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# Dynamic modelling of prosthetic chorded mitral valves using the immersed boundary method

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**Abstract**

Current artificial heart valves either have limited lifespan or require the recipient to be on permanent anticoagulation therapy. In this paper, effort is made to assess a newly developed bileaflet valve prosthesis made of synthetic flexible leaflet materials, whose geometry and material properties are based on those of the native mitral valve, with a view to providing superior options for mitral valve replacement.

Computational analysis is employed to evaluate the geometric and material design of the valve, by investigation of its mechanical behaviour and unsteady flow characteristics. The immersed boundary (IB) method is used for the dynamic modelling of the large deformation of the valve leaflets and the fluid–structure interactions. The IB simulation is first validated for the aortic prosthesis subjected to a hydrostatic loading. The predicted displacement fields by IB are compared with those obtained using ANSYS, as well as with experimental measurements. Good quantitative agreement is obtained. Moreover, known failure regions of aortic prostheses are identified. The dynamic behaviour of the valve designs is then simulated under four physiological pulsatile flows. Experimental pressure gradients for opening and closure of the valves are in good agreement with IB predictions for all flow rates for both aortic and mitral designs. Importantly, the simulations predicted improved physiological haemodynamics for the novel mitral design. Limitation of the current IB model is also discussed. We conclude that the IB model can be developed to be an extremely effective dynamic simulation tool to aid prosthesis design.


*Keywords:* Immersed boundary; Chordae; Mitral valve; Aortic valve; Prosthetic heart valve; Static and dynamic simulations; Fluid–structure interactions

## 1. Introduction

Currently, biological (human cadaver, bovine pericardial or porcine tissues) or mechanical prosthetic valves are used for mitral valve replacement. Biological prostheses have excellent mechanical and haemodynamic properties, however they have a limited lifespan due to tissue failure and calcification. Mechanical prostheses are more durable, but are unable to replicate physiological flow conditions. These introduce high velocity jets into the blood flow, which results in elevated shear stresses, promoting platelet aggregation (King et al, 1996). Consequently, lifelong anticoagulation therapy is required to prevent thrombosis and thromboembolism. In addition, neither type of valve is specifically designed for the mitral position: these are designed for the aortic position and reversed for the mitral position, thus they do not function in a similar haemodynamic way to the native mitral valve. More importantly, they do not retain the function of the chordae, which play a significant role in the preservation of left ventricular function.

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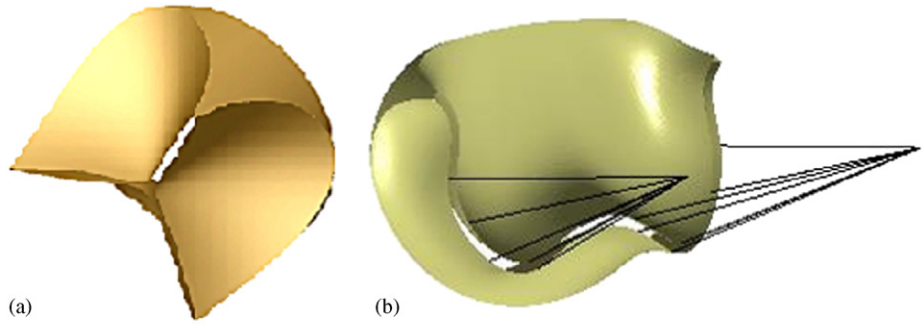


Fig. 1. The Glasgow designed aortic (a) and chorded mitral (b) prostheses.

In this paper, a new mitral polyurethane bileaflet valve prosthesis is evaluated. It's geometry and mechanical properties are designed, to a large extent, to mimic the native mitral valve (Wheatley, 2002), see Fig. 1b. It is designed to combine the advantages of mechanical and bioprosthetic heart valves, i.e. long-term durability without the need for permanent anti-coagulation. The valve has a larger anterior leaflet and incorporates chordae, which originate from the valve annulus and traverse the leaflet, exiting at the leaflet edges to attach to the papillary muscle regions of the ventricle. The chordae act to reinforce the leaflet structure and prevent prolapse of the leaflets when the valve closes. They also help to maintain the geometry and functionality of the ventricle.

Experimental analysis, mechanical insight and experience, are integral to the development of a good valve design. Once the basic design is formulated, computational analysis can prove extremely useful: it may highlight weaknesses in the design that are not intuitive, such as locations subject to critically high stresses; material and geometric parameters may easily be varied to obtain an optimum design.

In the past, we have developed trileaflet polymer valves based on the native aortic valve design (Bernacca et al., 1995; Mackay et al., 1996a, b; Bernacca et al., 1996, 1997a, b, 1999, 2002; Wheatley et al., 2000). These designs showed potential, but some issues remained in balancing the conflicting requirements of good hydrodynamic function and excellent durability. A material that is sufficiently flexible to provide competitive hydrodynamic function (Bernacca et al., 2004) may not have the required durability due to its low modulus. A higher modulus material may be durable but may require manipulation of the design or the leaflet thickness distribution to provide competitive hydrodynamics. Optimisation of hydrodynamic function would reduce the incidence of thrombogenesis, minimising the likelihood of failure associated with leaflet stenosis due to thrombus deposition and/or calcification.

To effectively assist prosthesis design we need to consider both a static analysis, to study the mechanical behaviour of valves under peak loading conditions (systolic pressure), and a dynamic analysis with large deformation fluid–structure interactions, to study the properties of the flow though the valve, and the dynamic behaviour of the leaflets. Although there are a limited amount of static and quasi-static simulations on mitral valves (Kunzelman and Cocharan, 1993; Kunzelman et al., 1996), to the best of our knowledge, to date no work has been carried out for simulating chorded mitral valves in the fully fluid–structure interactive opening and closing phases. Many currently available commercial packages, e.g. ANSYS, CFX, Fluent, all have difficulty modelling the strongly coupled fluid–structure interaction (FSI) of heart valve prostheses due to the inherent large deformations and severe distortions to the fluid mesh.

In this paper, we use the immersed boundary (IB) method to simulate the FSI problem in which an elastic material interacts with a viscous incompressible fluid (Peskin, 2002). IB's effective dynamic capability has been used to model blood flow patterns in the heart (McQueen and Peskin, 1997; Peskin, 1972, 1977; Peskin and McQueen, 1995, 1996), platelet aggregation during blood clotting (Fogelson, 1984), flow and transport in a renal arteriole (Arthurs et al., 1998), wave propagation in the cochlea (Beyer, 1992), aquatic animal locomotion (Fauci and Peskin, 1988), wood pulp fibres dynamics (Stockie & Green, 1998), and to assist prosthetic valve design (McQueen and Peskin, 1983, 1985). However, it was not initially designed to consider hydrostatic loading and modelling of arbitrary geometries. This is because the IB models needed to be created from a network of interconnecting fibres whose geometries and stiffness are defined to mechanically represent the structure to be modelled. The geometry of the fibres was generally defined by equations, an impractical approach if arbitrary and complicated valve geometries are to be constructed.

We have extended the current IB method so that it can effectively model arbitrary geometries, as well as be subjected to static loading. The ability to model static loading is important, since comparisons and validation