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

Erin Abbott, BS,[a] Sanjeev Dhara, BS,[b] Kameel Khabaz,[a] Seth Sankary, MD,[c] Kathleen Cao, PhD,[c] Nhung Nguyen, PhD,[c] Trissa Babrowski, MD,[c] Luka Pocivavsek, MD, PhD,[c] *and* Ross Milner, MD,[c] *Chicago, Ill*

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.[4] Type 1a endoleaks with stent migration or aortic dilation lead to "de-adhesion" of the endograft in the proximal seal zone.[5] Endoleaks 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.[7,8] 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]

The College,[a] Pritzker School of Medicine,[b] and Section of Vascular Surgery and Endovascular Therapy, Department of Surgery,[c] University of Chicago, Chicago.

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Correspondence: Luka Pocivavsek, MD, PhD, Department of Surgery, University of Chicago, 5841 S Maryland Ave, Ste J-557C, Chicago, IL 60637 (e-mail: lpocivavsek@bsd.uchicago.edu).

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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.[8,9] 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.[10] 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.[11] At the proximal seal zone, the interfacial contact between the graft and the aorta controls the stability of the repair.[12] 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.[13] The helical screws are indicated for patients with hostile neck geometry,[14] that is, those with short, wide, conical, or hyperangulated necks, but routine EVAR procedure patients also benefit.[12,13] The addition of endoanchors at the proximal seal zone, as a preventative measure or treatment of type 1a endoleaks, has low adverse event rates and is successful.[15] A majority of patients with EVAR with endoanchors have greater sac regression compared with EVAR without endoanchors, indicating the positive effect of increased proximal seal zone stability.[16,17] Although there is growing clinical understanding of the effectiveness of endoanchors in EVAR, a biophysical understanding of the role of endoanchors in preventing device failure, and in appropriate patient selection, is needed for this adjunct endovascular therapy. Therefore, although many clinical factors may impact seal stability, hypertensive conditions are studied because of the known impact of fluid dynamics and drag on endograft stability. This paper studies the interaction between endoanchors and initial aortic-endograft adhesion strength in a single cylindrical seal zone geometry. The output of our computational study is seal zone stability measured as preservation of contact area between the endograft and aortic wall within the seal zone. The parameters varied include the number and location of endoanchors, adhesion strength, and luminal pressure. Seal zone geometry is kept constant and is taken from a single patient with a traditional cylindrical neck anatomy. There is no direct modeling of oversizing in our simulations. The study was accomplished with a computational model using image segmentation and finite element analysis (FEA).

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

METHODS

Geometry segmentation

The details of the segmentation method are outlined in our prior work.[12] Patient-specific geometries are extracted from deidentified three-dimensional computed tomography image data sets. The axial images are segmented using Simpleware ScanIP (Synopsys, Inc, Mountain View, Calif) with a semiautomated custom algorithm. Institutional review board approval was not required as no patient identifiers were used in the analysis. Disks are made for the internal aortic wall every three images, with linear interpolation to form a solid aneurysm part. The external aortic wall is approximated by dilating the part by 2 pixels. A shell for the aortic wall (approximately 3 mm thick) results from a Boolean subtraction operation with the internal solid part from the external solid part. The endograft and the intraluminal thrombus (ILT) are segmented into the model with the same method. The endograft geometry is modeled as a homogeneous cylindrical solid (Fig 1). The endoanchors are segmented as bolts placed circumferentially through the thickness of the endograft and aortic wall in the proximal seal zone. We generate four separate models with the same parts for the aortic wall, endograft, and ILT, and either 0, 2, 4, or 8 endoanchors are placed medially in the proximal seal zone (Fig 2). The model does not include the renal or visceral vessels and is cropped 7.5 cm distally from the celiac artery because our focus was the infrarenal neck seal zone. The cropped model is then meshed with C3D4 elements using the +FE Simpleware free volume algorithm with smart mask smoothing.

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[18,19] 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,

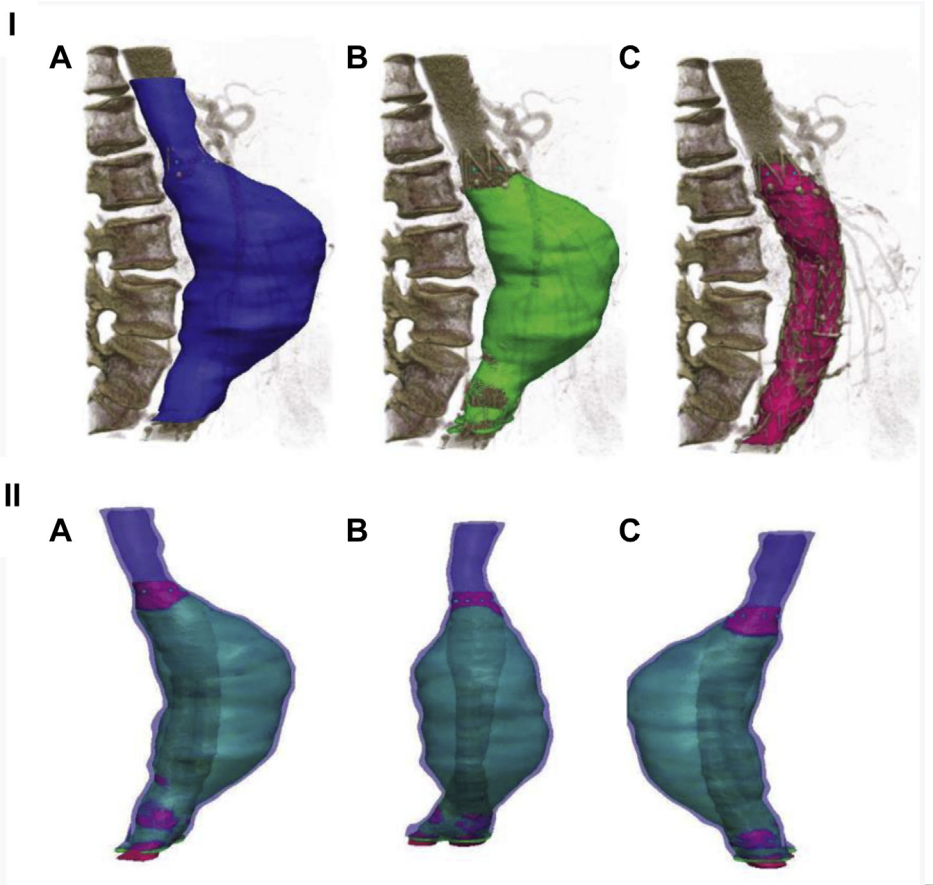

Fig 1. Segmented postoperative geometries for the study patient abdominal aortic aneurysm (AAA) with **(I)** individual components in the model for the aneurysm sac **(A)**, intraluminal thrombus (ILT; **B**), and endograft **(C)** overlaid on computed tomography (CT) images. **(II)** Multiple views of the complete model with endoanchor placement; in **(I)** and **(II)**, the three parts are constructed, meshed, and imported into Abaqus.

D21 = 0; and ILT: mass density = 1.12e-6 kg/mm^3, 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/mm^3, 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

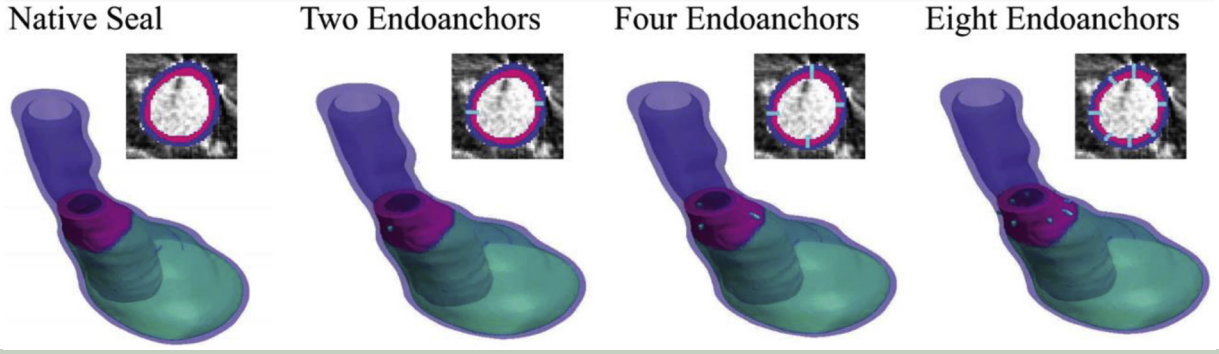

Fig 2. Segmented and cropped postoperative geometries for the study patient abdominal aortic aneurysm (AAA) with zero, two, four, or eight endoanchors added to the proximal seal zone of the model. The four parts, aortic wall (*purple*), ILT (*green*), endograft (*pink*), and endoanchors (*blue*), are constructed, meshed, and imported into Abaqus.

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. *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.[20-22] Effectively modeling seal zone stability, including ultimately interfacial failure, is important to our model, as the problem is highly nonlinear.[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, σ) is linearly related to the separation (δ) as $\sigma = K\delta$, where $K =$ 5 kN/mm^3 is the cohesive stiffness based on the above-mentioned properties describing the attachment of the two surfaces before deadhesion. When σ reaches a critical value σ_{max}, the spring fails and the aorta and graft surfaces can separate, with corresponding adhesion energy that corresponds to an adhesion energy $G = \sigma_{max}\delta = K\delta^2 = (\sigma_{max})^2/K$.[21-23] There are currently no direct experimental data on the adhesion energy, G, or the maximum interfacial stress, σ_{max}, for the wall-stent interface.[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 σ_{max} is the stress at which rupture of the aortic wall would occur (0.5 MPa).[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.[26] Using dimensional analysis, we define a middle range for σ_{max}: $\sigma_{max} \sim (F/A) \sim$ 0.01 MPa. This is approximately 50 times smaller than the value where wall rupture is considered to occur.[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.[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 $\sigma_{max} \sim (F/A)$. To obtain the order of magnitude for σ, we set F equal to published in vitro displacement forces.[26] Oversizing places an additional load onto the seal zone, δF_o, making the interfacial stress $\sigma_{max} \sim (F + \delta F_o/A)$; however, δF_o has remained poorly characterized in the literature.[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

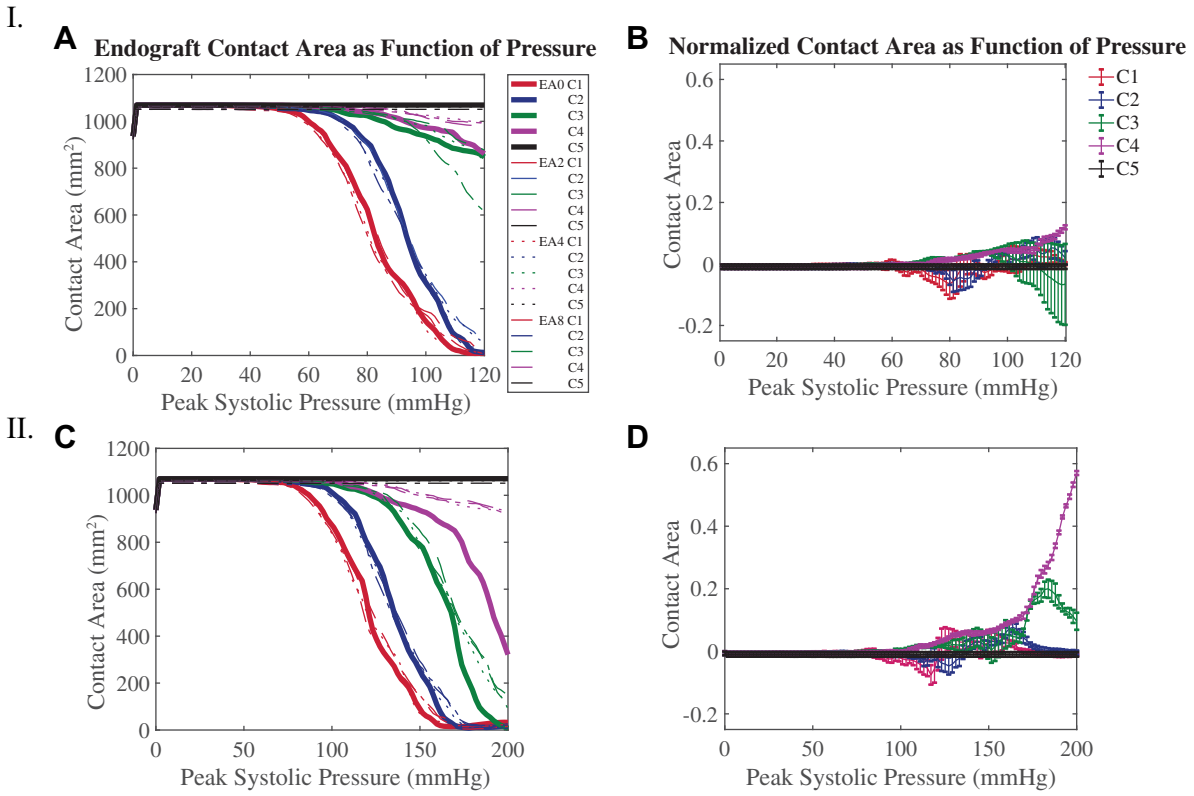

Fig 3. Proximal seal zone loss with pressurization at varying adhesion strengths and number of endoanchors placed. The addition of any number of endoanchors increases the stability of the wall-stent interface, especially for biological adhesion strengths, contacts 3 and 4 in *green* and *purple*. Analysis 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 (mm^2) 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)**.

endoanchors increases 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.

Fig 4 similarly displays that the endoanchor supported endograft extends the pressure that the interface with varying strength can withstand compared with the native endograft. Hoop strain around the neck of the aorta increases with an increase in pressure. Logarithmic strain increases as the interface fails as there is delamination between the aorta and the endograft. Without endoanchors, seal is lost at normotension only in the weakest adhesion but is lost at high peak systolic pressure at every adhesion strength except for the strongest, C5. With endoanchor support, the seal is maintained similarly for lower pressures, but with high peak systolic pressure, seal is maintained for biological adhesion with C4. Therefore, the addition of endoanchors extends the seal zone preservation at high pressures compared with no endoanchors.

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,

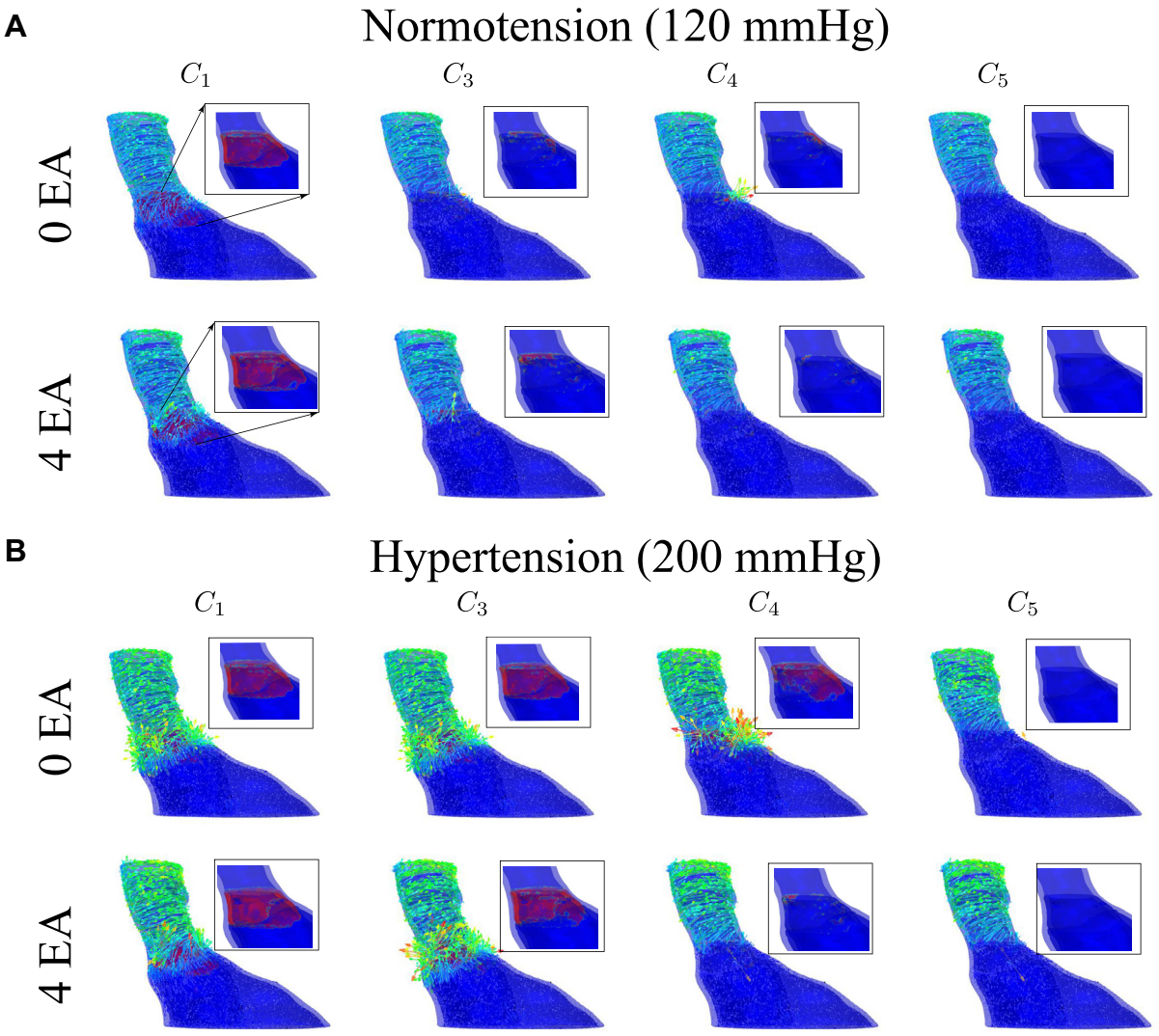

Fig 4. Contact loss and strain at normotensive and hypertensive peak systolic pressures with and without endoanchors. Seal is maintained at stronger adhesion strengths and in the presence of endoanchors at moderate adhesion strengths. Seal is lost, indicated by *red* in the proximal neck, for normotensive peak systolic pressure **(A)** and hypertensive peak systolic pressure **(B)**, and with the addition of zero or four endoanchors. The presence of *solid blue* in the proximal seal zone as opposed to *red* indicates that the aortic wall and the endograft are still in contact. The addition of endoanchors, especially for hypertensive peak systolic pressure in **(B)**, extends the seal zone stability compared with zero endoanchors. Logarithmic strain (*LE*) is displayed in *blue* and *green*, resulting in hoop strain in the proximal aorta and turning outward with more strength (*orange* and *red arrows*) where the proximal seal zone has failed.

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.

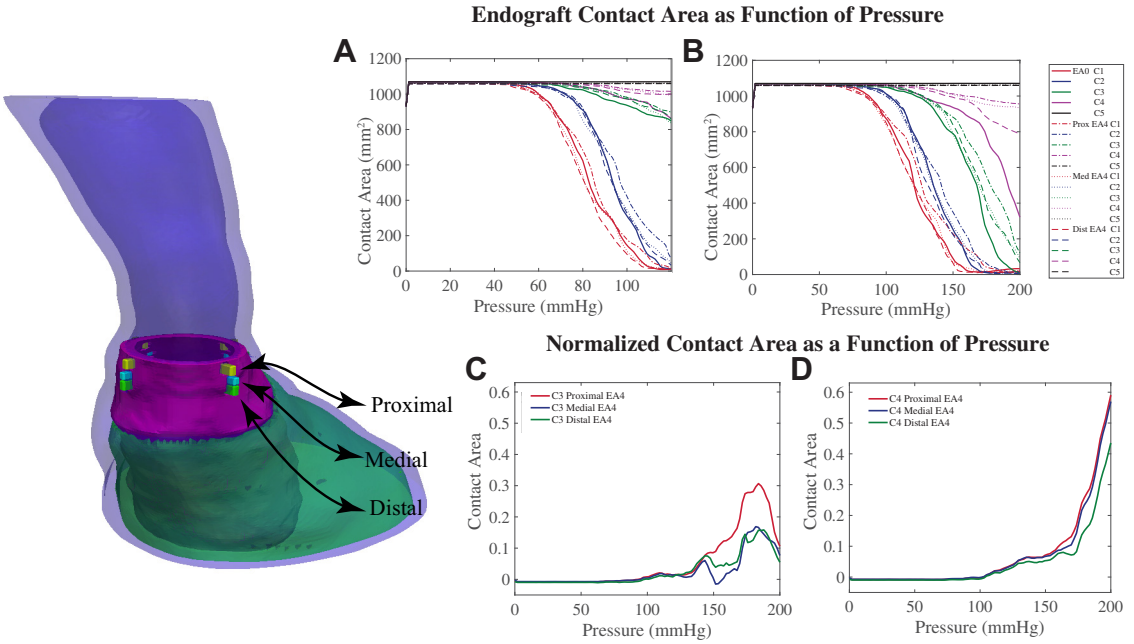

Fig 5. Proximal seal zone loss with pressurization at varying adhesion strengths and endoanchor positions. The position of four endoanchors proximally in the seal zone increased the stability of the wall-stent interface, especially for biological adhesion strengths. Analysis with varying endoanchor number run at five adhesion strengths: contacts 1 through 5, from weakest to strongest (*red, blue, green, purple, black*). Visualization of endoanchor placement in the model on the left. Raw and normalized contact area maintenance (mm^2) vs pressure (mm Hg), with an increase in pressure to 120 mm Hg in **(A)** and 200 mm Hg in **(B)**. Contact loss for zero endoanchors indicated by *bold lines* and contact loss with various positioning of endoanchors indicated with *dashed lines*. Normalization is endoanchor position compared with not having endoanchors, normalized by initial contact area for moderate adhesion, contact 3 **(C)** and contact 4 **(D)**.

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 σ_{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 noncircumferential 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 mode-mixing 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
Statistical analysis: Not applicable
Obtained funding: Not applicable
Overall responsibility: LP

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