{helix} of 0°, ±30°, ±45°, ±60°, and 90° and the Huyghe model are used. The magnitude of the contracting force is calculated as follows. The contracting force per unit volume is derived from the result of calculation by the Kyoto Model. Then, the volume of each hexahedral mesh is multiplied by a constant k.

For the material properties, an elastic material with a Young's modulus of 20 kPa and a Poisson's ratio of 0.49 was used. When a contracting force of about $k = 1$ was used in the experiment to calculate the shape change of the left ventricle, the computation became unstable because of too large contracting forces. Thus, the maximum value of k for which the shape returns almost to the initial shape at the end of a period was sought. The obtained value of k was $k = 2.0 \times 10^{-3}$.

The computer used in the experiment was as follows. The CPU was Pentium4 (3GHz). The memory was 1 Gbyte. The OS was Red Hat Linux 8. Marc (MSC Corp.) was used as the FEM solver.

For evaluation of the simulation result, the ejection fraction (EF), which is a general evaluation parameter for cardiac function, was used. It is generally understood that the normal value is 60 to 80% [21, 22]. In the real heart used for acquisition of the three-dimensional left ventricular shape model in this experiment, the ejection fraction was 66.7%.

5.2. Experimental results

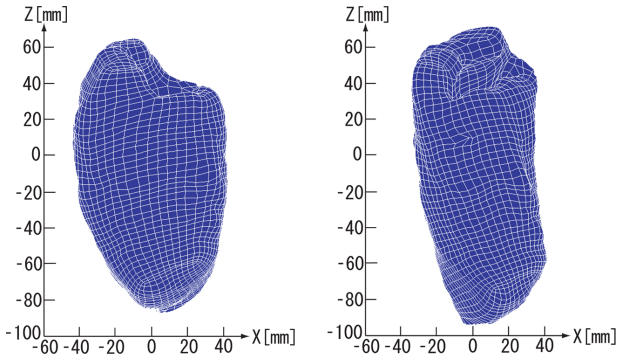
For the model without myocardial infarction, the left ventricular motion was simulated using the respective cell orientation models. The coefficient k to be used in the calculation of the contracting force was set to the value derived above, and also to 1/2 of that value.

Table 1 shows the ejection fraction for each cell orientation. Figure 12 shows the shape at end diastole and at end systole when a simple orientation model of –30° is used. We see from the result that the systolic state is dependent on the cell orientation model. The ejection fraction is large when α_{helix} is small, that is, when the contracting force is stronger in the circumferential direction. It is also seen that the ejection fraction changes when the sign of α_{helix} is reversed.

In the Huyghe model, an ejection fraction close to that of the simple orientation model of –30° was obtained. The shape data used in this experiment were complex and close to the actual shape of the heart, which produced a strong nonlinearity in the calculation of the mechanical shape deformation. Also when a different contracting force was used, the ejection fraction in the simple orientation model was larger when α_{helix} was smaller. Between the models with the same absolute value of α_{helix}, the ejection

Table 1. Ejection fraction of each cell orientation model under different contraction coefficient k

Orientation model		$k = 2.0 \times 10^{-3}$	$k = 1.0 \times 10^{-3}$
Simple model	90°	9.7	1.7
	60°	7.1	4.1
	45°	22.0	13.3
	30°	44.0	24.0
	0°	63.2	34.0
	−30°	45.8	25.5
	−45°	25.2	15.0
	−60°	13.9	7.0
Huyghe model		48.7	25.8



(a) End diastole (b) End systole

Fig. 12. Result of LV deformation with –30° simple cell orientation model.

fraction was larger in the model with a negative angle. The ejection fraction of the Huyghe model was consistently close to that of the simple –30° orientation model.

Based on these results, it is possible to evaluate the relation between the cell orientation model and the heart wall motion. It seems better to use the simple 30° orientation model and the Huyghe model when the cell orientation models are compared with the ejection fraction as the evaluation criterion. It also seems better to use the Huyghe model with a cell orientation close to that of the actual heart when discussing the effects of other component models of the simulation model, such as coronary artery model.

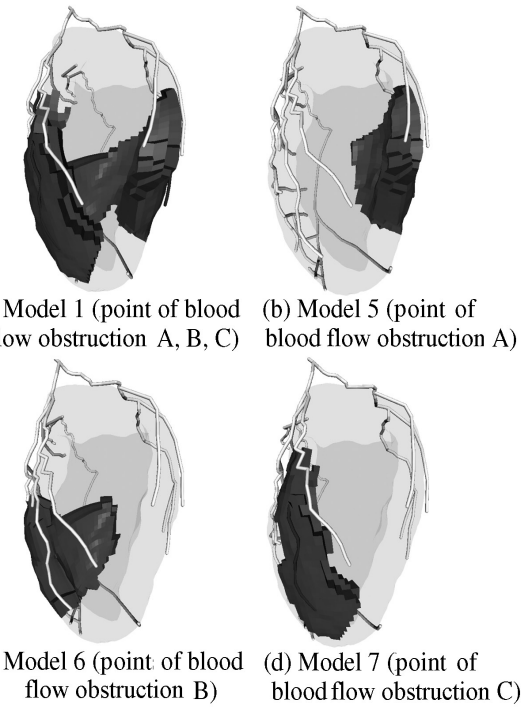
Next, the following experiment was performed to investigate the effectiveness of the interface for simulation model construction. Support for surgical planning in cases with multiple myocardial infarctions was assumed. An attempt was made to evaluate the degree of ejection fraction improvement by adjusting the point of blood flow recovery.

A point (A) in the circumflex branch and two points (B and C) in the anterior descending branch of the left coronary artery were specified as the points of blood flow obstruction. For each of the myocardial infarction areas corresponding to the points of blood flow obstruction, the ejection fraction anticipated from the blood flow recovery was examined. The ejection fraction was investigated for the following selections of recovery points.

- The original state in which the blood flow is obstructed at three points (model 1).
- The state in which the blood flow is recovered at one of three points A, B, and C (models 2, 3, and 4, respectively).
- The state in which the blood flow is recovered at two of three points (models 5, 6, and 7, respectively).
- The state in which the blood flow is recovered at all points (model 8, the same state as the normal heart).

Figure 13 shows the myocardial infarction regions in models 1, 5, 6, and 7. In this study, the volume ratio of the myocardial infarction area to the left ventricular wall is called the infarct size. Table 2 shows the infarct size and the ejection fraction in each model. The simple –30° orientation model was used as the cell orientation model. It is evident from the table that the ejection fraction decreases with increasing infarct size. It is expected that the effects of the recovery from myocardial infarction at points B and C and the recovery from myocardial infarction at point A are almost the same. Thus, the selection of surgical plans in bypass and other surgeries can be evaluated somewhat quantitatively. That is, it seems possible to utilize the model as an effective tool in the discussion of surgical plans.

The above experimental results were obtained by the preliminary simulator described in Section 4. There are still many phenomena that are not considered in the motion calculation, such as the internal pressure of the left ventricle and the cell excitation propagation model, and it is expected that the calculated shape deformation will differ greatly from the actual motion of the left ventricle. However, it will be possible to evaluate the effect of the model and its parameters related to the left ventricular motion. The inter-



(a) Model 1 (point of blood flow obstruction A, B, C) (b) Model 5 (point of blood flow obstruction A)

(c) Model 6 (point of blood flow obstruction B) (d) Model 7 (point of blood flow obstruction C)

Fig. 13. MI region of each model.

Table 2. Volume of MI and ejection fraction for the MI models

Myocardial infarction model No.	Blood flow obstruction point	Infarct size (%)	Ejection fraction (%)
1	A B C	38.3	26.0
2	A B	29.9	29.7
3	A C	24.8	33.5
4	B C	22.0	37.2
5	A	16.3	36.8
6	B	13.5	40.0
7	C	8.5	43.5
8	-	0.0	45.8

face for generating a model in which these elements can easily be modified is very important in the examination of the element models.

6. Conclusions

This study has considered the construction of a simulation model for left ventricular motion, and has presented an interface for model generation in which the element models composing the simulation model can easily be exchanged or modified. By implementing the proposed interface, it is made possible to generate a simulation model by combining arbitrary element models and parameters, and the effect of the combination on the left ventricular motion can be evaluated comprehensively.

The model for left ventricular motion is composed of the three-dimensional left ventricular model, the cell orientation model, the coronary artery model, the cell electrophysiological model, and the material property model. There has hitherto been no systematic evaluation of the combination of these proposed models, and it is very important to evaluate the models and their combinations in order to construct a precise ventricular model. By using the proposed interface, not only the proposed models, but also their combinations can easily be evaluated, which will greatly aid studies of precise simulation model construction for left ventricular motion.

For the motion model constructed using the proposed interface, the motion was simulated by a preliminary simulator. It was verified that the cell orientation model and its parameters had a large effect on the left ventricular motion. It was also verified that when myocardial infarction is produced, the ejection fraction decreases with the increase of its volume. It is thus shown that the implementation of the interface for model generation is important in the construction of biological models with complex structure, as in the case of the heart.

The motion model and the simulator which have been implemented at the present stage do not yet include various phenomena, such as the fluid dynamics of the blood and the membrane excitation propagation model, which are expected to have large effects on the left ventricular motion. It is planned in the future to expand the interface so that these models can be handled.

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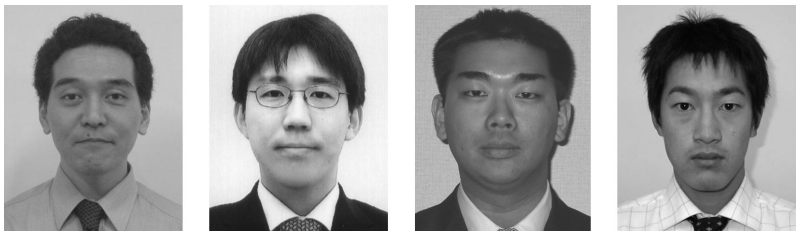

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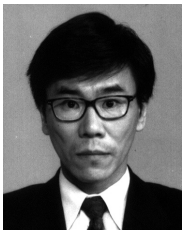

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