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Key Words: left ventricle assistant device, LVAD anastomosis geometry, aortic insufficiency, computational fluid dynamics, wall shear stress at aortic root

EDITORIAL COMMENTARY

Computational fluid dynamics: Solidifying fluid concepts in left ventricular assist device therapy

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See related article on pages 696-704.

The opinions that are held with passion are always those for which no good ground exists; indeed the passion is the measure of the holders lack of rational conviction.

—Bertrand Russell, *Sceptical Essays*, 1928

The study of viscous fluids, predicting their behavior under the influence of outside forces and constraints, is of ever-increasing importance for the everyday cardiac surgeon, and is especially important to the heart failure surgeon. To understand the short- and long-term consequences of

differing device designs and anastomotic factors (angle, length, location in relation to structural components), and the impact that conduit selection, design, size-matching, and tailoring have on outcomes after cardiac surgery, requires an appreciation for the parameters that influence blood flow.

The first description of the physiology of the circulatory system, where the blood is described as a fluid pumped by the heart, was made by William Harvey in 1628 in the *Exercitatio Anatomica de Motu Cordis et Sanguinis in Animalibus*. Theoretical advances in our understanding of blood flow, and the ability to predict the impact of various anatomic defects or structural lesions on blood flow, would have to wait for developments in the fundamental understanding of fluid dynamics as ultimately described by the Navier-Stokes equations.

The Navier-Stokes equations are mathematical relationships describing the behavior of viscous fluids, applying the general principles of the conservation of mass, energy, and momentum within the context of Newtonian dynamics. This set of nonlinear partial differential equations describes a velocity field in a given volume of space. The velocities at various points can be utilized to calculate other useful characteristics of the fluid, such as pressure, stress, or shear. Given that they are nonlinear equations, the solutions are cumbersome to derive, and whether

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FIGURE 1. Computational fluid dynamic map of a novel, continuous flow, total artificial heart. (The assistance of Dr William E. Cohn and Dr Daniel Timms, in sharing this image, is appreciated.)

solutions exist, mathematically, for all sets of these equations in 3-dimensional (3-D) space, is unclear. The utility of these equations in modeling the behavior of fluids did not become a reality until the advent of powerful computational devices.

Francis H. Harlow and colleagues (Group T-3) at the Los Alamos National Laboratory were the first to use a computational methodology to compute the 3-D behavior of fluids governed by the Navier-Stokes equations. This group established the scientific framework for the nascent field of computational fluid dynamics (CFD). Understanding the steps involved in performing CFD analysis allows the cardiac surgeon to better appreciate the message from these studies.

All CFD research starts with an anatomic model to define the geometric constraints on which the equations are expected to function. Next, the volume occupied by the fluid is divided into smaller sections, called the mesh. The characteristics of the mesh can be defined and can be uniform throughout the volume, or nonuniform. Following this step, the behaviors of the fluid at the boundary of the anatomic model are predefined (boundary conditions), and the equations governing the fluid within the model are specified. Generally, these equations incorporate Newtonian concepts of mass and energy. Finally, the equations are solved in an iterative process using high-level

computing power, and the results are processed into the nice color maps typical of this kind of investigation (Figure 1).

In the current issue, Callington and colleagues¹ systematically assess varying left ventricular assist device (LVAD) outflow graft anastomotic locations, with respect to the sinotubular junction, and angulation, with respect to the coronal and transverse planes of the aorta. These are used to predict pressure and wall shear stress (WSS) in the aortic root, utilizing 3-D reconstructed imaging from computed tomography angiography of an LVAD patient to construct their anatomic model. They demonstrate that the location and angle of the LVAD outflow aortic anastomosis has a dramatic impact on the aortic pressure and WSS, and that these effects are different in the aortic root versus the aortic arch or descending aorta. Intermittent aortic valve opening and flow through the aortic valve had a minimal effect on pressure and WSS.

If abnormal aortic root pressure and WSS contribute to the occurrence of late aortic valve insufficiency after LVAD implantation, then this study indicates that a low anastomotic position, with an inclination angle of 90° and an azimuthal angle of 60°, will have the most physiologic aortic root parameters. However, this configuration is shown to have decreased pressure and flow in the aortic arch and descending aorta. The continuous nonpulsatile

flow in the carotid bulb may be associated with the formation of thrombus and could lead to stroke in patients on long-term LVAD support.² The effect of varying anastomotic configurations on flow patterns in the main branch vessels that arise from the aortic arch will have to be studied to determine which location and angulation are likely to result in the best overall circulatory support for the patient.^{3,4} Ultimately, studies of this nature permit the surgeon to plan—based on preoperative 3-D computed tomography reconstructions of the heart, mediastinum, and aorta—for the optimal inlet location, pathway for the outflow graft, and outflow graft anastomosis site and orientation. Individualizing LVAD therapy, to account for patient-specific factors (both structural and

functional) should allow surgeons to obtain the best possible outcomes.

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