

## Article

# Development and Mechanical Evaluation of a Stent Graft for Endovascular Aneurysm Repair Using Finite Element Modeling

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## Abstract

An abdominal aortic aneurysm (AAA) poses a significant risk of arterial wall rupture, which critically endangers the patient's life. To address this condition, an endovascular aneurysm repair (EVAR) is required, involving the insertion and expansion of a stent-graft within the aorta, to support and isolate the weakened vessel wall. In this context, this article aims to approach the problem from a mechanical perspective and to simulate the expansion and deployment procedure realistically, utilizing the Finite Element Analysis (FEA). The process initiates with the computation evaluation of the aortic structure in order to identify critical regions of stress and strain in an aneurysmatic aortic region. Then, a customized 3D-designed stent graft model was developed for the aorta and positioned properly. Applying all the necessary boundary conditions, a complex nonlinear FEA was conducted until the stent-graft expanded radially, reaching a final diameter 25% larger than the aorta's vessel wall while withstanding mean stress and strain values close to 400 MPa and 1.5%, respectively. Finally, the mechanical behavior of the stent-graft and its interaction with the internal aortic wall, during the expansion process, was evaluated, and the extracted results were analyzed.



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**Keywords:** abdominal aortic aneurysm; stent-graft; endovascular aneurysm repair; finite element analysis

## 1. Introduction

Abdominal aortic aneurysm (AAA) is a pathologic condition, usually located in the infrarenal aorta, with progressive abdominal aortic diameter dilatation of 3.0 cm or more [1]. It occurs when the abdominal aortic wall weakens, causing it to bulge or balloon, and poses a significant risk for a potentially fatal rupture. The primary risk factors that can contribute to the occurrence of an AAA are advanced age (65–75 years), male sex, and smoking habits. Other important factors associated with AAA development include a family history of abdominal aortic aneurysm, coronary artery disease, hypertension, peripheral artery disease, and previous myocardial infarction. All of these can be diagnosed by physical examination, an incidental finding on imaging, or ultrasonography [2]. For small abdominal aortic aneurysms (3–5 cm), no immediate intervention is typically needed, but periodic medical surveillance is recommended. A substantially increased risk for fatal aneurysmal rupture is mainly observed when the aneurysm exceeds the diameter of 5–5.5 cm [3]. Additional rupture indicators are an aneurysm size that is 2.5 times the normal aortic diameter, an aneurysm with an expansion rate that exceeds 1 cm per year, and a symptomatic aneurysm, such as one with accompanying back pain, a potential warning sign of impending rupture [4]. In all those cases,

an open surgery or an endovascular aneurysm repair (EVAR) is required. The standard has traditionally been the open surgical approach, but nowadays, EVAR is more common, faster, causes less stress on the body, and has a shorter recovery time.

EVAR is a minimally invasive endovascular procedure, and much safer than the open repair, which involves the placement of a bifurcated or tubular stent graft over the AAA to exclude the aneurysm from arterial circulation [5]. A stent-graft comprises a metallic (stainless steel or nitinol) skeleton covered with an impermeable (polytetrafluoroethylene or polyester) fabric and is implanted using fluoroscopic guidance through the femoral arteries. The placement of the endovascular device against the aorta is achieved by the radial expansion force of the stent graft on the aortic wall. Consequently, it leads to the isolation of the aneurysm from the main systemic aortic flow and aims to prevent a subsequent rupture [6].

Finite element analysis (FEA) has become an essential tool in the evaluation and design of stent-grafts for EVAR. Demanget et al. [7] investigated the mechanical performance of eight marketed stent-grafts under combined bending and pressurization, reporting significantly higher luminal reduction and peak stresses in Z-stented designs compared to spiral or circular configurations, which showed improved flexibility and durability. More recently, novel approaches have been proposed, such as the development of stent-graft limbs based on auxetic unit cells, which were shown through FEA to outperform traditional Z- and circular-stented devices by achieving low luminal reduction (4–12%), absence of kinking, and favorable stress–strain distribution [8]. De Bock et al. [9] advanced the field by creating a validated FEA model of bifurcated stent-graft deployment, accurately replicating in vitro experiments and confirming FEA's predictive capability for radial force, expansion, and neck dilation. Similarly, Mouktadiri et al. [10] applied a comprehensive patient-specific FEA simulation that accounted for guide-wire and catheter navigation through tortuous anatomies, highlighting the role of vessel heterogeneity in predicting stress concentrations during EVAR. In another contribution, Emendi et al. [11] validated patient-specific FEA models for predicting guide-wire-induced aortic deformations, showing high accuracy compared to experimental results and underscoring the method's clinical potential. Most recently, Corvo et al. [12] demonstrated that respiration-informed FEA can replicate stent-graft post-operative geometry with high spatial accuracy when compared to CT data, providing evidence that advanced computational modeling can support procedural planning and improve outcome prediction. Together, these studies highlight the evolution of stent-graft FEA research, from benchmarking commercial devices to validating patient-specific deployment simulations, yet they also indicate that challenges remain in capturing complex nonlinear wall–device interactions, which motivates the present work.

The objective of this study is to realistically simulate the expansion of a stent-graft within an aneurysmal aortic vessel wall, with the dual aim of demonstrating the efficiency of the EVAR procedure and identifying potential mechanical risks during deployment. To achieve this, a nonlinear finite element analysis (FEA) framework was developed to capture the complex mechanical behavior of the stent-graft and its interaction with the surrounding aortic tissue. Like many physical problems in science and engineering, this process can be described by partial differential equations that govern material deformation and contact mechanics [13]. FEA provides a powerful numerical tool to approximate these solutions and predict structural responses under physiological loading conditions [14]. The simulation procedure began with the definition of governing and boundary conditions, followed by the generation of a detailed three-dimensional aortic model and a customized stent-graft geometry. The geometry was discretized into a fine mesh of finite elements, each assigned appropriate material properties and interconnected by nodal points. Physical laws and constitutive equations were then applied to each element, assembled into a global system of equations, and solved iteratively to compute stresses, strains, and displacements

throughout the model. Importantly, two key nonlinearities were incorporated into the analysis: (i) the large deformation of the aneurysmal aortic wall and the expanding stent-graft, and (ii) the unilateral contact behavior governing their mechanical interaction. The novelty of this study lies in the simultaneous treatment of these nonlinearities within a realistic deployment framework, which distinguishes it from many prior FEA studies that often simplified vessel deformation or neglected detailed contact phenomena. The results highlight the spatial distributions of stress and strain across the stent-graft and aortic wall during expansion, offering novel insights into oversizing behavior and wall–device mechanics. Ultimately, this work demonstrates that advanced nonlinear FEA, when combined with realistic deployment conditions, is a powerful approach for assessing stent-graft performance and improving the design and safety of EVAR devices. Moreover, Figure 1 graphically illustrates the workflow of the current study.

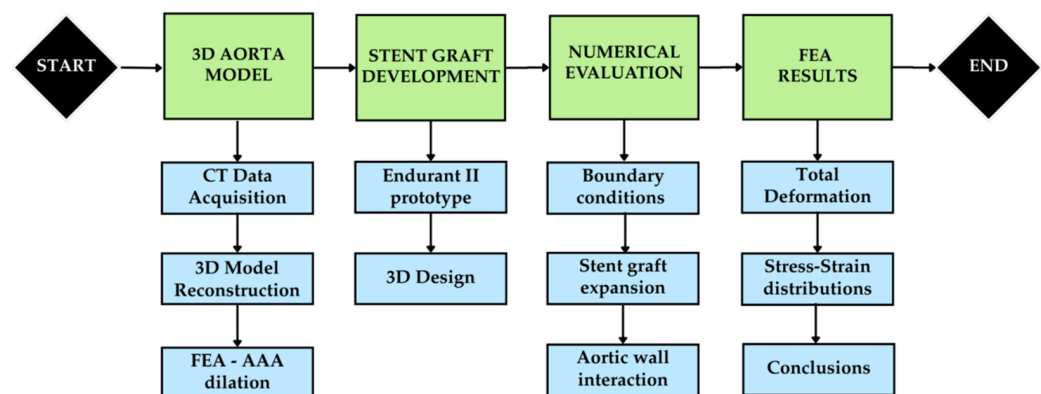


**Figure 1.** Flowchart of the study.

Regarding the structure of this study, in the Materials and Methods Section, the applied methodology and materials are presented, while in the Results Section, the FEAs’ results are analyzed, deriving the most crucial conclusions for this novel and patient-specific process.

## 2. Materials and Methods

### 2.1. 3D Reconstruction of the Abdominal Aortic Aneurysm Model

A key step in this study for preparing and conducting the simulation was the generation of an accurate three-dimensional model of an aorta with an abdominal aneurysm, derived from real clinical cases. For the purpose, a representative aortic anatomy was required. Based on the criteria outlined in a prior study [15], the T1-P6 aorta model—characterized by its suitability for stent graft deployment—was selected from a previous study. Therefore, a surface-based replica of this model was constructed in SolidWorks® Student Edition 2024 SP5.0, informed by the geometrical specifications of the stent provided in the cited work. This digital reconstruction served as a basis for further design and analysis related to the EVAR intervention [16].

The final reconstructed AAA model was later exported as an .stl file to Ansys Mechanical software (Ansys Student 2025 R2) in order to build the simulation environment. The 3D model was constrained with a Fixed Support at both proximal and distal ends and then subjected to internal pressure loads. More specifically, a hydrostatic pressure load derived from gravitational acceleration and blood’s fluid density of  $1060 \text{ kg/m}^3$ , along with an aortic pulse pressure of 44 mmHg, was applied to the inner surface of the aortic wall. The bounding box that describes the geometrical dimensions of the aorta is 82.588 mm in length across the X-axis, and 136.86 mm and 54.811 mm on the Y and Z axes, respectively. More details are shown below in Figure 2 and Table 1.



**Figure 2.** Visual 3D representation of Abdominal Aortic Aneurysm.

**Table 1.** AAA patient data and 3D model characteristics.

| Gender | Age | Aneurysm Diameter | Proximal Neck Diameter | Left Iliac Diameter | Right Iliac Diameter |
|--------|-----|-------------------|------------------------|---------------------|----------------------|
| Male   | 75  | 50 mm             | 15 mm                  | 10 mm               | 7.5 mm               |

## 2.2. Stent Graft 3D Design Process

The stent graft involved in this study’s FEA simulation was designed as a customized 3D model based on the Endurant II medical product, a real cardiovascular device manufactured by Medtronic Plc. The Endurant II stent graft is a new-generation device that is widely used in clinical practice, providing efficient and safe results for the treatment of abdominal aortic aneurysms. It is composed of two parts sewn together, a nitinol metal stent structure for external support and a multifilament polyester (PET—polyethylene terephthalate) graft fabric which provides resistance against type II endoleaks. The stent body is wire-formed in an M-shape that offers flexibility and circumferential conformability, contributing to an accurate and controlled deployment. [17]

The design and modeling process was carried out using Autodesk Fusion 360, a computer-aided design (CAD) software. For reducing the complexity and the computational cost of the simulation, the 3D stent graft model was designed as a simplified cylindrical curved tube with a uniform non-tapered diameter of 4.5 mm on both proximal and distal ends. The diameter value of 4.5 mm implies that the stent graft was designed in the compressed form (before deployment).

To reduce computational cost while still capturing the essential expansion mechanics, the stent graft was modeled as a uniform, unbranched cylindrical tube. This simplification implies that complex anatomical features such as tapering, bifurcation, and asymmetry of the iliac extensions were not included. While this geometry differs from the exact Endurant II clinical configuration, the cylindrical representation allows for an accurate evaluation of radial expansion, contact interaction, and stress–strain response, which are the primary focus of this study. Previous FEA investigations have also employed similar cylindrical approximations to reduce model complexity while maintaining predictive capability for local stress distribution and expansion mechanics [9,10,18].

The first step of the process was the creation of a circular base sketch, which was then extruded along a curved polyline following the aorta’s geometry. The created body represents the graft component. It was constructed with a diameter size of 4 mm, and its

multifilament fabric's thickness value was set equal to 0.127 mm [19]. For the development of the stent structure, a series of multiple advanced CAD commands, features, and patterns (Extrude, Fillet, Circular Pattern, etc.) were implemented in order to closely replicate the Endurant II characteristics. A total of 17 stent rings were constructed, each with a strut thickness of 0.5 mm [20]. The completed stent graft 3D model was subsequently exported as a step file to Ansys Mechanical for the conduction of the FEA simulation.

### 2.3. FEA—Stent Graft Radial Expansion

The main objective of the FEA simulation was to analyze and assess the mechanical behavior of the stent graft. The expansion target was to reach a diameter size of approximately 10–25% higher than the actual measured vessel diameter in order to achieve better sealing between the aortic walls [17]. In this case, the goal was to overpass the diameter of the ipsilateral leg of the aorta-left iliac artery- which is equal to 10 mm.

The nonlinear analysis was set up as a static structural problem. After both the reconstructed aorta and the customized stent graft models were imported to Ansys software, it was necessary to input the engineering data of the problem, or in other words, the material properties. The material assigned to the stent body was the superelastic nitinol, whereas to the graft component, the polyethylene terephthalate (PET).

Nitinol was modeled as a superelastic material using transformation stress parameters that characterize the forward and reverse transformations between the austenite (high temperatures) and martensite (low temperatures) phases. Depending on the phase, the crystal structure of nitinol transforms. Below the starting temperature of the martensite phase, the nitinol alloy has a low yield strength, allowing it to deform with ease and adapt to new shapes. Upon heating above the austenite finish temperature, the material undergoes a reverse transformation and returns to its original shape [21–23]. To account for hysteresis and pseudoelastic behavior, the constitutive model includes stress-strain paths for both loading and unloading. The forward transformation (austenite → martensite) is defined by the start and finish stresses,  $\sigma_{SAS}$  and  $\sigma_{FAS}$ , while the reverse transformation (martensite → austenite) uses  $\sigma_{SSA}$  and  $\sigma_{FSA}$ . The evolution of martensitic fraction,  $\xi$ , is governed by the following relations:

$$\xi = \begin{cases} 0 & \text{if } \sigma < \sigma_{SAS} \\ \frac{\sigma - \sigma_{SAS}}{\sigma_{FAS} - \sigma_{SAS}} & \text{if } \sigma_{SAS} \leq \sigma \leq \sigma_{FAS} \\ 0 & \text{if } \sigma > \sigma_{FAS} \end{cases} \quad (1)$$

More precisely,  $\sigma_{SAS}$  and  $\sigma_{FAS}$  represent the starting and final stress values for the forward transformation (from austenite to martensite), while  $\sigma_{SSA}$  and  $\sigma_{FSA}$  correspond to the starting and final stresses of the reverse transformation (from martensite to austenite). These parameters are responsible for the shape memory effect of the material under loading and unloading conditions. Additionally, the maximum residual strain, denoted as  $\epsilon$ , quantifies the maximum strain associated with the superelastic plateau. The parameter  $\alpha$  measures the difference between material responses in tension and compression, and lastly,  $E_s$  represents the elastic modulus of the full martensite phase. The nitinol properties are demonstrated in Table 2 below.

**Table 2.** Nitinol properties [24,25].

| Property       | Value  | Unit |
|----------------|--------|------|
| Young Modulus  | 60,000 | MPa  |
| Poisson ratio  | 0.36   | -    |
| $\sigma_{SAS}$ | 520    | MPa  |
| $\sigma_{FAS}$ | 600    | MPa  |

**Table 2.** *Cont.*

| Property               | Value  | Unit                |
|------------------------|--------|---------------------|
| $\sigma_{SSA}$         | 300    | MPa                 |
| $\sigma_{FSA}$         | 200    | MPa                 |
| Epsilon ( $\epsilon$ ) | 0.07   | mm mm <sup>-1</sup> |
| Alpha ( $\alpha$ )     | 0      | -                   |
| $E_s$                  | 60,000 | MPa                 |

The PET material was modeled as isotropic elastic, and its properties are demonstrated in Table 3.

**Table 3.** PET properties [26,27].

| Property      | Value  | Unit               |
|---------------|--------|--------------------|
| Young Modulus | 60,000 | MPa                |
| Poisson ratio | 0.36   | -                  |
| Density       | 1.38   | g cm <sup>-3</sup> |

Following the material assignment of the stent graft, to ensure numerical stability and accurate stress-strain distribution during the simulation, a proper mesh for both models had to be generated. The mesh grid was created individually for each component by selecting the Sizing command. Specifically, the element size values of the stent rings and the graft body were set to 0.8 mm and 0.75 mm, respectively, by using tetrahedral elements. Regarding the aorta body, because of its larger and less complex geometry, it was meshed with the hex dominant method with a higher element size of 8.5 mm. The coarser mesh of the aorta contributed to the minimization of the number of nodes and elements, thereby leading to a substantial reduction in the overall computational cost of the simulation. The total of the three meshed bodies reached the number of 18,831 nodes and 31,565 elements.

The contact settings of the static structural problem were adjusted after the meshing was complete. Two types of contact were applied—one to describe the relation between the nitinol stent and the multifilament graft and one to simulate the interaction of the assembled stent graft with the aorta during the expansion. A No-separation contact was assigned between the internal face of each stent ring (target faces) and the external graft surface (contact face), ensuring they remain completely attached to each other throughout the analysis, without allowing separation. For preventing extensive penetration during the expansion, a Frictionless contact with a Pure Penalty formulation was applied between the stent graft (contact body) and the aorta (target body). Pure Penalty is a method that treats contact as a stiff spring that provides resistance and allows minimal penetration. This resistance is modeled in the form of contact stiffness and was set to the default option. The *Small Sliding* parameter of the contact assumes that the contact interface will have relatively small sliding motion (less than 20% of the contact length) during the entire analysis. However, the radial expansion of the stent graft is large, and significant relative movement is expected between the two bodies. For this reason, the Small Sliding was disabled. The rest of the contact parameters were set to program-controlled.

Finally, the last step for the construction of the static structural analysis was the insertion of the bounding and loading conditions. The aortic model was constrained with a *Fixed Support* condition, which was applied to all the circumferential edges of the iliac legs and the neck, in order to restrict its degrees of freedom and maintain the whole body in a fixed position. The stent graft was subjected to an Elastic Support bounding condition with a foundation stiffness of 0.1 N/m<sup>2</sup> permitting controlled displacement while avoiding excessive deformations. For the radial expansion simulation of the endovascular device, a pressure load was implemented in

order to replicate the shape recovery mechanism of the stent's superelastic nitinol skeleton, which expands to its original shape when exposed to the human body temperature. The magnitude of the pressure was set to 1.8 MPa and was applied to the inner surface of the stent graft.

### 3. Results

#### 3.1. Evaluation of Aorta 3D Model

A finite element analysis was initially conducted for the reconstructed aneurysmatic aortic model in order to evaluate its mechanical behavior and assess the severity of the aneurysm's deformation. The extracted results revealed the stress and strain distribution across the aorta body and highlighted the critical risk of the aneurysm progression.

Figure 3 illustrates the simulation results across the regions of the aorta body. Figures (a) and (d) refer to the total deformation range, whereas Figures (b), (e) and (c), and (f) correspond to the equivalent stress and strain distributions. Across the aneurysmal bulge region, the aortic wall was deformed up to 2.2 mm, increasing the vessel diameter to 52.2 mm and subsequently enhancing the danger of a potential rupture. The stress and strain maximum values were mainly located on the inner curvature of the aorta, opposite the dilated side, and were found equal to 0.10727 MPa and 0.12799 mm/mm, respectively. Regarding the von Mises stress results, the mechanical behavior of the aorta was within the acceptable limits, as the mean peak von Mises stress for non-ruptured aortas can reach the value of 0.62 MPa with a standard deviation of  $\pm 0.3$  MPa [28]. The equivalent elastic strain distribution shows that the model underwent an average elongation of 5%. In cases of abdominal aortic aneurysms, the maximum equivalent strain of the aortic wall has been reported to reach  $32\% \pm 9\%$  [29], indicating that the model remains within the safe limits. Overall, although the 3D abdominal aorta body did not exceed the failure thresholds, the observed aneurysm deformation appears to be approaching a critical point, which brings out the importance and the necessity of the aneurysm treatment.

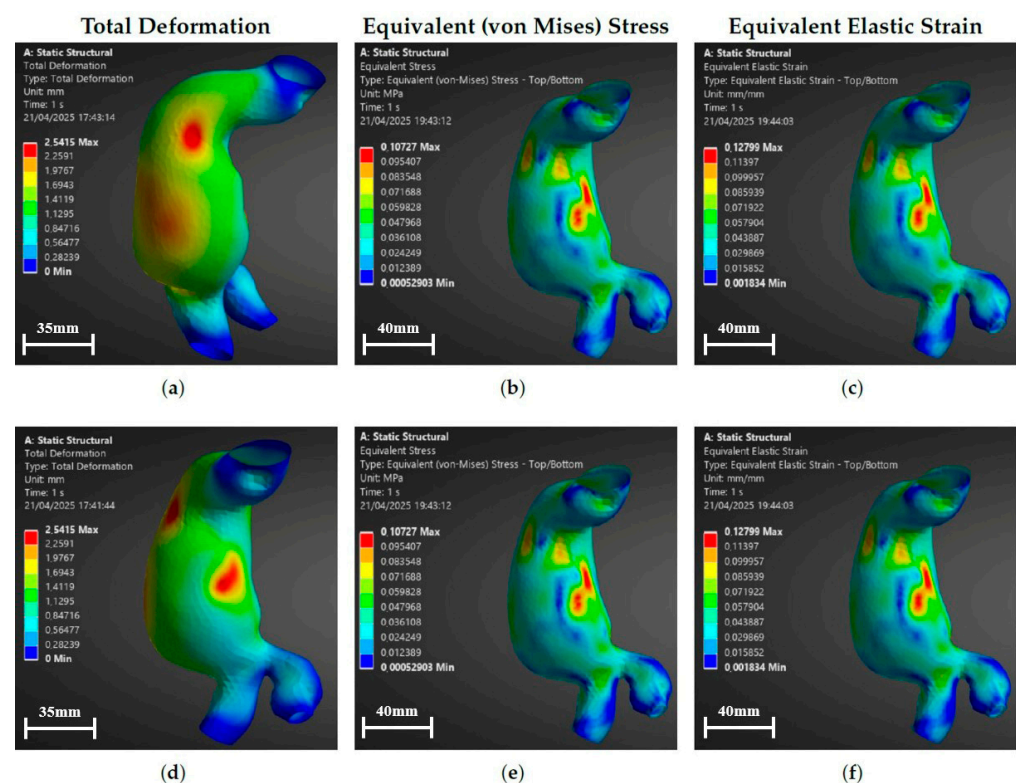


**Figure 3.** FEA results for the aorta: (a) Total deformation of aorta (front view); (b) Equivalent (von Mises) stress of aorta (front view); (c) Total elastic strain of aorta (front view); (d) Total deformation of aorta (back view); (e) Equivalent (von Mises) stress of aorta (back view); (f) Total elastic strain of aorta (back view).

### 3.2. Stent Graft 3D Model

The final stent graft 3D product is presented in this section, showcasing its geometric characteristics. The design approach was based on the Endurant II endovascular device prototype, aiming for a realistic and accurate model. It was composed, as mentioned in Section 2.2, of two components, a polyester graft from PET fabric and a nitinol framework for external support.

Figure 4 illustrates different views of the model and the real medical product. In Figure 4a, a single part of the metallic stent is presented, while in Figure 4b, the connection between the stent and graft components is depicted. Lastly, Figure 4c,d overlay the fully assembled 3D model and the Endurant II design reference, respectively. The designed model was created in a compressed form (before the deployment), as the objective is to simulate the expansion by applying internal pressure loads. It is notable that, similar to the Endurant II, it follows the same geometry curvature and features an M-shape skeleton pattern. On the contrary, the constructed stent graft does not incorporate a bifurcation and has a straight uniform diameter throughout its length.

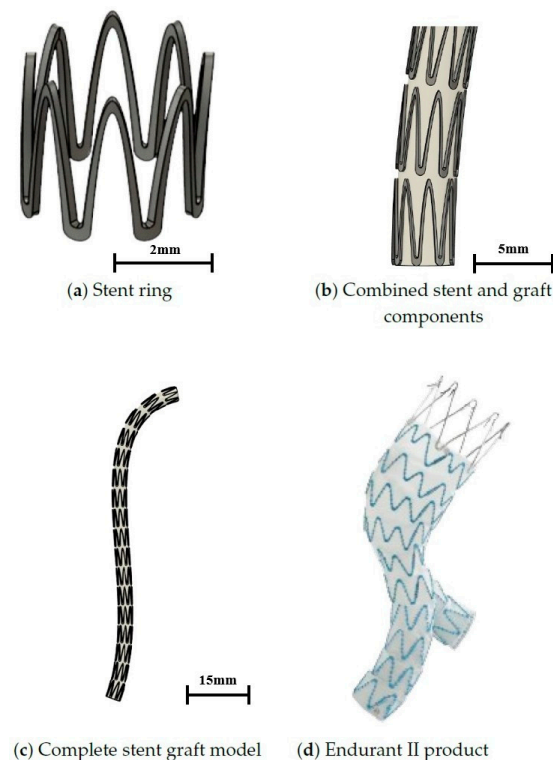


**Figure 4.** Visual representation of the designed 3D model: (a) Stent ring; (b) Combined stent and graft components; (c) Complete stent graft; (d) Endurant II endovascular device [30].

### 3.3. Stent Graft Expansion Assessment

The performance of the stent graft deployment and its interaction with the aortic vessel wall are evaluated in this section. The results of the expansion simulation showcased the mechanical behavior of the designed endovascular device and highlighted the efficiency of the EVAR procedure.

In Figure 5a,b images present the total deformation of the stent graft along with the numerical value spectrum, while Figure 5c,d, cross-sectional views of the combined models, prior to and after the deployment, are shown. The radial expansion corresponds to an average of 4 mm, resulting in a final diameter of 12.5 mm. This dimension lies within the desired target range, as it is 25% larger than the inner diameter of the aortic vessel and ensures proper and accurate sealing. Finally, the contact behavior between the stent graft

and the aorta revealed optimal interaction and no signs of penetration, confirming the successful mechanical performance of the models under the subjected loading conditions.

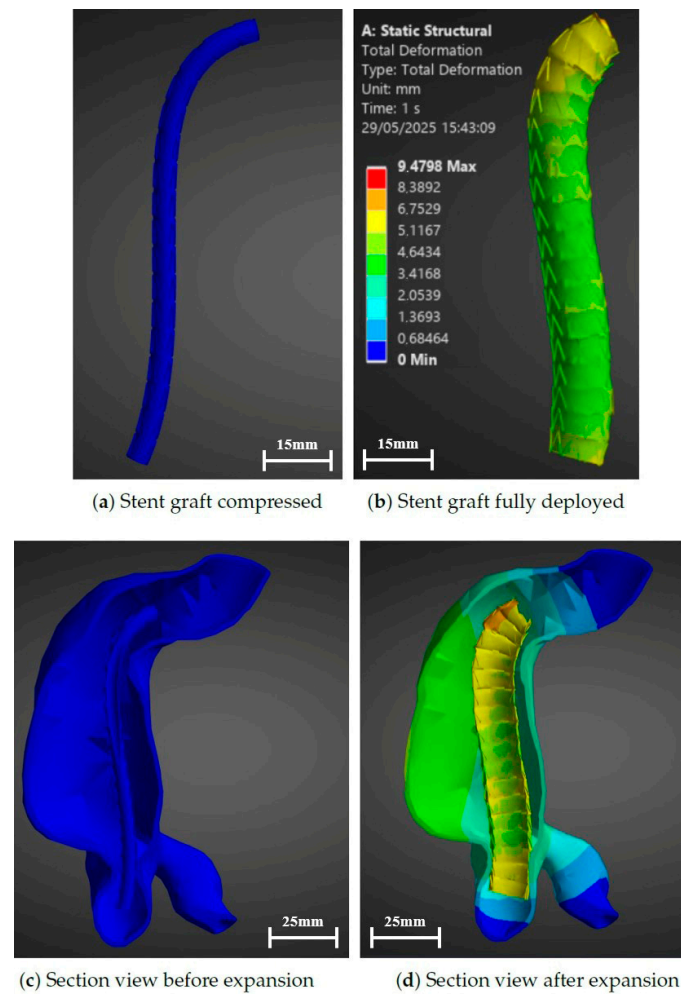


**Figure 5.** FEA total deformation results for: (a) Stent graft compressed; (b) Stent graft fully deployed; (c) Section view before the stent's expansion; (d) Section view after the stent's expansion.

Figures 6 and 7 below illustrate the equivalent von Mises stress and equivalent elastic strain that the designed stent graft model underwent throughout the finite element analysis.

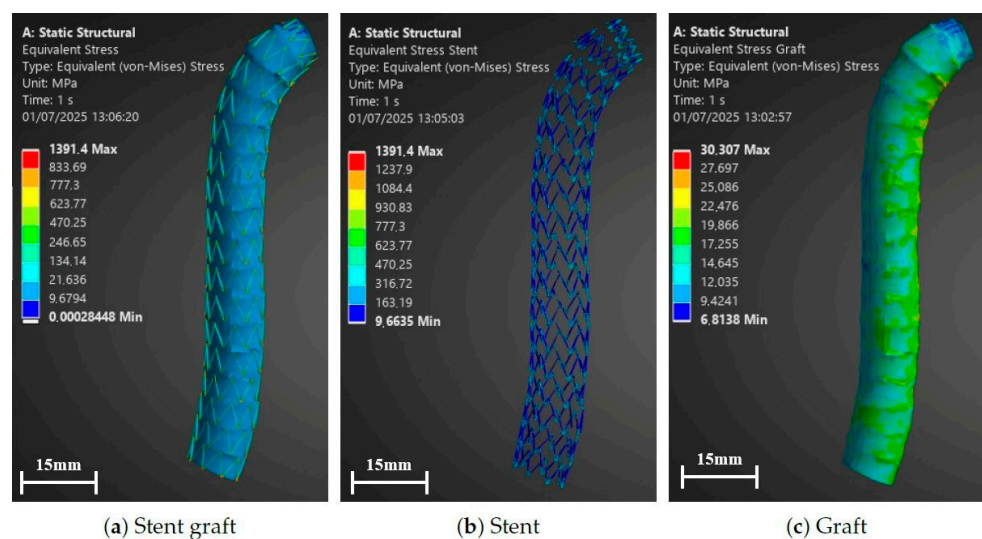


**Figure 6.** FEA equivalent von Mises stress results for: (a) Stent graft; (b) Stent (isolated); (c) Graft (isolated).

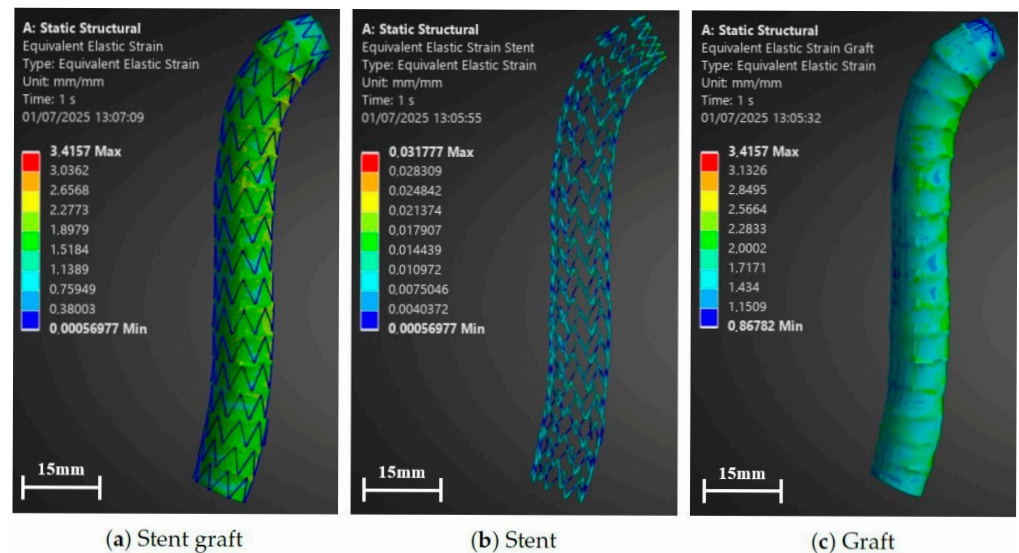


**Figure 7.** FEA equivalent elastic strain results for: (a) Stent graft; (b) Stent (isolated); (c) Graft (isolated).

Regarding the nitinol structure, the developed stress distribution ranged between 9.6 MPa and 1390 MPa according to the solver output. However, this was not the actual representative spectrum as the extreme values are attributed to minor local irregularities that do not affect the analysis. The actual mechanical stress range observed on the stent was 60–400 MPa. The lower values were distributed across the central surface of the stent’s M-shape skeleton, whereas significant stress concentrations appeared in regions with high geometric curvature. These equivalent stress levels are below the material’s yield strength (560 MPa) [31], indicating that the stent graft operates within the superelastic range of nitinol. This allows the activation of the shape memory effect and confirms the stent’s ability to provide adequate support to the aortic wall against the aneurysm. The polyester graft component also exhibited acceptable stress values that range from 15 MPa to 20 MPa uniformly across its surface and showed effective mechanical response under the applied loads.

Based on Figure 7b, the developed elastic strain of the nitinol stent was found to be equal to 1% with an amplitude of  $\pm 0.5\%$ . The distribution was similar to the stress concentrations, as the higher strain values were located at the curved areas of the metallic framework. It has been shown that between 1.5–7% mean strains, the nitinol structure experiences enhanced fatigue life [32], which results in an increased endurance against failure. Therefore, the mean strain of the stent in the current simulation is placed at the lower bound of the optimal range, indicating sufficient mechanical strength and a safe, durable deployment. The PET graft’s elastic strain differed significantly from that of the metallic stent structure. The overall strain values ranged between 0.87 mm/mm and 3.42 mm/mm, although the actual representative strain for the graft material corresponded to approximately 170–200%. This high percentage is attributed to the low elastic modulus of the fabric material ( $E = 10$  MPa), as well as the large enlargement of its diameter. In detail, the PET fabric are attributed to the size effect inherent in the multifilament structure of the material. The individual fibers’ thin and flexible nature leads to localized deformations, resulting in elevated strain values at the microstructural level, even when the bulk material remains within its elastic limits.

## 4. Discussion

### 4.1. Validation and Reliability of FEA for Stent Graft Deployment

The present study developed an FEA framework to simulate the deployment and mechanical behavior of a customized stent graft based on the Endurant II device. Despite the simplifications introduced in the stent-graft geometry, including a uniform cylindrical

shape, our model captures the essential mechanical interactions between the nitinol stent, the PET graft, and the aortic wall. Acknowledging that real clinical devices exhibit more complex geometries and branching; however, the simplifications were necessary to reduce computational cost while still providing accurate insight into stress and strain distributions. Previous studies have demonstrated that such FEA approaches can reproduce the general trends of stent expansion and interaction with vascular walls, showing good agreement with in vitro and clinical observations [3,4,7].

Regarding the mechanical response, the nitinol stent was modeled using a superelastic constitutive model that accounts for forward and reverse phase transformations between austenite and martensite, residual strains, tension-compression asymmetry, and hysteresis behavior. This approach allowed realistic simulation of the stent's shape memory effect and its interaction with the aorta. PET graft strains were observed to reach high values (170–200%), consistent with the fabric's size-dependent mechanical behavior under significant expansion [33,34]. Contact interactions between the stent, graft, and vessel were carefully defined to ensure numerical stability and realistic mechanical responses.

Finally, it is worth mentioning that the current study focused on a single stent-graft design, establishing a methodology that can be applied in future work to explore parametric studies, shape optimization, and rapid validation using AI-assisted FEA tools. While direct experimental validation was not performed in this study, prior literature supports the reliability of FEA models in predicting stent-graft mechanics, suggesting that our results provide meaningful insight into the device performance [35].

#### *4.2. Potential Applications of Additive Manufacturing*

Although the current stent graft was modeled based on an existing commercial device, the methodology developed here is directly relevant to emerging additive manufacturing (AM) approaches. AM processes, such as selective laser melting (SLM) for metallic stents and fused deposition modeling (FDM) or electrospinning for polymeric graft components, could enable the production of patient-specific stent grafts with customized geometry, porosity, or mechanical properties [36]. Such processes would allow rapid prototyping and testing of alternative designs, potentially integrating multi-material components within a single device. Future studies could combine the FEA methodology presented here with AM-based fabrication to explore novel stent-graft designs that optimize mechanical performance and clinical outcomes.

## **5. Conclusions**

The current study aimed to simulate the stent graft radial expansion and its interaction with the aorta during the endovascular aneurysm repair procedure. Certain patient imaging data from CT scans were initially acquired and modified in order to reconstruct a 3D abdominal aortic aneurysm model. A finite element analysis was performed then, using Ansys software (Ansys Student 2025 R2), highlighting the critical aneurysm condition under the applied pressure loads and the need for EVAR. Relying on Endurant II medical product characteristics, the next step was the 3D design of a stent graft device comprising a nitinol wire-formed metal structure and a multifilament polyester fabric. Afterwards, a mutual static structural problem was created for both the aorta and stent graft models, including all the necessary realistic constraints and boundary conditions. Lastly, a final FEA simulation was conducted showcasing the adequate mechanical performance of the endovascular device and its ability to seal to the aortic wall, allowing smooth and safe blood flow and providing support against the aneurysm.

Future research could focus on several key aspects. Firstly, optimizing stent deployment through advanced control techniques to minimize contact and stress concentrations

in the aorta during the procedure. Secondly, exploring novel stent designs with hybrid auxetic structures [37,38], to enhance geometrical configuration and mechanical performance. Finally, integrating model order reduction methods and digital twin technologies could enable real-time implementation, providing valuable support during clinical operations.

In conclusion, this study contributes to both engineering and medical fields by establishing a theoretical framework for understanding abdominal aortic aneurysms and offering a mechanical perspective on EVAR treatment. It provides insights into the structural integrity of stent grafts and highlights their conformability with the aorta. Further research and simulation improvements could enhance model accuracy and yield data of potential clinical relevance for future investigations.

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**Conflicts of Interest:** The authors declare no conflicts of interest.

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