Computational analysis of endovascular aortic repair proximal seal zone preservation with endoanchors: A case study in cylindrical neck anatomy

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ABSTRACT

Background: Endovascular aortic repair is the common approach for abdominal aortic aneurysms, but endoleaks remain a significant problem with long-term success. Endoanchors have been found to reduce the incidence of type 1A endoleaks and can treat intraoperative type 1a endoleaks. However, little is known about the optimal number and position of endoanchors to achieve the best outcome.

Methods: Using image segmentation and a computational model derived from a reconstructed native patient abdominal aortic aneurysm geometry, the stability of the proximal seal zone was examined through finite element analysis in Abaqus (Dassault Systèmes, Providence, RI). The biomechanical parameter of contact area was compared for varying numbers (0, 2, 4, 8) and positions (proximal, medial, distal) of endoanchors under different adhesion strengths and physiologic pressure conditions.

Results: In every simulation, an increase in adhesion strength is associated with maintenance of proximal seal. For biologically plausible adhesion strengths, under conditions of normal blood pressure (120 mm Hg), the addition of any number of endoanchors increases the stability of the endograft-wall interface at the proximal seal zone by approximately 10% compared with no endoanchors. At hypertensive pressures (200 mm Hg), endoanchors increase the stability of the interface by 20% to 60% compared with no endoanchors. The positioning of endoanchors within the proximal seal zone has a greater effect at hypertensive pressures, with proximal positioning increasing stability by 15% compared with medial and distal positioning and 30% compared with no endoanchors.

Conclusions: Endoanchors improve fixation within the proximal seal zone particularly under conditions of high peak systolic pressure. Seal zone stabilization provides a mechanism through which endoanchor addition may translate into lower rates of type 1a endoleaks for patients. (JVS-Vascular Science 2021;2:170-8.)

Clinical Relevance: Endovascular aortic repairs are commonly used to treat abdominal aortic aneurysms. Type 1a endoleaks threaten the long-term durability of repairs. Endoanchors have been found to reduce the incidence of this complication. Herein, we examine parameters surrounding optimal endoanchor number and positioning to reduce endovascular aortic repair failure. The computational modeling allowed for testing of endoanchors in varied adhesion strength between the endograft and the aorta, as well as hemodynamic conditions to mimic normotension vs hypertension. The results of the finite element analysis suggest that the addition of any number of endoanchors in the proximal seal zone is beneficial, especially with hypertensive loading.

Keywords: Abdominal aortic aneurysm; EVAR; Endoanchor; Endoleak; Finite element analysis

Endovascular aortic repair (EVAR) is currently the predominant treatment strategy for infrarenal abdominal aortic aneurysms (AAA).1 After EVAR, device-related complications occur in 10% to 40% of patients.2,3 The correct placement of an appropriately sized endograft in the proximal seal zone is an important determinant of EVAR outcomes.3 Type 1a endoleaks with stent migration or aortic dilation lead to “de-adhesion” of the endograft in the proximal seal zone.3 Endoanchors result in increasing pressure on the residual aneurysm sac, typically requiring reintervention to avoid rupture.6 Arterial hypertension, a risk factor for AAA development, has been associated with conversion from endovascular to open surgical repair and has been implicated in endoleak development.2,8 Endoleaks may result in continued aneurysmal degeneration, neck dilation, and pulsality of the sac, indicating the importance of medical therapy to decrease the risk of device failure.8

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A paradigm shift is underway concerning endoleaks, with the emphasis on the vulnerable proximal seal zone rather than on collateral vessels. For open procedures, continuous mechanical fastening at the suture line allows for kinematic coupling, or a near perfect seal zone. However, with EVAR, the ability of the endograft to remain in place is dependent on fixation and seal. Fixation is the resistance to longitudinal displacement and seal is defined as the interfacial contact stability between the graft and the aortic wall. Various clinical factors that may influence the proximal neck fixation are aortic roughness, blood pressure, drag forces, device properties, angulation, and oversizing. At the proximal seal zone, the interfacial contact between the graft and the aorta controls the stability of the repair. Interfacial fragility or toughness can be approximated by the adhesion strength between the surfaces. In an attempt to address the rate of type 1 endoleaks, endoanchors were developed to secure the proximal endograft to the aortic wall. The addition of any number of endoanchors improves proximal seal, translating to reduced endoleak risk. Endoanchors have a greater benefit under biomechanical conditions of high peak systolic pressure. Proximal positioning of endoanchors within the proximal seal zone increases stability compared with other positions.

**Take Home Message:** Endoanchors provide improved seal zone stability under appropriate graft aortic adhesive strength in cylindrical neck anatomy.

**ARTICLE HIGHLIGHTS**

- **Type of Research:** Single institution retrospective analysis
- **Key Findings:** Using a computational approach, the addition of any number of endoanchors improves proximal seal, translating to reduced endoleak risk. Endoanchors have a greater benefit under biomechanical conditions of high peak systolic pressure. Proximal positioning of endoanchors within the proximal seal zone increases stability compared with other positions.

**Finite element analysis**

**Mesh, geometry, and constitutive relations.** The mesh is imported into the commercial FEA code Abaqus (Dassault Systèmes, Providence, RI). The dynamic explicit solver in the mm-kg-ms system of consistent units is used for the simulations. The polynomial hyperelastic model of Raghavan and Vorp is used for the aorta: mass density = 1.12e-6 kg/mm^3, C10 = 0 GPa, C01 = 0.000174, C20 = 0, C11 = 0, C02 = 0.001881, D1 = 117.
D21 = 0; and ILT: mass density = 1.12e-6 kg/mm³. C10 = 2.6e-5, C01 = 0, C20 = 2.6e-5, C11 = 0, C02 = 0, D1 = 1900, D2 = 0. The endograft and endoanchors are modeled as neo-Hookean hyperelastic materials: mass density = 6e-6 kg/mm³, C10 = 0.03 GPa, D1 = 1.6. The endoanchors are kinematically coupled to both the aortic wall and stent; kinematic coupling imposes the constraint that the nodes on a given endoanchor initially in contact with the wall or stent remain tied throughout the simulation, coupling their displacements. This mimics the suture-like
mechanism of endoanchors that secures the stent to the aortic wall at the points of insertion. The endograft placed in the patient was an Endurant IIS (Medtronic, Minneapolis, Minn). Our simulations do not model this particular device or its structure, nor do we model the suprarenal fixation present in this device. \textit{Cohesive zone model (CZM):} while endoanchors are kinematically coupled to the wall and stent, the interface between the aorta and the graft is modeled using CZM.\textsuperscript{20–22} Effectively modeling seal zone stability, including ultimately interfacial failure, is important to our model, as the problem is highly nonlinear.\textsuperscript{23} CZM is employed as it is a widely used numerical technique for studying interfacial fracture. In CZM, the interaction between the aorta and the graft is studied by relating the cohesive forces and the separations between them using a spring-like behavior. Specifically, the cohesive force (and consequently the interfacial stress, $\sigma$) is linearly related to the separation ($d$) as $\sigma = Kd$, where $K = 5$ kN/mm$^2$ is the cohesive stiffness based on the above-mentioned properties describing the attachment of the two surfaces before deathesia. When $\sigma$ reaches a critical value $\sigma_{\text{max}}$, the spring fails and the aorta and graft surfaces can separate, with corresponding adhesion energy that corresponds to an adhesion energy $G = \sigma_{\text{max}} d = Kd^2 = (\sigma_{\text{max}} F/F_{\text{c}})$.\textsuperscript{21,23} There are currently no direct experimental data on the adhesion energy, $G$, or the maximum interfacial stress, $\sigma_{\text{max}}$, for the wall-stent interface.\textsuperscript{22,23} To complete our model, we chose values for the interfacial adhesion strengths within a reasonable range based on several limits. First, the limiting strength of any interface between two bulk materials of different elastic moduli cannot exceed the elastic modulus of the weaker material. Therefore, the upper bound for $\sigma_{\text{max}}$ is the stress at which rupture of the aortic wall would occur (0.5 MPa).\textsuperscript{10,24,25}

In our model, the proximal seal zone area is $A \approx 1000$ mm$^2$ (diameter: 25 mm, length: 15 mm). Published studies show that a downward force of 10 N applied to the end of an endograft is sufficient to initiate graft migration.\textsuperscript{26} Using dimensional analysis, we define a middle range for $\sigma_{\text{max}}$: $\sigma_{\text{max}} \approx (F/A) \approx 0.01$ MPa. This is approximately 50 times smaller than the value where wall rupture is considered to occur.\textsuperscript{23–25} Stent migration should occur at a lower stress than required for wall rupture. We include various strength parameters for cohesive behavior in the interface as contact 1 (C1) through contact 5 (C5), with C1 being the lowest stress and C5 being the greatest stress; C1: $\sigma = 0.05$ MPa (weak); C2: $\sigma = 0.1$ MPa (moderately weak); C3: $\sigma = 0.25$ MPa (moderate); C4: $\sigma = 0.5$ MPa (moderately strong); C5: $\sigma = 5.0$ MPa (strong). The value for adhesion strength can be inferred to be within this range because clinically, it is clear that appropriately sized endografts do not fall out of the aorta nor rupture the aortic wall. We expect C3 and C4 to represent biological adhesion strengths.

**Loads and boundary conditions.** The nodes at the proximal and distal portion of the aortic wall part and at the distal portion of the endograft ($<$1% of total nodes) are fixed axially. Standard displacement, strain, and stress outputs are used, with two fields used to mimic peak systolic pressures of 120 and 200 mm Hg following our prior work.\textsuperscript{12} It is important to note that we do not directly model the effect of endograft oversizing. As outlined above, by dimensional analysis, the interfacial stress that enters our simulations is given as $\sigma_{\text{max}} \approx (F/A)$. To obtain the order of magnitude for $\sigma$, we set $F$ equal to published in vitro displacement forces.\textsuperscript{26} Oversizing places an additional load onto the seal zone, $\delta F$, making the interfacial stress $\sigma_{\text{max}} \approx (F + \delta F) / A$; however, $\delta F$ has remained poorly characterized in the literature.\textsuperscript{10} Therefore, we elect to simply study different magnitudes of interfacial stress and not relate this to the underlying forces, one of which will be degree of oversizing.

**RESULTS**

**Seal zone preservation with endoanchors.** Fig 3 demonstrates that as the AAA model is pressurized to normotensive and hypertensive pressures, an increase in the adhesion strength maintains proximal seal with an increase in pressure, with endoanchors further increasing stability. The contact area between the luminal surface of the aorta and the endograft is the measure used to define proximal seal zone stability over the pressurization of the model. There is a loss of seal at lower pressures for both weaker adhesion strengths and zero endoanchors compared with the models with stronger adhesion strengths or endoanchors. With the addition of endoanchors, there is a minimal difference between the seal zone stability vs no endoanchors for the weakest and strongest adhesions. For moderate adhesion, which is considered biologically plausible and the most clinically relevant scenario, the addition of any number of endoanchors increases the contact area stability with an increase in pressure. Because there was no difference in the stability of the interface between 2, 4, or 8 endoanchors, the data for endoanchor addition for each adhesion strength were summed to compare with no endoanchors. After normalizing for initial contact area, endoanchor addition had a small but positive increase in contact area maintenance of under 10%. The resulting data for the model with eight endoanchors with a moderate adhesion strength of C3 do not fit with the trend and could indicate that under certain conditions increasing endoanchor numbers destabilizes the interface (planned future work). With a pressure up to 200 mm Hg, seal is maintained for the weak and moderate adhesion strengths at low pressures but contacts 1 through 4 fail before the maximum pressure is reached. Therefore, increased pressure on the luminal surface of the aorta and the endograft is associated with proximal seal zone failure. The addition of any number of
Endoanchors increase the contact area stability compared with no endoanchors for moderate adhesion. At the highest pressures, we see that for biological adhesion strengths, the interface is 20% to 60% more stable with endoanchors compared to without endoanchors for any number of endoanchors. Endoanchors greatly improve fixation of the proximal seal zone for interfaces with biological toughness.

Effect of endoanchor positioning. Fig 5 demonstrates that the position of the endoanchors within the proximal seal zone affects seal zone stability at normotensive and hypertensive pressures with varying adhesion strength. Four endoanchors at proximal, medial, and distal positioning within the proximal seal zone generally maintain seal better than no endoanchors. At lower pressures and weak adhesion, the position of the endoanchors has a minimal effect on the stability of the interface. Similarly, at the strongest adhesion, the addition of endoanchors in any position has no impact on stability because the interface never fails. For biological adhesion strengths (C3 and C4), the position of endoanchors does have an effect.

Fig 3. Proximal seal zone loss with pressurization at varying adhesion strengths and number of endoanchors placed. The addition of any number of endoanchors increases the stability of the wall-stent interface, especially for biological adhesion strengths, contacts 3 and 4 in green and purple. Analysis with varying endoanchor number run at five adhesion strengths: contacts 1 through 5, from weakest to strongest (red, blue, green, purple, black). Raw and normalized contact area maintenance (mm²) vs peak systolic pressure (mm Hg), with a peak systolic pressure increase to 120 mm Hg in (I) and 200 mm Hg in (II). Contact loss for zero endoanchors indicated by bold lines and contact loss with endoanchors indicated with dashed lines (A, C). Normalization is endoanchors (averaged for all endoanchor numbers) compared with not having endoanchors, normalized by the initial contact area (B, D).
especially when pressure reaches 200 mm Hg. For C3 at hypertensive pressures, proximal endoanchor positioning does 15% better than medial and distal positioning and 30% better than no endoanchors. For C4 at hypertensive pressures, proximal and medial positioning do about the same and slightly better than distal, and approximately 60% better than no endoanchors.

**DISCUSSION**

To our knowledge, these results provide the first detailed biomechanical model enhanced proximal aortic seal zone stability when endoanchor technology is used. In this computational study of EVAR stability in a straight neck AAA anatomy, we find that adhesion strength dominates stability. In moderate adhesion, endoanchors augment the ability of the seal zone to withstand failure. Although there is no significant difference between the number of endoanchors placed and interfacial stability, the position of the endoanchors appears to play a role in seal maintenance. Our work shows that FEA is effective in studying the interaction of adhesion strength and endoanchors on seal and fixation.
We observed that increased adhesion strength is correlated with increased interfacial toughness between the luminal surface of the aorta and the endograft. Although we do not change the radius of the endograft in the models, changing $a_{\text{max}}$ in the simulations can be clinically understood as graft sizing. Weak adhesion would clinically represent an undersized graft or neck thrombus where there is a lack of true seal between the stent graft and the aorta. Moderate adhesion would represent the clinical scenario with an appropriately sized endograft for a given diameter with good apposition at the interface. Strong adhesion mimics an endograft being completely sutured into the aorta, as with an open AAA repair technique. Increasing adhesion strength at high pressure is especially important, as stronger interfacial toughness increases proximal seal zone stability as pressure reaches hypertensive levels. Adhesion strength stabilizes the proximal seal zone by increasing the seal, resisting delamination at the interface that would allow fluid to leak through, thereby preventing an endoleak.

For moderate adhesion strengths, which are biologically plausible and correlated with an appropriately sized endograft, the addition of endoanchors increases seal zone stability by resisting displacement between the graft and the aorta. The addition of endoanchors does not improve stability for the weakest or strongest adhesion strengths, indicating that endoanchors assist seal in an intermediate regime and have negligible impact when adhesion and seal are either too weak or perfectly adhered.

With the instructions for use indicating the placement of several endoanchors around the proximal seal zone depending on the diameter of the neck, we studied the impact of various endoanchor numbers on proximal seal zone stability. We found that there was no difference in the benefits of endoanchors regardless of the number placed. This trend applies to most conditions, excluding eight endoanchors for moderate adhesion, C3, at a pressure of 120 mm Hg.

We also discovered a gap in the research regarding ideal placement of endoanchors within the seal zone and sought to test the impact of proximal, medial, and distal positioning of endoanchors on seal zone stability. For the biological adhesion regime, as with an appropriately sized endograft, proximal positioning better withstood an increase in pressure compared with medial and distal positioning, indicating that placing endoanchors proximally in the seal zone may assist in preventing...
interfacial failure and therefore endoleaks. The knowledge gap concerning positioning exists because of the difficulty of knowing the precise location of an endoanchor during placement intraoperatively. Our computational model overcame this gap and reveals a novel insight for optimal anchor placement.

We found that addition of any number of endoanchors in our model provided greater stability under conditions of high peak systolic pressure, potentially preventing endoleak and allowing better aneurysm depressurization. These findings suggest that endoanchors may be beneficial in patients with uncontrolled or resistant hypertension. As pressure on the endograft and the luminal surface of the aorta proximal to the stent increases from normotensive to hypertensive levels, within the moderate adhesion regime, the wall-stent interface fails. Controlling for adhesion strength and number of endoanchors placed, there is a greater loss of contact between the aorta and the stent with an increase in pressure up to 200 mm Hg. With the addition of endoanchors, the proximal seal zone can better withstand the increase in pressure. For adhesion in the moderate regime (C3 and C4), the addition of endoanchors improves seal by 20% to 60% compared with no endoanchors at hypertensive levels. These data support the clinical literature that hypertension may be a risk factor for type 1 endoleaks.7

Although many studies indicate that the addition of endoanchors is associated with fewer endoleaks for short and hostile necks, we find that endoanchors improve seal for favorable neck geometries as well and should be considered for use in routine EVAR. Endoanchors play a role in preventing endoleaks and therefore may limit the risk of sac pressurization and potential rupture. Although endoanchors are costly, our research indicates that fewer endoanchors can be placed with a minimal difference in seal zone stability with increasing pressure. Therefore, instead of attempting to place up to eight endoanchors, which is technically difficult and time consuming, fewer can be placed with the same benefits. A study of explanted endografts found that suboptimal placement, such as with noncircular positioning, or deployment issues, such as the angle or depth of penetration, of endoanchors resulted in structural damage to the stent.27 This indicates that the technical aspects of endoanchor placement should be thoroughly considered before use.

The main limitation of our study is its single patient computational approach. The anatomy chosen was specifically selected to assess a cylindrical neck that is the classic “best scenario” for EVAR. We are not claiming that this work generalizes outside the scope of this specific anatomy. We chose to study the most common and simplest anatomy in particular because more hostile neck anatomy is even outside the scope of standard instructions for uses for EVAR. The computational process currently includes a brittle fracture model, but we believe that for more accurate biological modeling, we will need to incorporate a fully integrated modexing ductile fracture cohesive zone model with appropriate softening behavior in the traction separation law to more accurately analyze the aorta-endograft interface. This study uses adhesion strength as a surrogate for graft oversizing, as there is no current computational framework where an active adhesive interface and graft deployment can be studied in a controlled manner. Furthermore, even experimental data show that stent-graft behavior is highly nonlinear concerning the radial loads a graft exerts for a given degree of oversizing.28

Future experimental work is needed to measure the interfacial strength and toughness of aorta-endograft interfaces at different degrees of oversizing to provided precise input parameters for our models. Moreover, future work will incorporate patient-specific migration loads through serial imaging similar to recent publication on increased risks in endoleak development.29

Also, the aortic roughness is not considered in the material properties of the aortic wall in our model. The inclusion of calcium content is beyond the scope of this work but will be studied in future work. These improvements in the modeling process will improve the biological accuracy of our current adhesion model. Also, because this is mechanistic work analyzing the novel use of computation model to analyze the impact of endoanchors on a patient geometry, we aim to apply our improved adhesion model to different patient neck anatomies with nonlinear neck geometries to investigate endoanchor benefits for different patients. We hope to use a fluid-structure interaction model to simulate blood flow and therefore EVAR failure with endoleaks in our patient models. This would allow us to investigate the role of drag forces on endograft stability. In the future, we want to apply different loading conditions, including endograft oversizing that will involve development of a nontrivial growth model and neck dilation due to EVAR placement to understand the impacts of stent diameter and aneurysmal degeneration of the proximal neck.

CONCLUSIONS

Adhesion strength is a critical parameter in proximal seal zone stability. Endoanchors provide necessary fixation to maintain contact at the interface for endografts with moderate adhesion regardless of the number placed. This novel study investigated the biomechanical benefits of endoanchor fixation on seal zone stability using the biological consideration of high peak systolic pressures. Future considerations include studying different patient anatomies including angulated and short neck geometries, oversizing of the endograft, and dynamic fluid interactions.
**AUTHOR CONTRIBUTIONS**

Conception and design: EA, SD, KK, NN, TB, LP, RM
Analysis and interpretation: EA, SS, KC, LP, RM
Data collection: EA, LP
Writing the article: EA, LP, RM
Critical revision of the article: EA, SD, KK, SS, KC, NN, TB, LP
Final approval of the article: EA, SD, KK, SS, KC, NN, TB, LP, RM
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Overall responsibility: LP

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