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Finite Element Modeling of Mitral Valve Repair

The mitral valve is a complex structure regulating forward flow of blood between the left atrium and left ventricle (LV). Multiple disease processes can affect its proper function, and when these diseases cause severe mitral regurgitation (MR), optimal treatment is repair of the native valve. The mitral valve (MV) is a dynamic structure with multiple components that have complex interactions. Computational modeling through finite element (FE) analysis is a valuable tool to delineate the biomechanical properties of the mitral valve and understand its diseases and their repairs. In this review, we present an overview of relevant mitral valve diseases, and describe the evolution of FE models of surgical valve repair techniques. [DOI: 10.1115/1.4032125]

Keywords: finite element modeling, mitral valve repair, cardiac surgery

Introduction

The mitral valve is a complex structure regulating forward flow of blood between the left atrium and LV. The valve apparatus consists of the mitral annulus, a saddle-shaped ring of thickened connective tissue, which supports the two leaflets of the valve, anterior and posterior. These leaflets are anchored by a complex web of branching fibrous chords, the chordae tendinae, to two muscular promontories of the left ventricular wall, the anterior and posterior papillary muscles (PPM). These papillary muscles are extensions of and supported by the muscle of the ventricular wall (Fig. 1).

Pathophysiology

A number of processes affect the proper function of the mitral valve, leading to regurgitant flow of blood from the LV into the left atrium; this condition is known as MR. MR can be broken down into two overall categories. Disease processes intrinsic to the valve affect its function by altering the leaflets, chordae, and mitral annulus. This is known as primary MR. Processes extrinsic to the valve which cause deformation of the LV itself lead to altered mitral geometry and are referred to as secondary or functional MR [1].

In primary MR (Fig. 2(a)), one or more components of the mitral valve are diseased, allowing regurgitant flow of blood. The most common cause in developed countries is mitral valve prolapse. In young people, this is typically caused by a severe form of valvular degeneration known as Barlow disease, affecting the anterior and posterior leaflets and chordal tissues, but can also be a result of various congenital connective tissue disorders such as Marfan syndrome, and diverse other causes such as infection and

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Fig. 1 (a) Intraoperative image of floppy valve with “flail” posterior segment: AL = anterior leaflet, PL = posterior leaflet. **(b)** Intraoperative image of subvalvular apparatus demonstrating chordae and papillary muscles. **(c)** Intraoperative image of repaired and reshaped mitral valve.

trauma. Older populations generally present with milder fibroelastic deficiency, where lack of connective tissue leads to rupture of one or more chordae tendinae and isolated prolapse of a single area of the valve, most commonly the center of the posterior leaflet. Because these disease processes are intrinsic to the valve, surgery tends to be curative [1–4].

Secondary MR (Fig. 2(b)) occurs in the presence of a normal mitral valve, and is caused by underlying changes in the LV. These changes may be secondary to coronary artery disease, history of myocardial infarction, or other myocardial disease that leads to dilation of the LV. LV dilation displaces the papillary muscles and chordae tendinae, causing dilatation of the mitral annulus and tethering of the leaflets. This leads to restricted leaflet motion, poor leaflet coaptation, and regurgitant blood flow. Because MR is only one component of the disease process, treatment must address the causative cardiac disease as well as the valvular disorder. Secondary MR is therefore more difficult to treat surgically and recurs more often after surgery; patient outcomes are directly related to the severity of their underlying heart disease [1,5].

Mitral Valve Repair

Surgical repair is part of the treatment for most causes of MR, and is the definitive treatment for primary MR [1]. It is recommended for all symptomatic patients with primary MR, and in asymptomatic patients with significant regurgitation who are good operative candidates. Effective surgical interventions include valve replacement and valve repair; repair has better long-term outcomes, including improved patient survival, but is more difficult to perform than replacement, partially because there is no standard method of repair [6]. The spectrum of currently performed valve repairs ranges from simple reinforcement of the mitral annulus with a rigid or semi-rigid prosthetic band (mitral

annuloplasty (MA)), to resection of the prolapsing leaflet with or without annuloplasty, to recreation of disrupted chordae tendinae using gore-tex suture [7]. For secondary MR, the most common repair is placement of an annuloplasty ring. While this corrects the valvular dysfunction initially, it fails to address the underlying disease processes, and continued remodeling of the LV results in high rates of repair failure over time [8–10]. Additional repair types have therefore been proposed which seek to correct the ventricular dilatation, displacement of the papillary muscles and the outward traction on the mitral annulus and valve leaflets [11–13]. For patients who are poor surgical candidates, percutaneous valve repair approaches have been developed, which mimic open surgical procedures [14].

Across the spectrum of valve repair, durability is an issue. Reoperation rates for recurrent MR after an initial repair reach 20% at 20 years. We hypothesize that this recurrence rate is due to a combination of initial post-operative valve geometry, and remodeling over time secondary to stress on the valve and ventricle. To understand the geometry of each type of repair, as well as the alterations in stress and strain across the valve leaflets, papillary muscles and LV, sophisticated models are necessary, incorporating the mitral valve, the subvalvular structures, and the LV.

Understanding Stress in Cardiac Tissues

The myocardium is a mechanically complex, nonhomogeneous material, with mechanical properties that vary during the cardiac cycle and evolve over the life of the organism. Understanding stress distributions in the myocardium and across the cardiac valves is clinically important because stress influences regional coronary blood flow, myocardial oxygen consumption, hypertrophy, and remodeling [15,16]. Directly measuring stress across intact heart valves and myocardium in vivo has been difficult, secondary to tissue damage caused by implanted transducers and the

Fig. 2 (a) 2D echocardiogram of MR secondary to isolated prolapse of the center segment of the posterior leaflet (P2), with color image showing regurgitant blood flow. (b) Echocardiogram of patient with functional MR, showing normal valve leaflets and regurgitant flow of blood. (c) Mitral valve after surgical repair, showing normal leaflet coaptation and no regurgitant flow of blood.

large biologic deformations of the tissues involved [17,18]. Consequently, computational models of the heart have been developed to explore these constitutive properties. The last 30 years have seen an exponential increase in model complexity: Initial models, applying LaPlace's law to the ventricle as a whole, led to early computational models which assumed that the myocardial stress-strain relationship is linear, isotropic, and only undergoes small deformations. From there, current techniques sprang up that incorporated directionality of myocardial muscle fibers and the large biologic deformations of cardiac tissues.

LaPlace's Law. Wall stress in the LV and cardiac valves has traditionally been estimated with LaPlace's law. This is based on the assumption of a thin and uniform ventricular wall, and states that stress is directly proportional to the ventricular cavity pressure and radius of curvature of the endocardium, and inversely

proportional to the ventricular wall thickness. It can only be used to calculate average stress across a material. Under conditions such as myocardial infarction or valvular disease, cardiac tissues undergo regional changes in systolic and diastolic material properties, and the ventricle and valves become mechanically complex structures [19]. The use of LaPlace's law is, therefore, severely limited.

FE Simulations. An alternative approach to quantifying cardiac stress and stiffness is mathematical modeling based on the conservation laws of continuum mechanics. To solve the governing equations of equilibrium for a pressure vessel with complex geometry, boundary conditions, and material properties, computational techniques are required. The most versatile technique for analyzing cardiac mechanics is the FE method [20]. This method is capable of predicting regional stress, regional deformation, and quantifying stroke volume.

Modeling the Isolated Mitral Valve

Early FE models of the mitral valve assume the papillary muscles are fixed in space, and examine the valve in isolation. For example, Kunzelman et al. developed a sophisticated static and quasi-static FE model of the isolated mitral valve with geometry that was derived from serial sectioning of resin filled casts [21]. This model includes thickness differences in the valve leaflets from annulus to free-edge. A small-angle-light-scattering mapping of local mean collagen fiber orientation together with site specific digests of noncollagen proteins, permitted site variable stiffness data to be deduced and verified with uniaxial stiffness measurements [22]. Tissue descriptions were anisotropic, but linear, with stiffness values taken to be post-transition values from uniaxial tests [23,24]. Fluid effects were not directly considered but were accounted for by increasing valvular inertia.

Further work to define the valve leaflet material properties was performed by Krishnamurthy, et al., who developed an ovine FE model of the anterior leaflet [25]. Building on previous in vivo studies, they implanted radiopaque markers on the mitral annulus, papillary muscle tips and anterior leaflets of 17 sheep to test the hypothesis that valve leaflets, which appear to be extremely compliant in vitro, are significantly stiffer in vivo, potentially due to the presence of neurally controlled contractile tissues within the leaflet. They obtained videofluoroscopic studies of the beating ovine heart after marker implantation, and based their analysis on images of three consecutive cardiac cycles in sinus rhythm. From these marker positions they developed a mesh and assigned material properties based on prior histologic studies of leaflet tissue thickness. Their results indicate that leaflet tissue in vivo is significantly anisotropic, with much greater stiffness in the circumferential as opposed to the radial direction. Their results also demonstrated much stiffer leaflet tissue in vivo than shown previously in studies of excised tissue, supporting their hypothesis. More recent efforts have expanded to include the subvalvular structures, examining the effects of isolated changes in annular and papillary muscle geometry in explanted ovine hearts, further establishing the experimental foundation necessary for accurate in silico modeling of the entire mitral apparatus [26,27].

The natural extension of these techniques was the first model of the isolated human mitral valve based on patient imaging data, developed by Choi et al. [28]. Patient-based FE models of the isolated mitral valve have now been used to determine the effect of artificial chordae on valve deformation, the effects of leaflet and annular curvature on leaflet stress, and the effects of innovative annuloplasty ring shape [29–32].

Evolving in parallel with these patient-imaging-based models, recent studies of the isolated, excised mitral valve are reaching new levels of complexity. In 2015, Toma et al., published results of a fluid-structure interaction model of papillary muscle forces, structurally based on MicroCT of excised porcine mitral valves. Their methods incorporate the specific architecture of the subvalvular apparatus, including the entire 3D network of branching

chordae tendinae and papillary muscles, and utilize previous work on collagen–fiber orientation described above, to give microscopic-level detail of the forces acting on individual papillary muscle tips and chordae [33].

As medical imaging continues to advance in resolution and computing power increases, we anticipate the union of these complex ex vivo models with patient-specific imaging data, providing new insight into the interactions of the mitral valve and subvalvular apparatus. Single patient results have been reported for multi-slice CT as a potential method to accomplish this union [34].

Modeling the Isolated LV

To perform FE analysis of the myocardium, the material stress–strain relationship must be specified. Many different constitutive relationships have been formulated for cardiac tissues, and we employ the FE method developed by Drs. Hunter, McCulloch, Guccione and colleagues specifically for continuum analysis of the heart (CMISS/Continuity). It includes several features that are uncommon in conventional FE methods, including allowing modeling of the ventricle as nonhomogeneous and anisotropic. The models are based directly on 3D fiber angle and sheet angle distributions (measured with a variety of techniques) and have been validated experimentally under a wide range of conditions [35–37].

Modeling the Mitral Valve + LV

Models of the isolated mitral valve developed our understanding of valve geometry and stress/strain relationships, and can usefully model isolated valvular repair. However, many of the processes affecting mitral valve function involve not just the valve, but also pathologic changes in the LV. For example, models of functional MR require incorporation of the dilated left ventricular wall and outwardly displaced papillary muscles. Various surgical repairs alter geometry of the papillary muscles, chordae tendinae, and wall of the LV. Even repairs that may at first glance appear to be isolated to the valve have broad-ranging effects on the ventricle as well, as the valve is a crucial component of normal ventricular function [38,39]. To understand these repairs, and the complex interrelationship of stress on the valve and ventricle, a more complete model of the mitral valve and LV in continuity is necessary.

Ovine Models. The first combined FE model of the MV and LV was developed in 2010 by Wenk et al. [40] (Fig. 3). High-resolution MRI and pressure transducing catheters in the LV were used to collect data on a sheep with ischemic MR secondary to induced myocardial infarction. The MRI images were contoured along the endocardial and epicardial surfaces to construct a model of the LV. The mitral valve annulus and leaflets were identified and contoured similarly, and the annulus was constructed from 12 points measured around the perimeter of the mitral valve. The chordae tendinae were not visible on the MRI images, and were therefore approximated based on published anatomic data from an excised heart, which included the chordae and papillary muscles

Fig. 3 Wenk model of LV + MV [40]. (a) Exterior view of the LV showing postero-basal MI. (b) Interior view of LV with chordae connected to the papillary muscles.

[41]. The acquired LV contours were used to create surface meshes, and the spaces between the endocardial and epicardial surfaces were filled with nine-node trilinear brick elements to generate a mesh that was four elements thick. Remote, borderzone and infarct regions of the ventricle were determined based on LV wall thickness, with the borderzone defined as the steep transition in wall thickness between remote and infarcted regions [42]. Each region was assigned different material properties. The mitral leaflets were meshed by creating B-spline curves fit to the points generated by the MRI. The thickness of the leaflets were measured on MRI and assigned corresponding thickness in the model. The posterior and anterior leaflets were assigned different material properties, to correspond with the higher stiffness of the anterior leaflet, and the leaflets were assumed to act as flexible membranes. The chordae were created from beam elements, with strut and edge chordae assigned different material properties. Cardiac myofiber angles were assigned at the epicardium and endocardium, with a linear transmural variation. The fibers in the mitral valve were assigned with the circumferential and radial directions of each leaflet. The LV and mitral valve were modeled as nearly incompressible, transversely isotropic, hyperelastic materials. The chordae were modeled as cable elements, which allows a load to be carried only in tension and does not sustain bending. The final model was observed to correlate closely with experimentally measured volume, pressure and geometry.

Human Models. Obtaining adequately high-resolution images of the mitral valve in human subjects with MRI poses greater challenges than in experimental animals, because of the time involved and the potential toxicity of contrast materials required for image capture. However, combining MRI with transesophageal echocardiography, Ge et al. developed an FE model of the human LV + mitral valve based on data from a single patient with MR secondary to isolated P2 prolapse [43]. They relied on a 3D echocardiographic images obtained intra-operatively to contour the mitral valve and subvalvular apparatus, while using MRI as described above to obtain measurement data from the LV. Images from the two modalities were manually co-registered to produce the final model (Fig. 4).

Modeling Mitral Valve Repair

Of the broad spectrum of mitral valve repairs currently performed, a number have been successfully simulated using FE

Fig. 4 Ge model of human LV + M V with P2 prolapse [43]

techniques. The following is a selection of published repairs demonstrating the versatility of FE modeling in the realm of mitral valve repair. All repairs are performed on a combined model of the valve + LV except where indicated.

Virtual Suture. To model any surgical repair, it is necessary to have a mathematical description of suture placement. Wong et al. described such a technique in 2012, where each suture is modeled as a discrete beam element (*MAT_071, LSDYNA) [44]. This material has the unique property that axial tension can mechanically pull the two ends of each beam element together, allowing approximation of two pieces of tissue, or the mitral valve annulus with an annuloplasty ring. After attachment, the virtual sutures are changed from deformable to rigid elements using the *DEFORMABLE_TO_RIGID function in LS-DYNA.

Annuloplasty. A variety of annuloplasty bands are available on the market, with a range of sizes and shapes. Wong et al. used the sheep LV + MV model in Fig. 3 to investigate the differing effects of saddle-shaped versus asymmetric MA rings on the geometry of the mitral valve, as well as on stress across the valve and basal LV [44]. A representative annuloplasty band of each type was photographed and digitized using KINEMAT MATLAB toolbox. A 3D B-spline was used to represent the 3D geometry of a ring. FE meshes of each ring were created using beam elements, and rings were assumed to be rigid. When applied to the mitral annulus with virtual suture, both ring types restored normal valvular geometry, decreasing the size of the mitral annulus and eliminating MR. Both types also decreased valvular stress on both the anterior and posterior leaflets, as well as the base of the LV, and had no effect on fiber stress at the apex of the LV.

Translating this work to human models, Morrel et al. investigated the leaflet coaptation, valvular stress, and suture stress for three different annuloplasty band shapes, each of varying size, on the combined model of the MV + LV shown in Fig. 4 [45]. They determined that all annuloplasty shapes increased leaflet coaptation and reduced stress across the mitral leaflets and base of the LV; saddle-shaped devices were the most effective, as were smaller sizes across all ring shapes.

Leaflet Resection. Resection of prolapsing leaflet tissue can be performed in a number of ways. The two most common are triangular resection, where a V-shaped wedge of tissue is excised from the valve and sutures are used to reapproximate the resultant tissue gap; and quadrangular resection, where an entire section of leaflet is excised. The former has the advantage of nondisruption of the mitral annulus, and is simpler to perform, while the latter allows for surgical reduction in size of the annulus. Both are typically performed with the addition of an annuloplasty ring, which results in increased long-term durability of repair.

Ge et al. explored triangular resection on an MV + LV combined model (Fig. 5), by removing elements in the P2 region of the posterior leaflet to model the area of surgical excision, and repairing the defect with virtual sutures [43]. This model was derived from imaging data of a patient who underwent surgical mitral valve repair with triangular resection + annuloplasty, and their FE model had good correlation with post-operative echocardiographic images obtained by the operating surgical team. The model demonstrated that resection without annuloplasty increased stress across the posterior leaflet, but decreased anterior leaflet stress both at end-systole and end-diastole. The addition of an annuloplasty band somewhat, but not entirely, mitigated the effect of the increased posterior leaflet stress, while further decreasing anterior leaflet stress.

Modeling Subvalvular Repairs

In functional MR, the LV undergoes remodeling and dilation, which can distort the mitral annulus and induce MR. Further contributing to geometric disruption, the papillary muscles anchored in the left ventricular wall move outward from their normal

position, placing tension on the chordae tendinae, and tethering the valve leaflets. Several “subvalvular” techniques of mitral valve repair attempt to address the distortion of the LV and papillary muscles. To understand the stress placed on the mitral valve after ventricular remodeling, and the stress relieved after ventricular repair, modeling these repair types requires a valve + LV combined model.

Papillary Muscle Anchoring. The technique of papillary muscle anchoring addresses outward displacement of the papillary muscle secondary to ventricular dilatation by creating a physical suture anchor between the PPM and a point on the anterior ventricle or anterior mitral annulus. The anchor pulls the papillary muscle inward, reducing the size of the dilated ventricle and attempting to restore normal valvular anatomy; it is used in conjunction with an annuloplasty ring.

Pantoja et al., using multiple ovine models, explored the effect of PPM anchoring on overall left ventricular myofiber stress across regions of the ventricle [46]. They anchored the PPM to the anterior commissure (AC), posterior commissure (PC), or center point (CP) of the anterior mitral annulus (Fig. 6). For each anchor point, they tested a range of relocation distances, from 10% to 40% of the unshortened diastolic length of the suture. Following anchoring, virtual ring annuloplasty with a saddle-shaped device was performed as described above. Their results demonstrate that stress is reduced in MI, borderzone, and remote regions of the LV with PPM anchoring, regardless of chosen anchor point, and that stress reduction is proportional to the percentage of displacement of the papillary muscle relative to its normal position.

Ventricular Repairs. Several devices exist to correct the dilatation of the LV as a primary means of eliminating MR [13]. An FE model of one such device, “Coapsys,” was created by Carrick et al. [47]. The device is a single chord anchored between the anterior and lateral walls of the LV, reducing ventricular size and therefore theoretically reducing stress on the ventricular wall. Pre- and post-operative MRIs from a single patient, a 58-year-old man with ischemic MR secondary to anterolateral myocardial infarction who underwent Coapsys placement in conjunction with coronary artery bypass, were used to create the model according to the procedure described by Ge et al. [43]. The Coapsys device was modeled as a discrete beam element with a Young’s modulus of 1×10^6 Pa to eliminate stretching, and attached anteriorly and posteriorly to the ventricle in the center of two anchoring pads. Length of the device was assigned based on the patient’s post-operative MRI ventricular dimensions, and axial tension was assigned to the pre-operative model to bring both ends of the beam toward each other until the appropriate length was achieved. They determined that placement of the Coapsys device decreased stress to the LV, both at the area of infarction and remotely, and restored normal mitral valve geometry, decreasing the dimensions of the mitral annulus and eliminating MR. In this way, despite decreasing the size of the LV, they demonstrated augmentation of forward flow of blood out of the heart (stroke volume), because of the eliminated regurgitant blood flow.

Prosthetic Chordae. Another technique to address the abnormalities of leaflet anchoring that lead to leaflet prolapse is placement of prosthetic chordae tendinae between papillary muscles and prolapsing leaflet edges. Typically, these prosthetic chordae are created from polytetrafluoroethylene (PTFE) suture of the approximate length of normal chordae anchoring the nonprolapsing leaflet segments. Variable numbers of prosthetic chords are created, depending on surgeon preference and individual patient anatomy.

Sturla et al. created an isolated MV model of posterior leaflet prolapse to investigate the biomechanical effect of various arrangements of prosthetic chordae [29]. They acquired preoperative cardiac MRI of four patients with isolated posterior leaflet prolapse. This imaging data was combined with intraoperative

Fig. 5 FE simulation of triangular resection of P2 prolapse and annuloplasty [43]. (a) Pre-operative model showing prolapse of the P2 leaflet. (b) After triangular resection of the P2 section of the posterior leaflet. (c) Postoperative model in early diastole. (d) Postoperative model at end-systole with valve closed.

measurements of anterior and posterior leaflet anatomy and number/type of chordae involved in each prolapsing lesion, as well as whether these chordae were ruptured or elongated, and the location of papillary and leaflet edge insertion. For each patient, a model was generated of the mitral valve with papillary muscles and chordae, followed by a second model for each of physiologic MV coaptation, simulating the geometry of the valve in a nondiseased state. They used this physiologically normal model to determine appropriate suture length for five different types of neo-chord implantation, and simulated these repairs, analyzing effects on leaflet geometry, coaptation area, and stress across native chordae, PTFE chordae and regions of the posterior leaflet. The results demonstrated that all repair types corrected MR by restoring leaflet coaptation, and redistributed stress across the valve leaflets and prosthetic chordae, relieving excessive mechanical stress on the normal leaflets. The repairs with more than one prosthetic chord implanted resulted in the greatest redistribution of stress across the repaired area of posterior leaflet.

This isolated valve model assumes the papillary muscles are fixed in space. To better understand the stress on the ventricle after chord placement, our lab is currently exploring neo-chord placement in a valve + LV model.

Future Directions

There are wide-ranging potential applications of the above FE modeling techniques in the realm of mitral valve repair. For example, one could consider development of new medical devices. With these techniques, it is possible to test a novel mitral valve repair device, such as a clip used for percutaneous valve repair, and examine its effects on valve geometry and stress prior to studies on large animals or humans.

The benefits of FE modeling for preoperative planning are manifold. If the surgical team can generate a patient-specific model of mitral valve pathology pre-operatively, and perform a variety of repairs on the model, it should be possible to determine the geometrically and biomechanically superior repair for that patient prior to ever entering the operating room. This may be particularly useful in low frequency, high complexity situations such as congenital malformations of the mitral valve. These conditions are much less common than the lesions discussed here, and surgical correction can involve a combination of many of the techniques described in this paper, such as annuloplasty, leaflet resection, and papillary muscle relocation, into a single procedure. FE modeling has not yet been applied to this type of MV disease.

Fig. 6 PPM anchoring suture placed between the PPM and either the AC, PC, or the CP [46]

There are several major limitations that have prevented the widespread adoption of FE modeling as a surgical tool: first, currently available automated processes to generate FE models from MRI images are prohibitively expensive. Manually generating any model similar to those described above requires long hours of extensive input by scientists with specific computational and engineering expertise. Second, the computing time required to solve any complex FE problem makes this unrealistic for real-time simulation of surgical procedures. Faster computational methods have been applied to circumvent this issue, but none have been widely used [48]. In the future as faster and more automatic techniques are developed, with more accessible user-interfaces, FE modeling has the potential to become a surgeon's best friend, arming them with advanced knowledge about the best operation for their patient.

Surgical simulation is an important, and mandatory, component of surgical education, and no tool has the potential to create higher fidelity simulators than FE modeling. By presenting a surgical trainee with accurate models of the mitral valve and LV which can be manipulated in 3D, they have the ability to explore repair types, and develop their understanding of repair techniques, prior to practicing on live patients. We are currently exploring the FE basis of a surgical simulator suitable for resident and surgeon education, by presolving a variety of the repairs described above on a patient-specific combined LV + MV model, and compiling these results into a library of possible repairs. The human and computing time required to build a model and perform simulated repair is therefore front-loaded, with immediate access to solutions by the end-user. In this way, trainees can investigate the precise biomechanical effects of a variety of mitral valve repairs in real time.

Conclusion

The mitral valve is a complex structure with intricate connections to its surroundings. Diverse processes, both intrinsic and extrinsic to the valve, affect its proper function, and disorders of these processes lead to MR. FE modeling is a powerful tool to accurately describe these pathologic processes, and provide insight into the geometry, stress and strain across the mitral valve and LV; this information is crucial to understanding valvular

diseases and their surgical treatments. The interrelationships between the valve and ventricle can only be understood in FE models that incorporate the entire mitral valve, subvalvular structures, and ventricle. Using such models, a variety of simulated surgical repairs have been performed as described above, and we believe these techniques have extensive applications in realms such as device development, preoperative planning, and surgical education.

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