The looped heart does not save energy by maintaining the momentum of blood flowing in the ventricle

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Watanabe H, Sugiura S, Hisada T. The looped heart does not save energy by maintaining the momentum of blood flowing in the ventricle. Am J Physiol Heart Circ Physiol 294: H2191–H2196, 2008. First published March 7, 2008; doi:10.1152/ajpheart.00041.2008.—Previous studies suggested that the reconstruction or maintenance of physiological blood flow paths in the heart is important to obtain a good outcome following cardiac surgery, but this concept has no established theoretical foundation. We developed a multiscale, multiphysics heart simulator, based on the finite element method, and compared the hemodynamics of ventricles with physiological and nonphysiological flow paths. We found that the physiological flow path did not have an energy-saving effect but facilitated the separation of the outflow and inflow paths, so avoiding any mixing of the blood. The work performed by the ventricular wall was comparable at slower and faster heart rates (physiological vs. nonphysiological, 0.864 vs. 0.874 J; heart rate = 60 beats/min; and 0.599 vs. 0.590 J, heart rate = 100 beats/min), indicating that chiral asymmetry of the flow paths in the mammalian heart has minimal functional merit. At lower heart rates, the blood coming in the first beat was cleared almost completely by the ninth beat in both models. However, at high heart rates, such rates, the blood coming in the first beat was cleared almost completely by the ninth beat in both models. Models, however, do not exist for the heart rate. Stress and heart rate, such complete clearance was observed only in the physiological model, whereas 27.0% of blood remained in the nonphysiological model. This multiscale heart simulator provided detailed information on the cardiac mechanics and flow dynamics and could be a useful tool in cardiac physiology.

computer modeling; blood flow; hemodynamics; biomechanics

DURING EMBRYOGENESIS, the heart develops from the straight heart tube and undergoes a morphogenetic process referred to as cardiac looping (20). As a consequence, the blood flow paths show chiral asymmetry in the adult mammalian heart, with asymmetric redirection of the stream in the atria and ventricles (13). Because of its unique characteristics, the flow pattern in the left ventricle has been the subject of computer modeling (3, 14, 15) and of measurements by various means (13, 19, 27). The prevailing view emerging from these studies is that the asymmetric redirection has a functional advantage because it can avoid collision of the flows. Furthermore, the rotational redirection of the inflow from the atrium, often referred to as the vortex formation, is thought to support the continuous motion of blood and thereby minimize loss of momentum, such that a contraction of the ventricle only serves to add extra energy for ejection of the blood already moving toward the aortic valve (13, 31). If it exists, such an energy-saving effect of flow redirection would be of paramount importance for not only the basic physiology of the heart but also the clinical cardiology, since aberrant flow patterns are observed in congenital heart diseases (24) or after mitral valve replacement (19). However, because of technical difficulties, none of the experimental or simulation studies mentioned above has provided concrete evidence to support this concept. Alterations of the flow path can be achieved surgically (17, 19), but precise intraventricular mapping of the resultant flow and pressure, which are prerequisites for the estimation of the momentum and energy balance, are not available using current experimental techniques. Similarly, simulation studies that can realistically reproduce the natural blood flow and behavior of the ventricle by integrating the molecular mechanisms of excitation-contraction coupling, mechanics, fluid mechanics, and cardiac structure in three-dimensional space (12) have not been realized to date.

We have developed a finite element method (FEM), based on a model of the left ventricle, in which the molecular mechanism of excitation-contraction coupling implemented in each element is activated by the propagation of excitation (29). In this simulation, the equations governing the dynamics of the fluid (blood) and the structure (ventricular wall) are solved by a strong coupling method (33), and we can therefore obtain detailed information regarding the intraventricular flow velocity and pressure, as well as the stress-strain relationships in the ventricular wall, which are necessary for energetic considerations. Taking advantage of this model, we have examined the effects of the inflow direction in the left ventricle on its performance under conditions where other experimental variables, including ventricular morphology, myocardial properties, and pre- and afterloads, are completely maintained. Such a controlled study is only possible with this kind of in silico experiment.

METHODS

Details of the formulations adopted for the model have been described in our previous report (29). Here we briefly present the framework of the model. For the details of formulation, see supplementary data (note: all supplemental material cited in this article may be found with the online version of this article).

FEM of the left ventricle. The present simulation is based on the FEM (2). The model of the left ventricle consists of 9,792 elements divided into six layers with distinct fiber directions ranging from −60° to +60° from the endocardium to the epicardium. In response to electrical stimulation applied to the endocardium, the excitation propagates toward the epicardium to induce the contraction/relaxation cycle of each element, representing the myocytes. In this simulation, the FitzHugh-Nagumo (FHN) model (7, 21) was coupled with the monomodern propagation model (7, 11, 21) to reproduce the excitat
are constant. Because we did not model the conduction system, we applied the stimulation signal to all the elements on the endocardial surface to initiate the excitation.

Fluid-structure interaction analysis and study protocol. The blood in the ventricle (fluid part) was also modeled by FEM with 18,976 elements, and its interaction with the ventricular wall (structure part) was analyzed by the fluid-structure interaction FEM, which has been developed by us (29, 33). For the analysis of blood flow, the arbitrary Lagrangian-Eulerian form of the Navier-Stokes equations was discretized by a standard FEM. For the analysis of ventricular wall motion, the total Lagrangian formulation was discretized by an FEM, and a nonlinear system of equations was obtained. These equations were solved by a strong coupling method. For the determination of the intraventricular flow pattern, the modeling of hemodynamic pre- and afterloads was needed, in addition to the interactions between the blood and wall properties. To simulate the systemic arterial tree, the windkessel model (30) was connected to the aortic valve. An electric analog model of left atrium and pulmonary circulation, proposed by Alexander and colleagues (1, 29), was modified and connected to the mitral valve to simulate the preload of the left ventricle (Fig. 1).

To study the impact of cardiac looping, the hemodynamic parameters of two model ventricles with different inflow angles were compared. First, with the physiological model, the inflow was tilted to the lateral wall by 10° from the long axis of the ventricle (light gray arrow in Fig. 1). This configuration was determined according to the magnetic resonance imaging data of Kilner et al. (13). Second, with the nonphysiological model, the inflow was tilted to the septum by 45° from the long axis (dark gray arrow in Fig. 1). A comparison was made at low (60 beats/min) and high (100 beats/min) heart rates.

RESULTS

In the physiological model, blood coming in from the atrium first proceeded to the apex and then turned toward the aortic valve to form a clockwise vortex (Fig. 2A, and supplemental movie 1), thus reproducing previous experimental and clinical observations (13, 17, 27, 32). Since the inflow angle deviated from the physiological angle, the intraventricular flow path changed drastically and a counterclockwise vortex finally appeared (Fig. 2B, and supplemental movie 2). However, despite such remarkable variations in the flow direction, the performance of the heart was minimally affected. As shown in Fig. 2C, the pressure-volume trajectories of the ventricle simulated with different inflow angles did not differ appreciably in terms of their height (developed pressure) or width (ejected blood volume).

We next evaluated the momentum of the blood at the aortic valve (red loop A in Fig. 2A) and beneath the outflow tract (red loop B in Fig. 2B). To evaluate the effect on ejection, the
velocity component normal to the loop plane was quantified and the upward direction was defined as positive. At a heart rate of 60 beats/min, the blood momentum measured at the outflow (loop B) differed slightly between the physiological model (black solid line) and nonphysiological model (red dashed line), but this difference disappeared at the aortic valve (loop A; physiological model, blue solid line; and nonphysiological model, green dashed line; Fig. 3A). Since Kilner et al. (13) previously suggested that the heart used the momentum of blood efficiently during exercise, we also simulated such a condition by increasing the heart rate (100 beats/min). In this case, the difference in the outflow became significant, but again it disappeared at the aortic valve (Fig. 3B). These results can be taken to indicate that the momentum of the blood heading for the outflow tract was negligible compared with that in the aorta and that the kinetic energy of the ejected blood was mainly provided by the work performed by the left ventricular myocardium, and, when we calculated the work done by the myocardium, the two types of ventricles produced similar amounts of external work (Table 1).

Therefore, what is the functional significance of cardiac looping? A possible consequence of the asymmetric flow redirection is the separation of the outflow from the inflow to avoid the mixing of blood. To evaluate the dynamics of the blood clearance by each cardiac cycle, we calculated the amount of external work (Table 1). The introduction of novel imaging techniques to cardiovascular science has revealed chirally asymmetric flow paths in the heart, produced by cardiac looping (13, 27). In particular, previous observations of flow redirection in the ventricle seemed to support the concept of an energy-saving, fluidic effect of cardiac looping (13, 31). The current simulation study contradicts this view, and clinical observations support our results. Maire et al. (19) compared two groups of patients with either physiological (parallel flow; clockwise vortex) or nonphysiological (cross flow; counterclockwise vortex) ventricular flow patterns induced by differences in the orientation of prosthetic mitral valves and found no differences between the morphologies, hemodynamics, or exercise capacities, except for a larger atrial size in the nonphysiological group. Since that study was performed more than five years after surgery in each patient, we expect that even a small functional difference, if it existed, would have accumulated and brought about deleterious clinical consequences. The absence of the energy-saving effect, even at the higher heart rate, was also consistent with earlier studies, showing that myocardial mechanoenergetics were insensitive to heart rate (6, 9).

Although our simulation could reproduce the flow dynamics during physiological contraction and relaxation of the ventricle based on the propagation of excitation, changes in the tissue properties and fluid-structure interaction analysis (16, 29), substitution of the left atrium with an electrical circuit analog, eliminated the analysis of the atrioventricular interaction. It is believed that a recoil of the ventricle away from the ejecting blood adds to the ventricular contraction and expands the atrium to enhance its filling during ventricular systole. In turn, the enhanced filling increases the momentum of the blood during the successive diastole, contributing to the slinging blood action. Although such an interpretation is correct overall, we remain skeptical about the significance

![A](image1.png) ![B](image2.png)

**Fig. 3.** Time and location dependence of the blood momentum. The momentum of the blood at the aortic valve (red loop A in Fig. 2A) and beneath the outflow tract (red loop B in Fig. 2B) were compared. The blood momentum measured at the outflow (loop B) for physiological model (black solid line) and nonphysiological model (red dashed line). At the aortic valve (loop A), physiological model, blue solid line; nonphysiological model, green dashed line. A: HR = 60 beats/min. B: HR = 100 beats/min.

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<th>Heart rate</th>
<th>Internal Work, J</th>
<th>External Work, J</th>
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<tr>
<td>60 beats/min</td>
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<td>0.864</td>
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<tr>
<td>100 beats/min</td>
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<td>0.599</td>
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**Table 1. Work performed by the ventricle wall**

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of the contribution of the fluidic action to this process. First, the reaction force of the ejecting blood is small compared with that generated by the muscle contraction. Second, since most of the filling takes place when blood flows into the ventricle directly from the pulmonary circulatory bed (8, 22) during the early rapid filling phase, an expansion of the atrium may not contribute greatly to the overall pumping function.

Fig. 4. Clearance of blood from the ventricle. The amount of remaining blood was defined as (the number of remaining particles in the nth beat)/(total number of inflow particles in the first beat). Solid line, the physiological ventricle; broken line, nonphysiological ventricle. A: HR = 60 beats/min. B: HR = 100 beats/min.

Fig. 5. Inflow blood motion. The distribution of blood coming into the ventricle in a specific beat during the following cardiac cycle. Activation level of the myocardium (force normalized by its maximum level) is shown by color coding. A: physiological model. B: nonphysiological model. A small but significant fraction of the blood remains in the inflow (right) side.
What, then, is the functional significance of cardiac looping? This detailed observation of the flow patterns may provide mechanistic information. In the physiological ventricle, the blood coming in as a jet moved along the ventricular wall, spread out gradually, and pushed the already-present blood toward the outflow side (Fig. 5A, and supplemental movie 3). On the other hand, in the nonphysiological model, a counterclockwise vortex developed just below the aortic valve and most of the swirling blood was ejected in the subsequent contraction, though a small fraction remained stagnant and was excluded from the circulation (Fig. 5B, and supplemental movie 4). Since the inflow and outflow blood in the left ventricle are both oxygenated, the first-in-first-out mode of function cannot have physiological importance, as is the case in the amphibian ventricle (26), whereas the blood stagnation that became apparent at higher heart rates may lead to an unfavorable outcome.

We must consider the possibility that several important properties of the heart, which were not modeled in the current study, may have significant effects. First, although the fiber orientation was arranged according to the histological data, the ventricular shape was modeled with simple symmetrical geometry. However, because many studies have shown that the ventricular shape can be approximated by a rotational ellipsoid (4, 10, 28), the effect of such simplification would be minimal. Of course, the influence of the valves and trabeculae need to be taken into consideration in future studies. Second, because we did not model the conduction system, we applied the stimulation signal to all the elements on the endocardial surface simultaneously to initiate the excitation. This is obviously not a physiological condition, but, according to the isochronal map of ventricular activation reported by Durrer et al. (5), the physiological activation propagates from the endocardium to the epicardium in an eccentric manner within 40 ms, except for the posterobasal portion and, therefore, does not differ significantly from the situation reproduced in this simulation. Finally, we adopted the constitutive equation for myocardium derived from the animal experiment by Lin and Yin (16), because such a detailed analysis of material property is not available for the human heart. Overall, we consider that the simplicity of this modeling did not influence our conclusion significantly, but further improvement in the model with respect to realistic shape, conduction system and valves, as well as a more detailed model of excitation-contraction coupling is needed to substantiate the current findings.

In conclusion, instead of working as a mixing chamber, the human left ventricle, by retaining a looping structure, functions in a first-in-first-out mode, but the energy-saving effect may not be significant. The present results have implications for cardiac surgery aimed at correcting the intracardiac flow (17, 18), as well as for studies on cardiac morphogenesis. Multiscale biological modeling connecting the cellular mechanisms of cardiac contraction to flow dynamics, combined with the wide variety of bioimaging modalities, could be a powerful tool for in silico biology (23).

**GRANTS**

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**REFERENCES**