◆ EXPERIMENTAL INVESTIGATION ◆

An Experimental and Numerical Comparison of the Rupture Locations of an Abdominal Aortic Aneurysm

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Purpose: To identify the rupture locations of idealized physical models of abdominal aortic aneurysm (AAA) using an in-vitro setup and to compare the findings to those predicted numerically.

Methods: Five idealized AAAs were manufactured using Sylgard 184 silicone rubber, which had been mechanically characterized from tensile tests, tear tests, and finite element analysis. The models were then inflated to the point of rupture and recorded using a high-speed camera. Numerical modeling attempted to confirm these rupture locations. Regional variations in wall thickness of the silicone models was also quantified and applied to numerical models.

Results: Four of the 5 models tested ruptured at inflection points in the proximal and distal regions of the aneurysm sac and not at regions of maximum diameter. These findings agree with high stress regions computed numerically. Wall stress appears to be independent of wall thickness, with high stress occurring at regions of inflection regardless of wall thickness variations.

Conclusion: According to these experimental and numerical findings, AAAs experience higher stresses at regions of inflection compared to regions of maximum diameter. Ruptures of the idealized silicone models occurred predominantly at the inflection points, as numerically predicted. Regions of inflection can be easily identified from basic 3-dimensional reconstruction; as ruptures appear to occur at inflection points, these findings may provide a useful insight into the clinical significance of inflection regions. This approach will be applied to patient-specific models in a future study.

J Endovasc Ther. 2009;16:322–335

Key words: Abdominal aortic aneurysm, in vitro model, experimental study, numerical, rupture, inflection

Aneurysms are permanent and irreversible localized dilatations¹ that can form in any blood vessel, but the majority of the more

serious ones occur in the infrarenal section of the abdominal aorta. If left untreated, abdominal aortic aneurysm (AAA) can eventually

This work was financially supported by the Irish Research Council for Science, Engineering and Technology (IRCSET) Grant RS/2005/340 and also Grant #R01-HL-060670 from the US National Heart Lung and Blood Institute, Bethesda, Maryland, USA.

The authors have no commercial, proprietary, or financial interest in any products or companies described in this article.

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rupture. Approximately 500,000 AAAs are diagnosed worldwide each year,² resulting in 15,000 deaths per year in the US alone.³

Typically, an AAA is surgically repaired once it reaches/exceeds a diameter of 5 to 5.5 cm. There have been reports that this threshold may lead to inaccurate surgical decision making because not only can smaller AAAs rupture,⁴⁻⁶ but also large AAAs can remain stable.⁶ By applying the definition of material failure, we know that an AAA will rupture when the locally acting wall stress exceeds the local wall strength. Currently, much research is aimed at examining the AAA wall stress distributions with a view to improving rupture prediction.⁷⁻¹⁶ Researchers have also examined methods to determine the strength of the AAA wall, both invasively from excised tissue¹⁷⁻¹⁹ and noninvasively.²⁰ While there is much focus on attempting to numerically understand the mechanics of AAA rupture, only limited work has focused on the development of experimental methods of determining rupture potential.

Previous work by Morris et al.²¹ reported the use of the photoelastic method in determining wall stress distributions in an idealized AAA case that used dimensions taken from a comprehensive study.²² Morris et al.²¹ observed rupture of the photoelastic model at the proximal and distal inflection points. This is the only publication in which rupture of an experimental AAA model has been reported. Callanan et al.²³ later validated this photoelastic work numerically. Other researchers²⁴ have also employed an idealized AAA in experimental studies, this time using strain gauges attached to a latex model to determine the stress in the AAA wall. The technique of creating rubber models of AAAs has been reported previously by Doyle and colleagues,²⁵ which was adapted from related techniques developed by O'Brien et al.²⁶

The primary purpose of this study is to identify the sites of rupture in idealized AAA models by performing in vitro rupture studies and to compare these findings with those predicted using numerical modeling. Mechanical characterization of the material used was also performed, along with the examination of silicone model wall thickness.

Figure 1 ♦ (A) Type 2 dumbbell specimen from BS ISO 37. (B) Modified trouser test piece from BS ISO 34-1; *a* indicates the direction of the cut. All dimensions are in millimeters, and the thickness of each specimen is 2 mm.

METHODS

Material Characterization and Validation

Silicone (Sylgard 184; Dow Corning, Midland, MI, USA), an elastomeric poly(dimethylsiloxane) (PDMS), was used to construct the physical models for this study. This transparent silicone has an ultimate tensile strength (UTS) of 7.1 MPa and a tear strength of 2.6 N/mm, according to the manufacturer's specification sheet. Supplied in 2 parts, Sylgard 184 is mixed in a 10:1 ratio of base to curing agent. Although silicone rubber and arterial tissue are not identical in their stress/strain behavior, silicone remains the most suitable analogue.

Sample preparation. Aluminum molds were designed and manufactured to be used with the injection-molding method to create the silicone tensile testing samples. The mold cavity for the tensile test samples conformed to a type 2 tensile test specimen as outlined in BS ISO 37 standards, with a second set of molds designed for creating trouser test samples to be used for tear testing, per BS ISO 34-1 standards. The dimensions of these samples are shown in Figure 1. The samples prepared for tear strength testing differed slightly in dimensions from those outlined in BS ISO 34-1.

The trouser test (Fig. 2) specimens had "legs" that were extended in opposite directions to determine the tear strength of the material. The protocol for preparing samples for testing is outlined in Appendix I. Briefly, the molds were thoroughly cleaned and bolted together, and the necessary amounts of silicone were manually mixed. Degassing was required at this stage as air bubbles can get trapped in the silicone during the mixing process. Once all air was removed, the

Figure 2 ♦ Illustration showing the loading directions that are applied to the trouser test specimen using the tensile test machine.

silicone was transferred from the mixing container to a syringe, where it was then injected into the mold cavity. Further degassing in a vacuum was required after the injection process. The mold was then placed into an oven at 45°C for 18 hours to ensure adequate curing of the samples. Once cured, the samples were removed from the mold and left at room temperature for 3 hours before further preparation for both tensile and tear testing.

Uniaxial tensile testing (Appendix II). Using a test apparatus (H25KS; Tinius Olsen, Surrey, UK) with a 1-kN load cell, an extension rate of 500 mm/min was applied, as recommended in BS ISO 37 for type 2 specimens. Each sample was preconditioned in an initial 10-cycle, 20% extension stretch-relax program to stabilize the stress-strain function of the material and overcome the Mullin's effect.²⁷ The sample was then stretched to failure. Nine samples were used (the BS ISO 37 recommends that at least 3 samples be used to determine material properties).

Tear testing (Appendix III). The Tinius Olsen test apparatus was used to apply an extension rate of 100 mm/min to each of 12 samples per BS ISO 34-1. Figure 2 shows the load directions that were applied to the trouser test specimen when the legs were pulled in opposite directions until the point of complete material failure.

Strain energy function. To mechanically characterize the material, the experimental data from the tensile tests was converted to engineering stress and engineering strain and

then a 4th order polynomial curve was fitted to the data to obtain a representative experimental data curve. The polynomial curve applied to the experimental data was then used as the material model for the finite element analysis (FEA), which was performed with the ABAQUS commercial FEA solver (version 6.7; Dassault Systems, Simulia, RI, USA). A dumbbell model identical to that of the tensile tests was replicated numerically, with the 4th order polynomial curve applied to represent the experimental data; this was used to determine the most suitable strain energy function (SEF). Various SEFs were examined, such as the Neo-Hookean, Ogden, Yeoh, and Mooney-Rivlin, to determine the most applicable SEF. Material constants could then be determined for any chosen SEF, thus allowing the material to be readily modeled in numerical studies.

Finite element analysis. By replicating the tensile test experiments using a numerical simulation, the accuracy of the material coefficients could be examined. For this, the numerical model was simulated in the same manner as that of the tensile tests, with a half-model analyzed due to symmetry. The model was constrained from all movement at one end, with a displacement constraint identical to that of the tensile test applied to the opposite end. Mesh independence was determined by examining the stress and strain experienced in a particular node at the center of the gauge length over the duration of the loading. This was performed for mesh densities of 260, 1752, and 5000 elements. A mesh of 1752 elements was deemed adequate as there was no significant difference in results when the mesh density was increased to 5000 elements.

Rupture Modeling

Experimental AAA model. In order to study the rupture of the aneurysm in vitro, a previously reported^{21,23,26} idealized AAA model was used. This ideal AAA was developed using realistic dimensions obtained from the EUROSTAR data registry.²² In brief, the ideal AAA was geometrically symmetric, with a maximum internal diameter of 50 mm, a total length of 260 mm, and a 2-mm

Figure 3 ♦ Experimental setup of rupture test. (1) Retort stand, (2) plug, (3) clamp, (4) silicone rubber AAA model, (5) plug, (6) pressure gauge, (7) mirror-walls, (8) pressure regulator, and (9) pneumatic air-line. Mirror walls allow 360° of AAA model to be examined.

uniform wall thickness. Experimental ideal AAA silicone models were manufactured using the lost-wax process previously reported by our group.^{25,26} Using this technique, male and female aluminum molds were designed using computer-aided design (CAD) and machined using a computer numerical control (CNC) 3-axis milling machine. The female mold allowed the creation of a wax model of the AAA that upon solidification was then placed into the outer male mold. This male mold had a uniform wall cavity of 2 mm that facilitated the injection of silicone around the wax model. Once the silicone rubber was cured, the complete model could be removed and the wax melted out from the model. To ensure the optimum uniform wall thickness of the resulting silicone rubber model, the female mold was heated to slow down the solidification process of the wax model, which reduces the amount of shrinkage involved with the cooling process, resulting in more accurate wax models. An accurate wax model allows a uniform 2-mm cavity to surround the wax upon placement in the male outer mold. Five AAA models to be used in the rupture study were manufactured using this technique, with a further 4 models created to determine wall thickness.

Experimental setup. The experimental test rig consisted of a series of mirrors, a pneumatic airline, a pressure regulator, a pressure

manometer, a high-speed camera and various clamps and silicone tubing. The use of correctly positioned and angled mirrors allows 360° of the model to be viewed, which is of paramount importance to the success of the experiments (Fig. 3). A high-speed camera (Olympus i-Speed; Olympus Corporation, Tokyo, Japan) was used to capture the point and location of rupture. This camera was capable of recording images at rates up to 33,000 frames per second (fps). A pixel resolution of 800×600 at 1000 fps was deemed adequate for this application, and images were recorded using a monochrome image sensor. Once the AAA model was attached to the test rig, the high speed camera was adjusted to ensure optimum focus and angle. After a satisfactory setup was established, the air pressure was increased and the camera set to record. Air pressure was incrementally increased by 20 mmHg every 30 seconds until rupture occurred. The pressure readings were also recorded in the video image and so could be examined post rupture.

Measurement of wall thickness. In order to determine the variations in wall thickness throughout the AAA models, 4 additional silicone rubber ideal AAA models were created. These models were sectioned at 20-mm increments along the longitudinal distance of each model (Fig. 4), similar to a measurement

Figure 4 ♦ Cross-sections of ideal AAA rubber model. Cross-sections (left) correspond to slices shown on model (right). Slices were taken at 20-mm longitudinal increments. Image shows relative uniformity of wall thickness throughout the rubber model.

technique previously reported.^{25,26} Cross sections were then measured for wall thickness at 4 90° equidistant locations. Results for each region were averaged across the 4 models so that regional variations in thickness could be compared. Wall thickness measurements were then analyzed using SPSS 15 (SPSS Inc., Chicago, IL, USA) to determine the 25th and 75th percentiles, maximum and minimum wall thickness, and the mean $\pm$ standard deviation.

Numerical Modeling

To correlate experimental and numerical results, the evaluated material constants were applied to the ideal AAA model and implemented in the FEA software. The pressure loadings observed experimentally were then applied numerically. Boundary conditions similar to the experimental setup were implemented in the virtual AAA model, so that the model was constrained from all movement at the proximal neck and iliac legs. Due to the symmetrical nature of the model, only half the model was numerically analyzed using symmetry constraints to reduce computational time. Each model was meshed using 3-dimensional (3D) stress elements with mesh independence performed as previously reported.¹³ This mesh independence test involves increasing the number of elements of

Figure 5 ♦ Representative numerical ideal AAA, uniform wall. (A) The finite element mesh and (B) loading conditions and constraints.

the mesh until the peak stress is <2% of that recorded with the previous mesh size. Initially, a uniform 2-mm thick model was subjected to an internal pressure of 364.6 mmHg, which is an average of the 5 rupture pressures observed during the experiments. The model was also tested using all 5 of the actual experimental rupture pressures. Peak von Mises wall stress was analyzed along with peak wall tension.¹⁹ In order to gauge the effect of wall thickness on the resulting wall stress, AAA geometries were created using varying wall thickness corresponding to the 25th and 75th percentiles, maximum and minimum wall thickness, mean wall thickness, a thick walled inflection region, and a thin walled inflection region. This required the creation of an additional 7 numerical models. Each of the models was subjected to an internal pressure of 120 mmHg, which is commonly regarded as the average peak systolic pressure of the cardiac cycle. By applying a common internal pressure, the stress on the AAA wall could be compared across each model. A sample numerical model can be seen in Figure 5, which shows the finite element mesh and boundary conditions used in the study.

RESULTS

Material Characterization and Validation

Uniaxial tensile test. Tensile testing revealed that the mean UTS of Sylgard 184

Figure 6 ♦ Engineering stress and engineering strain experimental data (gray lines) with 4th order polynomial curve applied to the results (black line). Engineering stress refers to the stress associated with the initial cross-sectional area of the material. Engineering stress is defined as $\sigma_E = F/A$, where σ_E is engineering stress, F is force, and A is cross-sectional area. Engineering strain is defined as $\varepsilon_E = \Delta l/l$, where ε is engineering strain, Δl is the change in gauge length, and l is the original gauge length.

was 7.7361 ± 1.6597 MPa (range 4.74–9.39) compared to 7.1 MPa on the manufacturer's specification sheet. Figures 6 shows the stress/strain curves for each sample, along with the 4th order polynomial curve fit.

Tear test. Tear testing showed that the mean tear strength of the material was 0.419 ± 0.094 N/mm. The manufacturers reported a 2.6-N/mm tear strength on the specification sheet, but it is known that results can depend heavily on the type of tear test sample used²⁸; therefore, it is difficult to compare our results with those of the manufacturer. For this study, the average tear strength of 0.419 N/mm was taken as the maximum wall tension that the material can resist prior to rupture.

Strain energy function. For this particular material, the 3rd order Ogden SEF proved to be the most suitable constitutive equation because it provided a good curve fit to the data ($R^2 = 0.9812$) and was also stable at all stresses and strains. The ABAQUS form of the Ogden SEF is shown in Eqn. 1:

$$W(\lambda_1, \lambda_2, \lambda_3) = \sum_{p=1}^N \frac{2\mu_p}{\alpha_p^2} (\lambda_1^{\alpha_p} + \lambda_2^{\alpha_p} + \lambda_3^{\alpha_p} - 3), \quad (1)$$

TABLE 1

Material Coefficients Determined for the 3rd Order Ogden Model for Sylgard 184

	μ	α
1	-304.235	1.2667
2	148.232	1.5962
3	157.156	0.9075

where W is the strain energy density per undeformed unit volume, $(\lambda_1, \lambda_2, \lambda_3)$ are the principal stretch ratios, α is a strain hardening exponent, and μ has the interpretation of the shear modulus under infinitesimal straining. The resulting material coefficients are presented in Table 1.

Finite element analysis. The results of the FEA and the uniaxial tensile tests compared favorably (Fig. 7), thus establishing confidence in the material characterization of Sylgard 184.

Rupture Modeling

Experimental rupture tests. Of the 5 silicone models ruptured in vitro, 4 models experienced rupture at a region of inflection on the surface of the model. An inflection point is defined as a point on the AAA surface at which the local AAA wall shape changes from concave outward to concave inward.²⁹ This finding is consistent with previous numerical and experimental reports by our group^{13,21,23} and others,²⁹ who noted peak stresses occurring at these regions instead of at the maximum AAA diameter. A summary of the rupture locations along with corresponding burst pressures is shown in Table 2. One silicone model ruptured at the iliac bifurcation. The sequence of events leading to rupture for Test 1 can be seen in Figure 8, which shows the frame where the material failure initiates, leading to rupture, and ultimately complete failure of the silicone model.

Wall thickness results. Wall thickness was assessed by slicing the 4 ideal AAA models into cross sections (Fig. 9). These 4 models were made using the same process as the previous AAA models. Table 3 shows the average regional wall thicknesses corresponding to the sections in Figure 9. This

Figure 7 ♦ Comparison of results for the experimental tensile tests and the numerical simulation of Sylgard 184. True stress refers to the stress associated with the current cross-sectional area of the material. True stress takes into account the change in cross-sectional area. True stress is defined as $\sigma_T = \sigma_E (1 + \epsilon_E)$, where σ_T is true stress, σ_E is engineering stress, and ϵ_E is the engineering strain. True strain is defined as $\epsilon_T = \ln(1 + \epsilon_E)$, where ϵ_T is true strain and ϵ_E is the engineering strain.

table also shows the wall thickness used to create numerical models, with thick and thin regions at the inflection points. The minimum wall thickness recorded in any model was 1.17 mm and was located at the iliac bifurcation region. The maximum wall thickness was 2.53 mm and was also found at the iliac bifurcation region.

The effect of wall thickness on stress distributions within the AAA sac is displayed in Figure 10. The uniform 2-mm wall model appears to represent the overall behavior of the AAA model regardless of variations in wall thickness, with the exception of the “min” thickness model. This study highlighted that the inflection region, in particular the proximal inflection region, experiences higher stresses than the maximum diameter region regardless of wall thickness.

TABLE 2

Summary of Experimental Rupture Results

Test	Rupture Location	Rupture Pressure, mmHg
1	Proximal inflection point	254.7
2	Proximal inflection point	278.6
3	Proximal inflection point	466.2
4	Distal inflection point	278.7
5	Iliac bifurcation	544.6

Numerical Modeling

Stress distributions on the surfaces of the virtual AAA model reveal that high stresses occur at the regions of inflection and not at regions of maximum diameter. This has been previously shown by our group for this particular idealized AAA model both numerically using FEA²³ and experimentally using the photoelastic method.²¹ The resulting von Mises stress distribution on the surface of the ideal AAA at a mean experimental pressure of 364.5 mmHg, along with the image captured at the point of rupture, is shown in Figure 11. Applying the same mean internal experimental pressure to the FEA model (364.5 mmHg) returned a peak stress of 1.16 MPa located at the iliac bifurcation. The figure also shows the qualitative agreement in stress distributions at this pressure loading with the rupture location found experimentally. Inflection zones experience higher stress than regions of maximum diameter. Wall stress results can also be expressed as wall tension results, as the failure tension of the wall is believed to be a better indicator of a region’s susceptibility to failure,¹⁹ with peak wall tension = peak wall stress $\times$ wall thickness. The numerical results from the 5 ranges of pressures applied during the experimental tests resulted in varied wall stresses and wall tensions (Tables 4 and 5).

DISCUSSION

The focus of this study was to experimentally rupture idealized silicone AAA models and compare these locations to numerical predictions. There has been little reported on the in-vitro rupture of AAAs, with much focus on the computational analysis of these aneurysms.^{7–16} Morris et al.²¹ observed rupture locations at the inflection regions the using the photoelastic method. These rupture locations correlate with the sites of rupture experienced in our study.

Nine models were manufactured using the injection-molding technique,^{25,26} of which 5 were experimentally ruptured and 4 were used to assess wall thickness. Although silicone rubber does not behave identically to arterial tissue when subjected to large

Figure 8 ♦ Sequence of events leading to model rupture of Test 1. (A) The inflated model, (B) the initial point of rupture, (C) propagation of the failure zone, and (D) complete failure of the silicone model. A similar sequence was observed for all 4 models that ruptured at regions of inflection.

strains, it is currently the most suitable analogue and has been widely used in previous studies.^{25,26,30,31} Until a more suitable material is available, one that possibly incorporates the layered structure of the arterial wall and the presence of a fibrinous network (collagen and elastin), silicone rubber will remain the most feasible arterial mimic.

Material Characterization

During tear testing, the tear in each sample was smooth, possibly indicating that the bond within the PDMS chains was not strong enough to prevent rupture under an applied force.³² Tensile testing documented a UTS that correlated well with the manufacturer's specifications, and numerical replication of

Figure 9 ♦ (A) Key regions and (B) wall thickness measurements of the ideal AAA model. Data points are individual measurements. Upper and lower bars represent 25th and 75th percentiles. Middle bar is mean wall thickness.

TABLE 3
Average Regional Variations in Wall Thickness

Model	Wall Thickness Variations by AAA Region, mm				
	1	2	3	4	5
Uniform 2 mm	2	2	2	2	2
25th Percentile	1.94	1.89	1.63	1.7	1.55
75th Percentile	2.16	2.29	1.87	2.18	2.17
Min thickness	1.72	1.65	1.34	1.26	1.17
Max thickness	2.49	2.51	2.04	2.44	2.53
Mean thickness	2.03	2.08	1.78	1.91	1.87
Thick inflection	2	2.5	1.5	2.5	2
Thin inflection	2	1.5	2.5	1.5	2

Thickness values were applied to numerical models to determine the effect of wall thickness on stress results. AAA regions correspond to Figure 9.

the type 2 dumbbell sample used for the tensile tests agreed well with the experimental data. The 3rd order Ogden model was selected as the optimum strain-energy function for this particular material; it allowed the material to remain stable at all stresses and strains within the ranges presented. To numerically model the actual rupture of the silicone models, a different approach to the traditional continuum mechanics method could be used. This could take into account the supersonic rupture of rubber³³ and may provide a further insight into the rupture behavior of these models. For the purpose of this article, we are interested in rupture regions of the models, not the numerical quantities per se.

Rupture Modeling

The use of a high-speed camera to record the point of rupture proved to be a very powerful experimental tool. As image quality was of paramount concern with this work, determining the optimum resolution and frame rate was crucial to the success of the testing. The mirror-wall arrangement also proved to be very effective for allowing all 360° of the model to be viewed. Upon pressurization, the AAA models did not inflate uniformly, resulting in asymmetry, possibly due to localized variations in wall thickness. However, it was shown that variations in wall thickness do not significantly affect the overall stress distributions, with the exception of the “min” thickness model. Rupture pres-

ures varied significantly for 2 of the 5 models. Four models experienced rupture at an inflection region. Elevated stresses at regions of inflection have been previously reported in experimental models, using both the photoelastic method²¹ and rosette strain gauge arrangements.²⁴ Flora et al.²⁴ observed stresses at the inflection region nearly twice those experienced at the region of maximum diameter. The one model that ruptured at the iliac bifurcation was unusual; in a clinical setting, real AAAs rarely rupture at the iliac bifurcation. Rather, rupture is typically at the posterior wall,⁵ so the iliac bifurcation region of our model may have had a localized flaw.

Wall Thickness

The mean wall thickness of the models was comparable to the 2-mm original design of the aluminum molds, so it was acceptable to use a uniform 2-mm wall to numerically represent the ideal AAA. The statistical method to determine the 25th and 75th percentiles of the wall thickness allowed us to create realistic wall thickness variations in the numerical AAA models. The maximum and minimum wall thickness values were also used to create “worst case scenario” AAA models with quite large variations in wall thickness. Also, two models were generated that had extreme changes in wall thickness (referred to as Thick and Thin Inflection in Table 3). Computations of the wall stress revealed that the inflection regions, particularly the proximal inflection zone, experi-

Figure 10 ♦ Effect of varying wall thickness on wall stress results. X-axis refers to the longitudinal distance from above the AAA sac to immediately above the iliac bifurcation. The line on the AAA model (between the brackets) indicates the path of stress results. The “25th” and “75th” refer to models created using the 25th and 75th percentiles of the wall thickness results; min, max, and mean refer to models created using the corresponding wall thickness values recorded.

Figure 11 ♦ In-vitro rupture locations and FEA stress patterns. High stress occurs at inflection points that correlate with rupture locations in experimental models. The undeformed image shows stress contours displayed on the AAA model prior to loading, with the deformed illustration showing the stress contours displayed on the model when an internal loading of 364.5 mmHg is applied. There is a noticeable change in geometry with increased internal pressure, particularly at the inflection regions.

enced higher stress than the region of maximum diameter, regardless of wall thickness variations.

Numerical Modeling

The peak wall stress located immediately prior to the iliac bifurcation was much lower than the UTS of the material, so one would not expect rupture of the model. Peak stresses at the iliac bifurcation are common in numerical computations of AAA wall stress. It is known that stress artifacts arise due to the inclusion of this bifurcation, and a method of evaluating and refining this localized stress region has yet to be developed.³⁴ The peak stress obtained numerically was on average only 13.8% of the UTS, with the stress at the inflection regions even less (8.4%). The manufacturers state that UTS values can have a very large range, so the stated 7.1-MPa UTS can actually be as low as 3.5 MPa.

Wall tension in these models exceeded the tear strength of the material, which may suggest that the model would experience rupture in all cases examined, if indeed tear strength is to be considered a possible indicator of material failure strength. At elevated pressures, silicone AAA models may rupture due to the pressure applied to the inner surface, causing an increase in wall tension and propagation of a nick or tear, ultimately leading to complete failure of the silicone wall. In realistic situations, calcified deposits embedded in the AAA wall may raise stress and could act as rupture initiators. Ruptures may occur at these areas, in particular if calcifications occur at regions of inflection. Analogue models to replicate calcified deposits can be included in future rubber models using available techniques at our laboratory.

Good agreement was observed when comparing the experimental rupture locations with the high stress regions found using FEA. It is believed that the geometry of the idealized AAA model may play a role in rupture behavior. The 3D nature of the model, along with slight deviations in wall thickness,^{25,26} may increase the likelihood of rupture. Another interesting point to note is that when the experimental model is inflated,

TABLE 4
Summary of FEA Wall Stress Results

Rupture Test	Pressure, mmHg	$\sigma_{\max}$, MPa	$\sigma_{\text{inflection}}$, MPa	$\sigma_{\max}$, % of UTS	$\sigma_{\text{inflection}}$, % of UTS
1	254.7	0.6808	0.4156	9	5
2	278.6	0.7375	0.4586	10	6
3	466.2	1.337	0.8668	17	11
4	278.7	0.7375	0.4586	10	6
5	544.6	1.733	1.1033	23	14

σ : stress.

UTS: ultimate tensile strength (7.7 MPa) of the material as determined from tensile testing.

the radius of curvature of the inflection region changes, while the numerical model does not experience such a significant change. This may also contribute to the difference in the wall stress results.

Also, it is known that no silicone model is absolutely free from defects and imperfections. The manufacturing process results in a silicone model that may contain microscopic flaws. Tensile testing determined the UTS of the material using planar dog-bone shapes, whereas the AAA model has regions of inflection, along with a bifurcation branching into the iliac arteries. These factors may all contribute to the propensity of the silicone model to rupture, and hence the FEA results would differ somewhat. It has also been reported³¹ using the Law of Laplace that actual AAAs can withstand up to 982 mmHg before rupture, with much variability between intrasac rupture pressures.³⁵

Our group has begun work on the rupture of realistic AAA models and the correlation with both FEA-predicted and photoelastic-predicted high stress regions. Also, our group has developed a successful method of creating silicone AAA models complete with a region of intraluminal thrombus (ILT) and

representative calcified deposits. Numerical modeling has shown that thrombus reduces wall stress,^{13,15,36} but it has a negligible effect on wall pressures in vivo,³⁷ so there is also a need to examine the role of ILT on in-vitro rupture tests. Calcifications can affect numerically computed wall stress and also decrease the biomechanical stability of the AAA,^{38,39} so they should be included in future work.

Limitations

A more suitable AAA wall analogue could be employed. Use of silicones with lower tensile strengths would allow lower, more realistic pressures to cause rupture. Work has begun in our group on the use of more suitable arterial vessel analogues.⁴⁰ Also, the use of a realistic cardiac pressure wave instead of static air pressure may alter the results. These factors are to be examined in future work.

Conclusion

According to the experimental and numerical findings presented here, AAAs experience higher stresses at regions of inflection com-

TABLE 5
Summary of FEA Wall Tension Results

Rupture Test	Pressure, mmHg	$\Omega_{\max}$, N/mm	$\Omega_{\text{inflection}}$, N/mm
1	254.7	1.3616	0.8312
2	278.6	1.475	0.9172
3	466.2	2.674	1.7336
4	278.7	1.475	0.9172
5	544.6	3.466	2.2066

Ω : wall tension.

Average tear strength = 0.419 N/mm.

pared to regions of maximum diameter. Ruptures of the idealized silicone models occurred predominantly at the inflection points, as numerically predicted. The use of a high speed camera is a useful experimental tool in observing rubber AAA rupture locations. To improve the method described, more suitable arterial analogues that mimic arterial properties more closely are required, thus possibly leading to an improved understanding of AAA rupture. Regions of inflection can be easily identified from basic 3D reconstruction, and as ruptures appear to occur at inflection points, these findings, along with future work, may provide a useful insight into the clinical significance of inflection regions.

Acknowledgments: The authors would like to thank (1) the Irish Research Council for Science, Engineering and Technology (IRCSET) for grant RS/2005/340; (2) the US National Heart Lung and Blood Institute for grant #R01-HL-060670; (3) Dr. Liam Morris from the Galway Medical Technology Centre, Galway, Ireland; (4) Kevin O'Flanagan from the Department of Manufacturing and Operations Engineering, University of Limerick, Ireland, for his assistance with the high-speed imaging; (5) Maria Ryan for her help with the experimental rupture modeling; (6) the Department of Vascular Surgery in the Midwestern Regional Hospital, Ireland, in particular, Mr. Eamon Kavanagh, Mr. Paul Burke, and Prof. Pierce Grace; and (7) Michel S. Makaroun, MD, Department of Surgery, University of Pittsburgh, Pennsylvania, USA.

APPENDIX I Sample Preparation

1. Clean the molds thoroughly using acetone.
2. Bolt the molds tightly together and seal the edges to prevent leaks.
3. Weigh appropriate amounts of components, i.e., silicone and catalyst (10:1 ratio), depending on the quantity of silicone material desired.
4. Combine the components in a mixing container and thoroughly mix for 3 minutes.
5. Degas the mixture at a pressure of 100 mBar for 15 minutes.
6. Remove from vacuum and suck mixture into a 60-mL syringe.

7. Slowly inject the mixture into the mold, ensuring no leaks.
8. Place the mold into vacuum to further degas for 15 minutes at a pressure of 350 mBar.
9. Remove and cure in oven for 18 hours at 45°C.
10. Upon curing, remove the samples from the mold and leave at room temperature for 3 hours (BS ISO 37).

APPENDIX II Tensile Testing

1. Set up the tensile test machine apparatus as recommended by the manufacturer.
2. Measure the thickness and width of each sample before testing.
3. Place the specimen in clamps and ensure proper location, i.e., the specimen is clamped in the same position each time.
4. Set up the software with the pre-determined loading and failure criteria.
 - a. *Dimensions* = Input thickness and width of sample (mm)
 - b. *Extension rate* = 500 mm/min (BS ISO 37)
 - c. *Preconditioning* = 10 cycles at 20% extension
 - d. *End criteria* = Stretch sample to failure
5. On failure of sample, the machine resets to pre-determined setting.
6. Software automatically generates UTS (MPa) and other user-defined variables.
7. Record data such as UTS and repeat test for the next sample.

APPENDIX III Tear Testing

1. Set up the tensile machine apparatus as recommended by the manufacturer.
2. Measure the thickness of each sample before testing.
3. Place the specimen in clamps and ensure proper location, i.e., the specimen

is clamped in the same position each time.

4. Set up the software with the pre-determined loading and failure criteria.
 - a. *Dimensions* = Input thickness of sample (mm)
 - b. *Extension rate* = 100 mm/min (BS ISO 34-1)
5. On failure of sample, the machine resets to pre-determined setting.
6. Software automatically generates tear strength (N/mm) and other user-defined variables.
7. Record data and repeat the test for the next sample.

REFERENCES

1. Sakalihasan N, Limet R, Defawe OD. Abdominal aortic aneurysm. *Lancet*. 2005;365:1577–1589.
2. Vande Geest JP, Sacks MS, Vorp DA. A planar biaxial constitutive relation for the luminal layer of intra-luminal thrombus in abdominal aortic aneurysms. *J Biomech*. 2006;39:2347–2354.
3. Kleinstreuer C, Li Z. Analysis and computer program for rupture-risk prediction of abdominal aortic aneurysms. *Biomed Eng Online*. 2006;5:19.
4. Nicholls SC, Gardner JB, Meissner MH, et al. Rupture in small abdominal aortic aneurysms. *J Vasc Surg*. 1998;28:884–888.
5. Darling RC, Messina CR, Brewster DC, et al. Autopsy study of unoperated abdominal aortic aneurysms. The case for early resection. *Circulation*. 1977;56:161–164.
6. Vorp DA. Biomechanics of abdominal aortic aneurysm. *J Biomech*. 2007;40:1887–1902.
7. Raghavan ML, Vorp DA, Federle MP, et al. Wall stress distribution on three-dimensionally reconstructed models of human abdominal aortic aneurysm. *J Vasc Surg*. 2000;31:760–769.
8. Venkatasubramaniam AK, Fagan MJ, Mehta T, et al. A comparative study of aortic wall stress using finite element analysis for ruptured and non-ruptured abdominal aortic aneurysms. *Eur J Vasc Endovasc Surg*. 2004;28:168–176.
9. Fillinger MF, Marra SP, Raghavan ML, et al. Prediction of rupture risk in abdominal aortic aneurysm during observation: wall stress versus diameter. *J Vasc Surg*. 2003;37:724–732.
10. Fillinger MF, Raghavan ML, Marra SP, et al. In vivo analysis of mechanical wall stress and abdominal aortic aneurysm rupture risk. *J Vasc Surg*. 2002;36:589–597.
11. Truijers M, Pol JA, SchultzeKool LJ, et al. Wall stress analysis in small asymptomatic, symptomatic and ruptured abdominal aortic aneurysms. *Eur J Vasc Endovasc Surg*. 2007;33:401–407.
12. Mower WR, Quinones WJ, Gambhir SS. Effect of intraluminal thrombus on abdominal aortic aneurysm wall stress. *J Vasc Surg*. 1997;26:602–608.
13. Doyle BJ, Callanan A, McGloughlin TM. A comparison of modelling techniques for computing wall stress in abdominal aortic aneurysms. *Biomed Eng Online*. 2007;6:38.
14. Rodriguez JF, Ruiz C, Doblare M, et al. Mechanical stresses in abdominal aortic aneurysms: influence of diameter, asymmetry, and material anisotropy. *J Biomech Eng*. 2008;130:021023.
15. Doyle BJ, Callanan A, Walsh MT, et al. A finite element analysis rupture index (FEARI) as an additional tool for abdominal aortic aneurysm burst prediction. *Vasc Dis Prev*. 2009;6:114–121.
16. Doyle BJ, Callanan A, Burke PE, et al. Vessel asymmetry as an additional diagnostic tool in the assessment of abdominal aortic aneurysms. *J Vasc Surg*. 2009;49:443–454.
17. Thubrikar MJ, Labrosse M, Robicsek F, et al. Mechanical properties of abdominal aortic aneurysm wall. *J Med Eng Tech*. 2001;25:133–142.
18. Raghavan ML, Webster MW, Vorp DA. Ex vivo biomechanical behaviour of abdominal aortic aneurysm: assessment using a new mathematical model. *Ann Biomed Eng*. 1996;24:573–582.
19. Raghavan ML, Kratzberg J, de Tolosa EM, et al. Regional distribution of wall thickness and failure properties of human abdominal aortic aneurysm. *J Biomech*. 2006;39:3010–3016.
20. Vande Geest JP, Wang DH, Wisniewski SR, et al. Towards a non-invasive method for determination of patient-specific wall strength distribution in abdominal aortic aneurysms. *Ann Biomed Eng*. 2006;34:1098–1106.
21. Morris L, O'Donnell P, Delassus P, et al. Experimental assessment of stress patterns in abdominal aortic aneurysms using the photoelastic method. *Strain*. 2005;40:165–172.
22. Laheij R, van Marrewijk C, Buth J. Progress report on the procedural and follow up results of 3413 patients who received stent graft treatment for infrarenal aortic aneurysms for a period of 6 years. EUROSTAR Data Registry Centre, 2001.

23. Callanan A, Morris LG, McGloughlin TM. Numerical and experimental analysis of an idealised abdominal aortic aneurysm. Presented at: European Society of Biomechanics. S-Hertogenbosch, Netherlands; July 4–7, 2004.
24. Flora HS, Talie-Faz B, Ansdell L, et al. Aneurysm wall stress and tendency to rupture are features of physical wall properties: an experimental study. *J Endovasc Ther.* 2002;9:665–675.
25. Doyle BJ, Morris LG, Callanan A, et al. 3D reconstruction and manufacture of real abdominal aortic aneurysms: from CT scan to silicone model. *J Biomech Eng.* 2008;130:034501.
26. O'Brien T, Morris L, O'Donnell M, et al. Injection-moulded models of major and minor arteries: the variability of model wall thickness owing to casting technique. *Proceedings of the International Mechanical Engineers, Part H: Journal of Engineering in Medicine.* 2005;219.
27. Mullins L. Softening of rubber by deformation. *Rubber Chem Tech.* 1969;42:339–362.
28. Brown R. *Physical Testing of Rubber*, 4th ed. New York: Springer Science+Business+Media, Inc.; 2006:159–167.
29. Vorp DA, Raghavan ML, Webster MW. Mechanical wall stress in abdominal aortic aneurysm: influence of diameter and asymmetry. *J Vasc Surg.* 1998;27:632–639.
30. Marins PA, Natal Jorge RM, Ferreira AJ. A Comparative study of several material models for prediction of hyperelastic properties: application to silicone-rubber and soft tissues. *Strain.* 2006;42:135–147.
31. Shergold OA, Fleck NA, Radford D. The uniaxial stress versus strain response of pig skin and silicone rubber at low and high strain rates. *Int J Imp Eng.* 2006;32:1384–1402.
32. Aziz T, Waters M, Jagger R. Analysis of the properties of silicone rubber maxillofacial prosthetic materials. *J Dent.* 2003;31:67–74.
33. Marder M. Supersonic rupture of rubber. *J Mech Phys Solids.* 2006;54:491–532.
34. Fillinger M. The long-term relationship of wall stress to the natural history of abdominal aortic aneurysms (finite element analysis and other methods). *Ann NY Acad Sci.* 2006;1085:22–28.
35. Vallabhaneni SR, Gilling-Smith GL, How TV, et al. Heterogeneity of tensile strength and matrix metalloproteinase activity in the wall of abdominal aortic aneurysms. *J Endovasc Ther.* 2004;11:494–502.
36. Wang DH, Makaroun MS, Webster MW, et al. Effect of intraluminal thrombus on wall stress in patient-specific models of abdominal aortic aneurysm. *J Vasc Surg.* 2002;36:598–604.
37. Schurink GW, van Baalen JM, Visser MJ, et al. Thrombus within an aortic aneurysm does not reduce pressure on the aneurysmal wall. *J Vasc Surg.* 2000;31:501–506.
38. Speelman L, Bohra A, Bosboom EM, et al. Effects of wall calcifications in patient-specific wall stress analyses of abdominal aortic aneurysms. *J Biomech Eng.* 2007;129:105–109.
39. Li Z-Y, U-King-Im J, Tang TY, et al. Impact of calcification and intraluminal thrombus on the computed wall stresses of abdominal aortic aneurysm. *J Vasc Surg.* 2008;47:928–935.
40. Doyle BJ, Corbett TJ, O'Donnell MR, et al. Design and development of a range of silicone elastomers for use in experimental studies. *Proceedings of the Fifteenth Annual Meeting of the Section of Bioengineering of the Royal Academy of Medicine in Ireland.* 2009;112.