# Aortic root simulation framework for valve sparing aortic root replacement surgery

Tamás Umenhoffer,[1] Márton Tóth,[1] László Szécsi,[1] Ágota Kacsó[1] and Balázs Benyó[1]

[1] Department of Control Engineering and Informatics, Budapest University of Technology and Economics, Budapest, Hungary

**Abstract**

*In this paper we present a framework for the elastic and fluid simulation of the aortic root. Our goal is to measure the effectiveness of a new aortic valve sparing surgery method, where the patients own valve angular distribution is preserved. Our workflow covers the patient specific aortic root geometry modeling for CT or MR data, virtual aortic valve sewing and blood flow calculation in the aortic root. Our simulation techniques use simplifications applicable directly for aortic valve simulation resulting in raised effectiveness.*

Categories and Subject Descriptors (according to ACM CCS): I.6.3 [Simulation and Modelling]: Applications

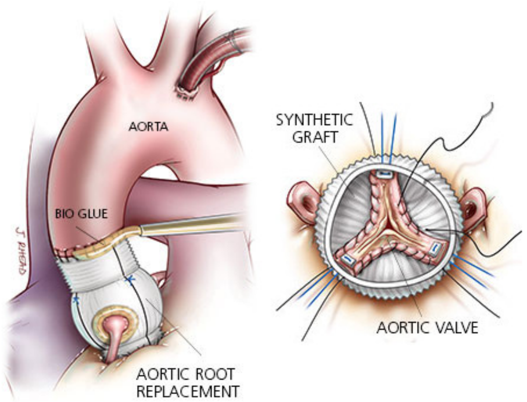


**Figure 1:** *Valve sparing aortic root surgery* †.

## 1. Introduction

Aortic root surgery is necessary if a disease has developed that affects the portion of the aorta closest to the heart: the aortic root. The most common diseases are high blood pressure and Marfan syndrome (when the connective tissues are weakened due to abnormal production of fibrillin), which causes aortic root enlargement or aortic root aneurysm. In case of these diseases the aortic valve is not diseased, unlike in case of aortic valve stenosis, when the valves are too narrow and stiff.

If the valves are not diseased valve sparing aortic root replacement can be used. During these operations the patient's aortic root with valves are removed and replaced with a synthetic tube called graft. However the patient's own valves are preserved and sewed back into the inside of the graft. The coronary arteries are also attached. This operation, unlike using synthetic valves, avoids the use of life-long blood thinners, thus recommended for active and young patients. The graft is highly durable and does not require any additional medicines, while the patients own valves potentially will last the rest of their lives.

Figure 1 shows the aortic root after valve sparing aortic root surgery and the inner side of the graft after the valves are sewed in. On the left side we can see a black line running on the outer side of the graft. This has special meaning, it shows where the comissures of the valves should be sewed in. These lines are drawn on the grafts by the producer of the graft and the three lines have even angle distribution on the circle of the tube. However, patients have their own specific comissure angle distributions, which is usually not considered by the surgeon and the valves are sewed in the predefined even distributions. This is because the measurement of patient specific angles are not straightforward dur-

† Image courtesy of Intermountain Medical Center Heart Institute (https://intermountainhealthcare.org/services/heart-care/treatment-and-detection-methods/aortic-root-replacement/)

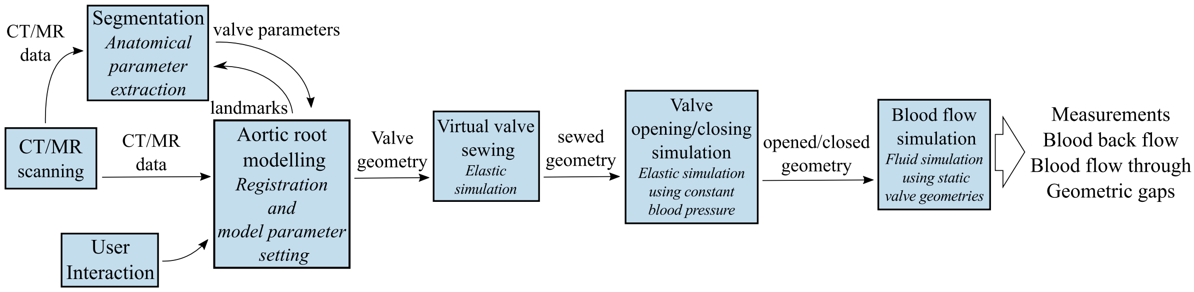


**Figure 2:** *The main workflow of our framework.*

ing he operation, but recently special tools have been developed that make this measurement accurate and fast, thus enables the sewing using the patient specific distributions. Yet no database exists that would show the physiological benefit of the more advanced surgery technique, though it probably has. Figure 3 shows the difference between the two valve sewing methods.

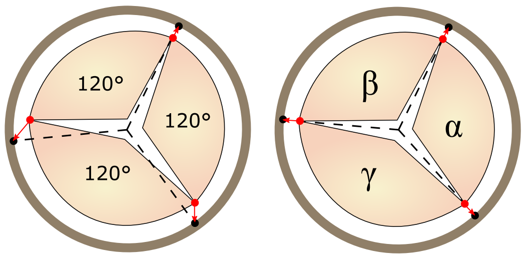


**Figure 3:** *Valve sewing methods with even and with patient specific leaflet angle distributions.*

Our goal is to try to prove the benefits using computer simulations [8]. We defined a framework to create patient specific geometric models of the aortic root and the valves and simulate valve opening and closing and blood flow in case of both surgery techniques. We believe that measuring the blood back flow and flow through or the geometric gaps between valve leaflets will provide us the necessary information.

## 2. Main workflow

The main workflow of our system is shown on Figure 2. As we need patient specific aortic root models we use medical imaging data, particularly CT or MR scans. These scans serve only as a base for the modeling procedure, which is not fully automated, it needs significant professional user interaction. If the geometric model is ready we can run our simulations.

In most aortic root simulation systems a coupled elastic and fluid simulation is used. It means that elastic and fluid simulation steps are run alternately and iteratively to take both phenomena into account. The main drawback of this technique is that data conversion is usually needed when changing to one solver to an other. We avoid this problem with making simplifications applicable specifically for the aortic valve surgery problem.

Our final measurements that show the effectiveness of a surgery technique would be the amount of blood that flows through the aortic root when the valves are in opened position. Similarly the amount of blood that can flow back if the valves are closed is also a very important value. We can also measure geometric gaps between the valve leaflets in closed position. These measurements does not require coupled elastic-fluid simulations, only fluid simulation with a static geometry (the opened or closed leaflets).

As the fully opened and closed state of the valves take only a short period of one hearth cycle, when the blood pressure in the aorta is defined by the maximal ventricular pressure, no fluid simulation is needed to calculate the opened and closed valve shapes, only elastic simulation. During these elastic simulations blood pressure is modeled with a constant force. Virtual sewing of the aortic valves should also be simulated with the purely elastic model.

As we are interested in the valve functioning after surgery, the exact patient specific aortic root model is not used, only the models of the valves. The grafts has a simple tube shape which can be modeled analytically in contrast to the more complex shapes of the sinuses of Valsava (see Figure 4).

Summing these simplifications together our workfow is the following. We created an aortic valve modeler software, that uses medical data and user interaction to create the geometry of the valves. These valves are virtually sewed onto the inner walls of a tube with given radius using elastic simulation. When the simulation reaches a stable state the geometry is saved and passed to a second elastic simulator, that applies constant force to the valves resulting on opening or closing based on the direction of the force. The geometry in the two extreme cases is saved and passed to a fluid simulator which simulates blood flow through the valves and measures the amount of passed through and back flow blood. The following sections describe the building blocks in more detail.

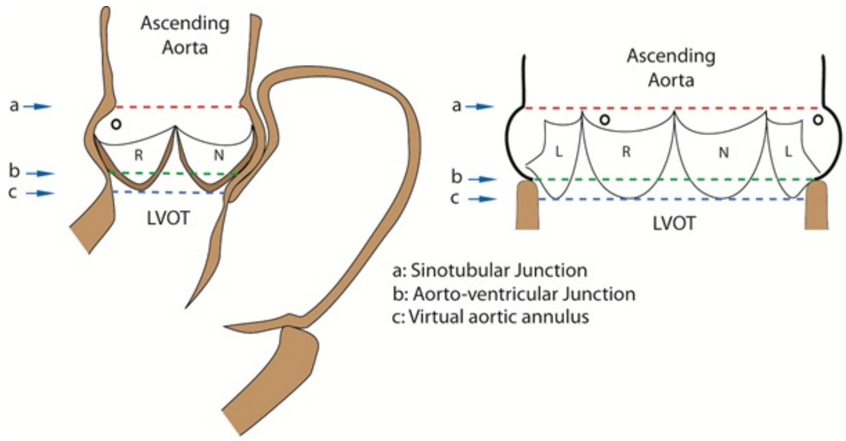


**Figure 4:** *The anatomy of the aortic root and aortic valves.*

| Name | Type | Default Values | Description |
|---|---|---|---|
| commissure1 | 3D vector | (0, 17.5, 13.44)mm | Common suspension point of the L and N valves. |
| commissure2 | 3D vector | (11.64, 17.5, -6.72)mm | Common suspension point of the R and N valves. |
| commissure3 | 3D vector | (-11.64, 17.5, -6.72)mm | Common suspension point of the L and R valves. |
| leafletTipPoint | 3D vector | (0, 15.03, 0)mm | Valve intersection point. |
| annulusRadius | float | 11.58 mm | The radius of the annulus |
| valsalvaRadius | float | 14 mm | The largest radius of the aortic root |
| valsalvaMaxRadiusHeight | float | 0.25 | Position of maximal radius on aortic root axis (0-1) |
| ostiumR | 3D vector | (0, 14, -14.2) mm | Base point of right coronal ostium |
| ostiumL | 3D vector | (-12.2, 14, 7) mm | Base point of left coronal ostium |
| valsalvaSlope | float | 0.05 | Slope of the sinuses of Valsalva |
| valsalvaCurvature | float | 3 | Curvature of the sinuses of Valsalva |

**Table 1:** *Aortic root modeling parameters.*

## 3. Modeler application

The purpose of our modeler application is to generate the patient specific geometric model of the aortic valves. We can also define the geometry of the aortic root (sinuses of Valsalva) and the left ventricle, though they are not used in the physical simulations, they serve as a big help in the rigid registration of medical data, and to determine the exact valve parameter values. All the ventricle and aortic root parameters can be given on a proper GUI panel, table 1 lists all these parameters, their type and their default values.

To determine the exact patient specific values of these parameters we need to visualize the medical data of the patient. The left side of Figure 5 shows the volume visualization of a CT data in our application. CT is more preferred as it has higher resolutions, so fine anatomic features can be identified. It can be seen that after loading this data it is misaligned with our model, so our first task is the rigid registration of the two models. Later, during simulations we always assume that the axis of the aortic root is always pointing in the up direction of the three dimensional virtual space. The axis is defined by the center of the aortic root annulus and the center of the sinotubular junction (STJ), i.e. the center of *a* and *c* on Figure 4. In practise this axis is never aligned with the axes of CT or MR scanners so the rotation and translation of medical image data must be determined.

Our application offers a completely manual workflow to determine this transformation. Beside the 3D volume visualization 2D scan slices also help to find the correct values, which can be seen on the right of Figure 5. To determine the axis of the aortic root the user needs to define two landmark points, one for the center of the annulus and one for the center of the STJ. For an experienced user with the use of sagittal, coronal and axial views finding these landmarks is not a hard problem. On the proper CT slice a cursor can be placed, which exactly defines a point in the three dimensional space of the scanner. Using these landmarks the CT model can be translated and rotated to its place. Only the axial rotation can not be determined automatically, but us-

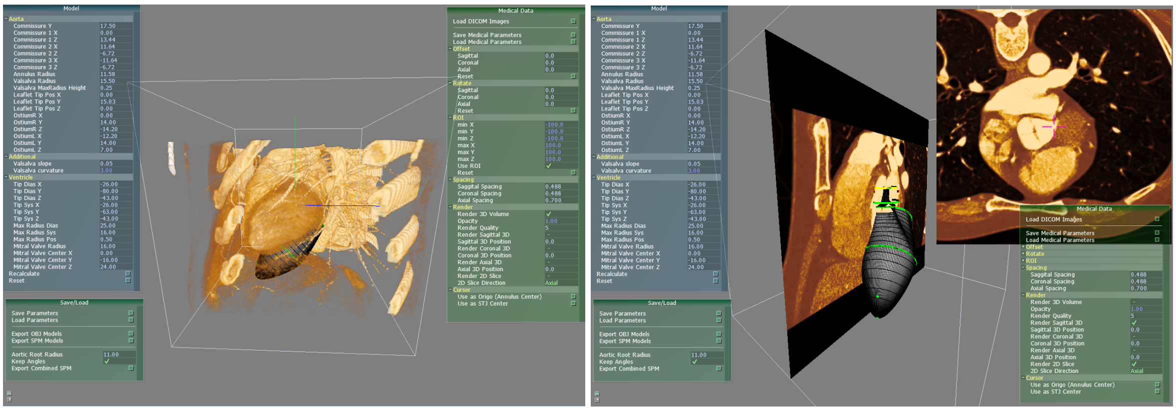


**Figure 5:** *3D volume and 2D CT slice visualization of CT data in our modeling software.*

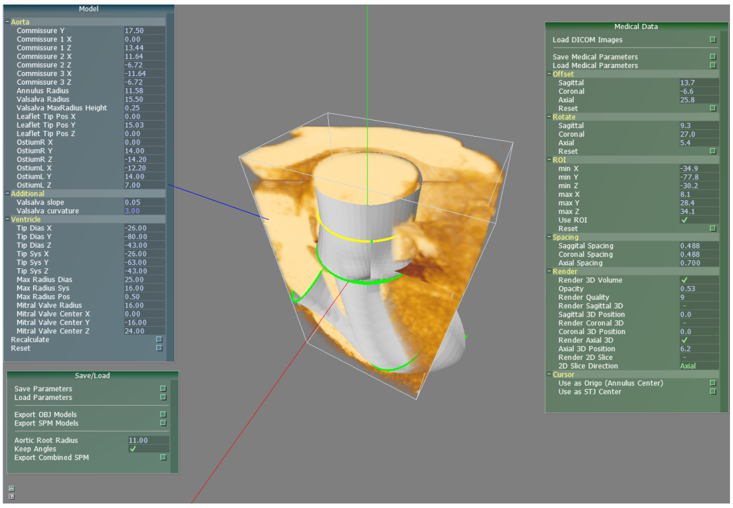


**Figure 6:** *A registered CT data with properly set anatomical parameters.*

ing the 3D volume visualization and examining the coronal arteries, the model can be easily rotated to align with our abstract aortic root model.

If the CT data is properly registered, we can easily set the aortic root parameters using the CT slice and 3D CT volume visualizations. Figure 6 shows a properly registered data with anatomical parameters set. The resulting geometry describes the patient specific anatomical features quite well. The transformation parameters used in registration can be saved and reloaded as well as the current values of the anatomical parameters. For an experienced user registration and parameter setting takes up to 10-15 minutes. The final model can be exported in our own file format, which will be processed by or simulation softwares. Additional information like the radius of the virtual graft, or whether we virtually sew the valves in even angles or keep their angle distribution can also be given and will be saved in the output file.

We also examine the possibility to use segmentation algorithms to determine the two landmark points automatically. We have successfully extracted the axis of the ventricle and the position of the annulus junction, but haven't integrated into the modeler yet. The segmentation will also help to hide unwanted features (tissues that are not part of the aortic root) in our volume visualization.

## 4. Elastic Simulation

We used a finite element simulation technique [1] for the simulation of the elastic deformations of the valve leaflets. Finite element methods divide the interior volume of the objects into small, simple geometric elements. These elements should fill the space without gaps and overlapping. The elastic behavior of one element can be handled analytically due to its simple geometry and assuming that the deformation field is linear inside the element. A common choice of element shape is a tetrahedron, which has four vertices and four faces. Figure 7 shows the deformation of such an element, where rest shape vertices are labeled with $x$ and the deformed vertices are labeled with $p$. Using the displacement of the deformed vertices, elastic forces on the faces of the tetrahedron can be calculated and be distributed on its vertices. All elements can be handled independently, but a small time step should be used, which makes the simulation a time consuming process. This explicit finite element method on the other hand can produce accurate results and was used successfully used in elastic simulations [2].

For solid geometries of arbitrary shapes the tetrahedral model is general enough to be used efficiently. However our valve model does not actually have a volume as it is an infinitely thin layer. To be able to use it in a finite element simulation we should give a thickness to it which can be complicated in the sharp corner areas near the leaflet tip points. If we assume that our material has weak resistance against bending, and we focus only on the stretching of the geometry, our model can be simplified and geometry extrusion can be avoided.

We defined a thin elastic model, which assumes that the material is flattened in a plane, has constant thickness and no shearing or tearing occurs along the plane normal. In other words a two dimensional version of the finite element simulation will be used. In the two dimensional space the simplest element shape is a triangle, and giving it a fixed thickness we get a triangular prism as our base elements of simulation. Figure 8 shows such an element and its deformation. Deriving the elastic equations for a triangular prism is not harder than for a tetrahedron, and without deformations along the extrusion axis our mathematical model will be even simpler and can be evaluated more efficiently.

One problem still remains, namely our simulation is actually not a two dimensional simulation but tree dimensional as the leaflets are transformed and bended in the three dimensional space. The solution to this problem is to run the thin elastic simulation in the tangent space of the leaflet geometries. Thus for each prism element we first transform it to its tangent space, evaluate the elastic forces and transform it back to three dimension. This transformation is a translation and rotation which does not affect deformation tensors. Thin layer elastic simulation has the advantage of being more stable, as it neglects large scale bending deformations that makes finite element simulations (and especially explicit versions) instable.

The same elastic simulation was used in the virtual sewing and in the valve opening and closing simulations. In case of virtual sewing some of the vertices have a fixed target position and a force is computed in each iteration which moves them towards these target locations. These vertices are the vertices corresponding to the valve attachment points on the graft. The second group of the vertices correspond to the valve edges sewed to the inner side of the graft. They don't have fixed positions but should be projected onto the virtual cylinder geometry of the graft during simulation. Again a force is computed that moves them toward the graft. These vertices will relax according to the elastic forces, but slide on the walls of the graft. The last group of vertices are the remaining vertices that can move without restrictions, elastic simulation has full control over them.

During valve opening and closing, the attachment points and attachment edges of the sewed valves are fixed, and remaining vertices are simulated elastically applying constant force on them. The amount of force can be calculated from the maximal ventrical pressure and the area of aortic annulus. The later is given by our anatomical model, and for the former an average human heart pressure can be used (120 Hg mm, i.e. 16 000 $N/m^2$), or in many cases we have patient specific information about this value. Our thin elastic model requires the elastic parameters and thickness of the material. We can find concrete elastic parameter values of the valves in several publications [6, 5]. We used 0.3 MPa for Young modulus and 0.38 for Poisson ratio. Though the Poisson ratio of real world material is higher, we had to decrease it to aviod instability. This limitation was described in other physical simulation papers too [4]. The thickness of the valve tissues is also a known value [9] and ranges between 0.36 and 0.8 mm, finally we used 0.6 mm.

Figure 9 shows the elastic simulation of the opening of the valves using our thin elastic finite element simulation method.

## 5. Fluid Simulation

If the fully opened and closed versions of the valve geometries are present we can use fluid simulation to check if they functioning properly. We have several options for fluid simulations, but as the closing valves have very narrow gaps between them, we should choose a method that can handle these geometric properties. Regular grid based methods are not suitable, but irregular grid based and particle based methods can be used. We investigated both cases, and created a smoothed particle hydrodynamics (SPH) [3] and a finite volume fluid solver.

The main advantage of the SPH method is that the simulation space does not have to be resampled on an irregular grid, the poligonal geometry result of the elastic simulation can be used right away. However the compressible property of blood is harder to ensure. The basic SPH method can also be extended to handle both elastic and fluid materials in a single simulation. We investigate this direction too, but we believe that the separated elastic and fluid simulations would result more accurate results. We performed basic fluid simulation test with our SPH solver, but it is not yet implemented into our framework.

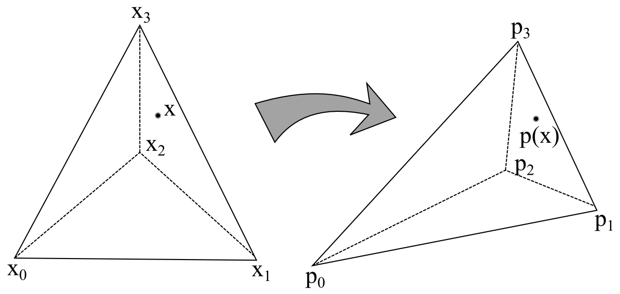


**Figure 7:** *Tetrahedral finite element and its deformation.*

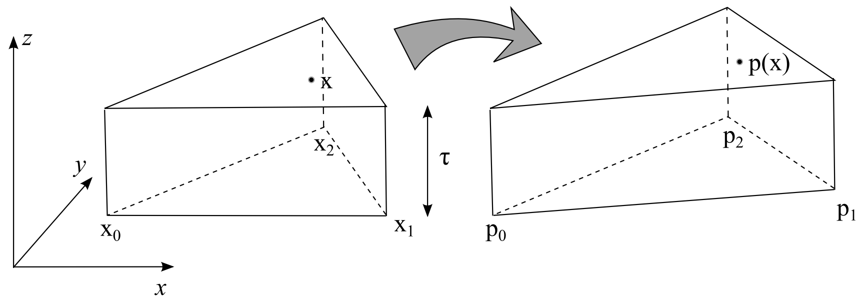


**Figure 8:** *A triangular prism finite element and its deformation. The prism should deform only in the xy plane, no shearing or tearing occurs along axis z. τ is the thickness of the material.*

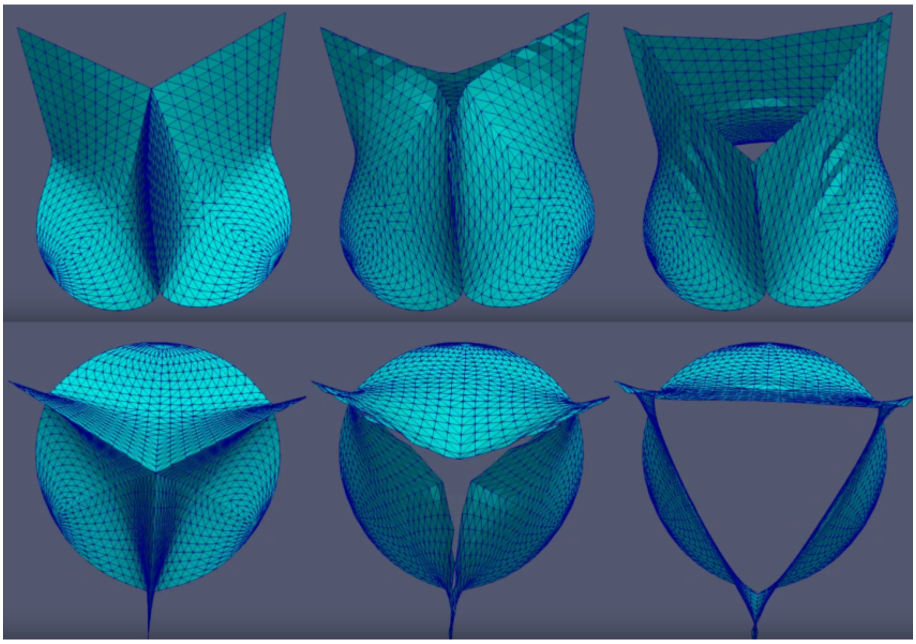

**Figure 9:** *Simulation of the opening of an aortic valve.*

The finite volume method is a finite element method used in fluid simulation, where the simulation space is divided to an unstructured grid and fluid flow is computed between these volume elements. We have implemented a control volume method [10], which stores fluid variables at the vertices of the unstructured grid, which makes fast rebuild of the grid possible. We implemented a 2D version of the algorithm, which will be extended to a 3D version and later can be integrated to our framework.

## 6. Implementation

Our modeler application is an OpengGL based interactive standalone application written in C++, yet running on Windows platform only. Our elastic solver is an OpenCL based standalone console application that uses the graphics processing unit (GPU) to accelerate computation [7]. It runs on Windows and Unix platforms. We created an efficient running environment, where several simulation parameter sets can be defined and the simulations can be batch executed on remote machines. The fluid solvers are also GPU based standalone applications using the CUDA technology.

## 7. Conclusion

In this paper we defined a complete framework to examine the results of a valve sparing aortic root replacement surgery using computer simulations. We implemented a modeler application to build a patient specific aortic valve geometry. We implemented a specialized elastic solver to simulate valve sewing and valve opening and closing. We also implemented fluid simulation solvers which will provide the final measurement that can help to judge the effectiveness of the surgery. Our system is still under development so evaluation of real scenarios are yet still to come.

## Acknowledgements

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