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Comprehensive assessment of mechanical behavior of an extremely long stent graft to control hemorrhage in torso

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Abstract

Traumatic vascular injuries, resulting from either civilian accidents or wounded soldiers, require new endovascular devices (i.e., stent graft) to rapidly control the excessive internal hemorrhage in torso region. Current stent designs are limited by their permanent nature, which is not well suited for emergent placement. A retrievable stent graft could regulate the internal bleeding temporarily, as fast as possible with the most feasible performance, until the patients arrive the hospital to receive the proper treatment. The novel endovascular device of this study is designed according to the anatomy of a porcine model with plans to transition to a human model in the future. The stent graft is manufactured using a substantially long nitinol backbone and covered selectively based on anatomic measurements, with highly stretchable expanded-polytetrafluoroethylene (ePTFE). In this study, our group comprehensively explored designing and manufacturing methods, and their impact on the stent graft performance. Geometric parameters and heat treatment conditions were investigated to show their effect on the radial force of the metallic backbone. As a retrievable device, the resistance force for retrieval as well as deployment were measured, and analyzed to be manipulated through ePTFE covering configurations. In vitro measurements for bleeding were measured using swine aorta to show the functionality of the stent graft under the simulated pulsatile flow circulation. Finally, the stent graft showed substantial effectiveness for hemorrhage control in vivo, using swine model. The new design and fabrication methods enable rapid hemorrhage control that can be removed at the time of a dedicated surgical repair.

Keywords

ePTFE; hemorrhage; nitinol; radial force; retrieval resistance; stent graft

1 | INTRODUCTION

In the past few decades, it has become increasingly popular to treat many vascular diseases using endovascular devices due to the nature of minimally invasive procedure (Andersen, Knudsen, & Hasenkam, 1992; King, Thompson, Steiner, & Mills, 1976; Palmaz, 1988; Simon, Kaplow, Salzman, & Freiman, 1977; Uflacker & Robison, 2001). One of the most important endovascular devices are stent grafts, also known as “covered stents,” which are widely used for repairing damaged blood vessels and diseased arteries or veins (Dotter, 1969; Palmaz, 1992). The representative diseases that can be treated by stent grafts include aneurysm (Parodi, Palmaz, & Barone, 1991), vascular coarctation (Tzifa et al., 2006), arteriovenous fistula (Thalhammer, Kirchherr, Uhlich, & Gross, 2000), and pseudoaneurysm (Maras et al., 2006). Recently, a new stent graft has been developed to control noncompressible hemorrhaging in torso region for civilians or soldiers injured by penetrating trauma (Chen et al., 2018). Commercial stent grafts have been classified as balloon-expandable (Serruys et al., 1994; Suárez de Lezo et al., 1995) and self-expandable stents (Lau et al., 1999; Palma, Marcondes de Souza, Rodrigues Alves, Carvalho, & Buffolo, 2002) depending on the deployment mechanism with the metallic materials used. The type of stent grafts is primarily dependent on the location, anatomy, and the difficulty level of deployment, as well as the length and diameter of blood vessel to be treated, such as thoracic (Dake et al., 1994) abdominal (Blum et al., 1997; Parodi et al., 1991) aorta, iliac (Razavi et al., 1995; Greenberg et al., 2006), and visceral (Larson, Solomon, & Carpenter, 2002) arteries. Typically, commercially available stent grafts have wide range of diameters up to 46 mm and length range from 40 to 250 mm (GORE® TAG® Thoracic Endoprosthesis, 2019; GORE® VIABAHN® Endoprosthesis, 2019). Even the longest of these current stents still remains well below the length that would be needed to cover the length of the aorta.

In most clinical settings, two or more short stent grafts are connected each other by overlapping in order to cover the desired length within the vessel. The delivery is sequentially carried out, where each stent graft is deployed, then the preceding one is deployed with keeping overlapped portion. This deployment procedure is considered relatively complicated and not well suited for emergencies when time is limited and the exact location of the injury may not be known (Grabenwöger et al., 2000; Zhang et al., 2013). A longer aortic stent allows all potential areas of injury to be covered until the exact site of the injury can be identified and repaired.

While there exist many studies on the mechanical behaviors of the typical short stent grafts (<200 mm in length), there is lack of study on the mechanical performance of long stent grafts. One of the essential parameter for assessment of stent graft's functionality is the radial force exerted by stent grafts to the vessels, which is associated with (a) the device migration and (b) the leakage between the vessel wall and the deployed device. Several other features were integrated toward the goal of simplifying rapid delivery of a retrievable stent.

First a tapered nosecone, integrated with the stent, eliminates several time-consuming steps of the stent delivery process, which would be essential in an emergency setting. Next, an extension wire (i.e., push rod) is a crucial component in the extremely long stent graft due to the relatively high resistance during the device movement and in the case of a temporary stent for trauma, is also needed for stent retrieval. Although a few studies show the deployment and retrieval processes of the neurovascular endovascular devices, there is no study conducted for larger and ultralong stent grafts.

Therefore, mechanical behavior study on the new long stent graft would be beneficial for understanding the performance of the device considering deployment, leakage, migration, and retrieval process. In this article, we present comprehensive study results on design, manufacturing, and test results of extremely long stent grafts that could be used for as a future endovascular device to treat traumatic injury. We explored the effect of the geometric parameters on the radial force exerted by nitinol stent grafts. Since nitinol is shape memory alloy, then the thermomechanical behavior is dependent on the thermal history. Therefore, we also investigate the effect of the thermal annealing process on the mechanical response of the stent grafts. In addition, this article examined in vitro quantification of leakage, device deployment, and retrieval processes, as well as in vivo test results in swine model.

2 | MATERIALS AND METHODS

2.1 | Materials: Nitinol, stretchable PTFE, sheaths, guidewires, and radiopaque marker materials

The stent graft consists of (a) metallic stent backbone, (b) polymeric covering membrane, (c) extension wire, (d) nosecone, and (e) radiopaque marker. The backbone was fabricated from 0.0155" diameter superelastic nitinol wire (NDC, Fremont, CA). A highly stretchable expanded-polytetrafluoroethylene (ePTFE), (ZEUS, Inc., Orangeburg, SC, original wall thickness of 0.005") was used to cover the backbone. A superstiff wire (Lunderquist® guide wire, COOK, Bjaeverskov, Denmark) was used to extend the proximal end of the device both for device deployment and retrieval. A custom-built nosecone was directly connected to the backbone, which contains a floppy tip of guidewire and a tapered nosecone mimicking the dilator. A 20 µm thick Tantalum strip (GoodFellow, Coraopolis, PA) was used to fabricate radiopaque markers on the desired locations on the stent graft for the in vivo fluoroscopic guided endovascular procedure.

2.2 | Stent graft fabrication processes

2.2.1 | Nitinol stent backbone: Laser welding and subsequent thermal annealing—Nitinol wires were joined by a precision micro-laser-welding system (LZR-100; Sunstone Engineering, Payson, UT). Welding parameters were optimized in order to acquire the best material and mechanical properties of the structure for nitinol wires with various thicknesses. The laser process parameters are 1.4 kW power, 0.7 mm spot size, 1.2 ms time duration, 3 Hz frequency, and rectangular wave. A machined aluminum mandrel based on the anatomy of the swine model was used to create the nitinol backbone with the desired cylindrical geometry, where the external diameter of the mandrel corresponds to the deployed device diameter. The mandrel diameter was oversized by 25% to produce the

oversized stent backbone for achieving sufficient radial force for the device. The mandrel with stent was heated up to 500°C by tube furnace (Lindberg/Blue M Mold-a-therm, Fisher Scientific, Pittsburgh, PA), then, the stent was rapidly cooled in water to 20°C in 10 s (quenching). Thermal treatment was performed to set the final shape of nitinol wires with required superelastic properties, allowing relieve of local stress concentration caused by welding.

2.2.2 | Prototype fabrication: Radiopaque marker, nosecone, extension wire, and covering—Radiopaque markers were mechanically trimmed from Tantalum strip with width of 1.0 mm and length of 10 mm. The markers were wrapped 5 times on the four difference locations on nitinol strut, such as a distal end, and proximal end. A nose cone was attached to the top end (closer to the heart during stent delivery) after trimming from the delivery sheath dilator, and a floppy tip was subsequently connected on the top nose cone for smooth delivery. This system enables the sheath to navigate through the blood vessels smoothly without any preinstalled guidewire. An extra stiff Lunderquist guidewire was integrated with the device using a 4.0–5.0 size silk sutures (UNIFY®, AD Surgical, Sunnyvale, CA) for the device delivery. The covering membrane, ePTFE (originally ID $0.394'' \pm 0.03$, wall thickness $0.005'' \pm 0.004$) was stretched via ballooning and attached onto a nitinol backbone using a biocompatible polymer adhesive internally or externally.

2.3 | In vitro tests to evaluate mechanical performance

2.3.1 | In vitro radial force measurement of nitinol backbone—Effect of geometrical parameters on radial force of nitinol backbone is quantitatively characterized using different types of prototypes. Radial forces were measured using crimping experiment similar to published papers (Chen et al., 2018; Mummert, Sirois, & Sun, 2013). The radial force was investigated using numerical analysis methods in our previous work (Chen et al., 2018) In this work, the results of experimental radial force measurements were compared to the numerical results. The crimping was performed by a mechanical test system (FLC-5E, Starrett, MA) that pulls one end of Dacron strap (Sailrite Enterprises, Inc., Columbia City, IN). The other Dacron end was fixed to stainless steel rod, and wrapped around the nitinol backbone as shown in Figure 1a. The load cell, attached to the mechanical testing system, measures the forces employed on the prototypes to deform radially as the testing system pulling the Dacron strap. The load cell records the displacement with respect to the hoop force applied on the stent. Radial force is calculated as the hoop force multiplied by 2π . The relation of radial force and, nitinol wire diameter, distance between two successive welds, and stents deployed diameter, were calculated. We examined 0.009'', 0.0120'', 0.0140'', and 0.0155'' wire diameters, while keeping the distance between welds 23 mm and the deployed diameter 20 mm. For distance between welds analysis, radial force was measured for 17, 23, and 30 mm distances between welds, for 0.0155'' wire diameter and 20 mm deployed diameter as in Figure 1b. We tested the effect of the deployed diameter for 15, 20, and 25 mm diameters, Figure 1c, whereas the wire diameter is 0.0155'' and the distance between welds is 23 mm. All these prototypes had heat treatment for 20 min, followed by the quenching process. The results were plotted as radial force with respect to the reduction percentage (%) of the initial stent diameter.

2.3.2 | Assessment of thermal annealing effect—Welding changes the microstructure of the affected zone around the weld, named heat affect zone, which requires thermal annealing to reconstruct the microstructure. Shape setting is performed through thermal annealing, which causes material relaxation at the desired equilibrium shape. Also, the heat treatment temperature and duration affect the mechanical properties of nitinol. Shape setting is usually carried out at temperature around 500°C using mandrel that has the desired dimensions (Shabalovskaya, Anderegg, & Van Humbeeck, 2008). We investigated the effect of the annealing temperatures 450, 500, and 600°C on the radial force. The samples tested are 0.0155" wire diameter, 23 mm distance between welds, 20 mm deployed diameter and heated for 20 min. We examined the effect of heating duration for 10, 15, 20, 25, and 30 min on the radial force, while keeping the temperature at 500°C, 0.0155" wire diameter, 23 mm distance between welds, and 20 mm deployed diameter. Finally, the effect of number of heat treatment cycles was tested on the radial force. Here, we consider the cycle is the heating of stent to the annealing temperature for a specific time, then quenching using water to room temperature. We measured radial force for 1, 2, and 3 cycles for 500°C annealing temperature, 20 min annealing time, 0.0155" wire diameter, 23 mm distance between welds, and 20 mm deployed. The results were plotted as radial force with respect to the reduction % of the initial stent diameter.

2.3.3 | In vitro device deployment and retrieval tests for evaluating resistance

—Retrieval stents represent a novel solution, as they enable the surgeons to use stent graft as temporary remedy for life threatening bleeding. One of the obstacles for fully retrieval of temporary stents is the resistance exerted by the stent in the catheter. The setup for resistance measurement is shown in Figure 2a. We investigated the deployment (pushing the stent from the catheter) and the retrieval (pulling the stent into the catheter) with respect to the ePTFE attachment to the nitinol backbone and the size of the catheter used. The pushing/pulling forces were measured for the backbone only, ePTFE attached outside the backbone, and ePTFE attached inside the backbone as shown in Figure 2b. The prototypes were deployed and retrieved in a 10 Fr sheath (OSCOR, Palm Harbor, FL). The backbones used for measurement were 500 mm long, 0.0155" wire diameter, 23 mm distance between welds, and 20 mm deployed. The heat treatment of the stents was carried out at 500°C for 30 min. The effect of the catheter size was investigated by recording the pushing/ pulling forces for 8, 9, and 10 Fr catheters. The stent, used in these tests, was similar to previous tests in terms of geometrical dimensions and heat treatment parameters. The results were plotted as deployment/retrieval force with respect to the deployed % of the stent outside the catheter.

2.3.4 | In vitro leakage test for different type of stents

—The ability of the stent to prevent leakage within a tubular structure is crucial for evaluating the performance of the stent graft. We used swine aorta extracted from Yorkshire pigs, and kept in 0.9% sodium chloride solution (Baxter, Deerfield, IL). The experimental setup is shown in Figure 3a. A mixture of glycerin and water (58.5:41.5% v/v; density, 1.16 g cm⁻³; dynamic viscosity μ , 4.0 cP) was used to simulate the viscosity of blood. The fluid was circulated using a pulsatile pump (Harvard Apparatus, Holliston, MA) at different pressures and flow rates. The flow rate was controlled by the pulsatile, and the solution pressure was measured using pressure probe (PressureMAT™ DPG, PendoTech, Princeton, NJ). The solution was heated to 37°C

using water bath (ThermoFisher, Waltham, MA) to simulate the body temperature. We measured the diameter of the distal end of the aorta (left in Figure 3b), to manufacture a stent graft to cover the aorta internally. The aorta diameter was 20 mm. Two stent grafts were used to control the leakage through the aorta, the first device diameter was 20 mm and the second diameter was 25 mm. The aorta used in the testing was 200 mm long, with internal diameters 20 mm on the top (proximal end to heart) of the aorta, 17 mm on the middle of aorta, and 13 mm on the bottom (distal end to heart) of the aorta. The aorta had several holes, as shown in Figure 3c. The aorta had two holes with 3 mm diameter, one holes with 5 mm diameter, and two holes with 6 mm. The leakage was recorded for uncontrolled leakage (without stent graft), and for the two stent grafts. The stent grafts used in all leakage tests are 300 mm long, 0.0155" wire diameter, 23 mm distance between welds and heat treatment 500°C for 30 min. The backbones were covered with ePTFE externally as shown in Figure 3c.

2.4 | In vivo tests

Yorkshire pigs 80–90 kg were obtained from Archer Farms, Inc. (Darlington, MD), and were placed under general endotracheal anesthesia, upon the approved IACUC protocol as part of University of Pittsburgh with inhaled isoflurane. Operative exposure was done to expose the aorta. In order to simulate the derangement of clotting after trauma, the animals were heparinized to overcome the rapid clotting occurred inside the porcine model. We used 22 Fr dilator to make an injury in aorta artery, then, a long 10 Fr sheath was used to deliver the device under fluoroscopic guidance. Several angiogram images were taken in different situation for the porcine model, an angiogram for injured thoracic aorta, and hemorrhage control for thoracic aorta. The injury was observed for 1 hr, and the leaked blood was qualitatively assessed. As a final step, the vascular sheath was advanced over the stent graft for collapsing and retrieval of the device.

3 | RESULTS AND DISCUSSION

3.1 | A prototype manufactured using optimized conditions

A functional prototype stent graft has been successfully manufactured using optimized conditions obtained from various in vitro mechanical experiments. Figure 4 shows the device image that is composed with a nosecone, first long PTFE covered region for thoracic covering, bare metal region for perfusion to organs, second short PTFE covered region for abdominal covering, and an extension wire for easy device delivery and retrieval process. In vitro test results to assess mechanical performance of this device are followed.

3.2 | In vitro tests to evaluate mechanical performance

3.2.1 | In vitro radial force measurement of nitinol backbone—Figure 5 shows the radial forces of the metallic backbones with varied geometric parameters, including nitinol wire diameters, welding distances, and deployed diameters. The radial forces are increased with thickener wire diameter as the backbone diameter is reduced by crimping. The highest radial force was observed as 6.5 N for 0.0175" nitinol wire thickness at 42% backbone diameter reduction as shown in Figure 5a. The distance between welds has the highest influence on increasing radial force by shrinking the welding distance as shown in

Figure 5b. The welding distance “17 mm” has the highest radial force of 8.9 N. Figure 5c represents the outcome of increasing the deployed diameter, with highest radial force 6.8 N for 25 mm deployed diameter.

3.2.2 | Assessment of thermal annealing effect—The heat treatment conditions show great impact on the radial force of the metallic backbones. Three parameters annealing temperature, annealing time, and number of cycles, are explored for evaluating the radial force of the nitinol backbone. Figure 6a illustrates the effect of the annealing temperature on the radial force with respect to the diameter reduction % of backbones in crimping process. The maximum radial force is observed as 7.3 N at annealing temperature 600°C. Annealing time was investigated by varying the time from 10 min to 30 min. For 10 min annealing time, the radial force was less than 1 N, however, the radial force increases significantly by increasing the annealing temperature as shown in Figure 6b. The radial force is saturated, when it reaches 20 min annealing time. Figure 6c shows that the number of cycles do not have significant differences in different test samples, presenting negligible influence to the radial force.

3.2.3 | In vitro for evaluating resistance occurred during the device deployment and retrieval—As a retrievable endovascular device, the resistance occurred both during deployment and retrieval significantly affects the performance of the stent graft delivery. We investigated the ePTFE configuration and the sheath sizes, 8, 9, and 10 Fr on the resistance forces of the deployment and retrieval of the stent graft. Figure 7a, b shows the resistance forces of the bare stent (No ePTFE) in different sheaths sizes. The highest resistance force was for 8 Fr sheath, while 10 Fr sheath had the lowest resistance forces. The resistance forces of retrieval and deployment for stent grafts with different ePTFE configurations are shown in Figure 7c–f. The resistance forces for outside ePTFE showed higher resistance forces, also, 9 Fr sheath revealed higher resistance forces compared to 10 Fr sheath.

Figure 8a,b compared the resistance forces of deployment and retrieval between different ePTFE configurations, bare (no ePTFE), inside ePTFE, and outside ePTFE stent grafts. The bare case showed the lowest resistance force as less material is deployed or retrieved. For inside and outside ePTFE cases, same material is deployed or retrieved, however inside ePTFE showed less resistance forces. We investigated the outcome on resistance forces for using several retrieval speeds, and the slower speed 250 mm/min showed less resistance force.

3.2.4 | In vitro leakage test for different type of stents—Leakage quantification is a critical parameter for successful device performance. We measured the leakage using an in vitro flow circulation that equipped with the harvested swine aorta for evaluating the performance under different pressures and flow rates. We compared the uncontrolled case, where no device is used, with cases 20 and 25 mm stent graft diameters are deployed inside the aorta. Figure 9a illustrates the results of leakage for pressures 85, 72.5, 60, and 52.5 mmHg, while keeping the flow rate at 2500 cc/min. The maximum leakage occurred in the uncontrolled cases, and with increasing the stent graft diameter, the leakage reduced. Also, the higher pressure of the fluid exerted showed higher amount of leakage through the aorta.

We investigated the effect of the flow rate, while fixing the pressure of the solution used. The leakage increased with higher flow rates, as well as, the leakage decreased with using larger stent graft diameter as shown in Figure 9b. This demonstrates that the stent is able to cover an injury and prevent simulated bleeding. Additionally, the potential leakage in different size of conduit has been evaluated as shown in Figure 9c. This qualitative test results conducted with silicone tubing, show that the stent was able to prevent leakage even when the conduit was as much as 50% smaller than the stent size.

3.3 | In vivo test with swine model

Swine model was used to demonstrate the stent graft functionality in vivo, where a controlled hemorrhage is induced inside the pig, and our stent graft is introduced to control the bleeding. Figure 10a shows the bleeding occurred in the aorta, where the dark color represents the blood flow outside the artery walls. Figure 10b demonstrates the device deployment inside the artery, and the hemorrhage is controlled, as the blood flows normally through the artery. Figure 10c shows a gross look to the exposed injured artery. The ePTFE is depicted on the site of the injury, and no bleeding is shown around the injured artery.

4 | DISCUSSION

The desired features of a new long stent, used for massive traumatic hemorrhage of the great vessels with limited experience and limited imaging, would be rapid delivery, the ability to control hemorrhage along the entire vessel even when the injury site is not known, and the ability to remove the stent at the time of a more focused permanent repair. This stent is intended for an ultrashort duration of implantation for emergent but brief control of life threatening hemorrhage as a patient is transported between hospitals or even with a hospital, with an expected duration of less than 1 hr. Patients can bleed out in less than 15 min, which is why this stent is necessary and will only be in place until a qualified vascular specialist can permanently repair the injury (and remove the stent). There are many parameters for stent design and fabrication. While radial force is essential to prevent migration of a permanent stent, it was expected that less radial force would be needed for a temporary stent and in fact would improve the ability to more easily collapse the stent back into a sheath after use. Since the device is placed in the vascular system temporarily only less than an hour, there would be no host tissue interactions with the stent (Scott, 2006; Weintraub, 2007) that enables the retrieval process easy with the reduced radial force in the extremely long stent graft. In addition, our stent is inserted from the groin, and is positioned in the aorta, which is considered a very straight vessel, thus no bending is expected in the proposed clinical application. Therefore, three major mechanical performance assessments have been conducted in vitro, including radial force measurement, retrieval resistance measurement, and leakage quantification. Then, the feasibility of device has been successfully demonstrated in vivo after the functional prototype construction with the in vitro data.

Radial force was first evaluated using various geometric parameters of the metallic backbone, which can provide device's refinement strategies to customize the radial stiffness. The radial force assessment is associated with both the device migration and blood leakage performance, therefore this study was first investigated. The radial forces of different nitinol

wire diameters (i.e., stent struts) showed that the radial force increased with larger nitinol wire diameter. The radial force can be expressed as the radial stiffness multiplied by the deformation. By assumption that stent graft is roughly cylinder tube, the radial stiffness was expressed to be directly proportional to tube thickness. As shown in (Ragab & Bayoumi, 1998), by increasing the wire diameter (in our case, the cylinder thickness is equivalent to wire diameter), the radial stiffness increases, which increases the radial force. The second parameter that determines the radial force of stent grafts is welding distance. The results demonstrated that the radial force increases by reducing the distance between welded joints, as the deformed length between welds decreases, which increases backbone's stiffness.

Oversized stent backbones can provide high radial force and potentially prevent device migration. Too much oversized devices could potentially create damages on the blood vessel wall or muscle, so it should be avoided. Therefore, fully deployed nitinol backbones with various final diameters were tested for assessing radial force with different level of oversizing effect. It was found that the backbones with larger deployed diameter exert higher values of radial force when the backbones crimped to the same diameter reduction %. Larger diameters need more radial displacement to achieve similar diameter reduction %, which requires more radial force applied on the larger diameter backbones.

Another unique parameter to determine the device's mechanical performance is thermal treatment that is used to fabricate shape-set nitinol backbones. Shape-setting process of nitinol governs the transformation temperature and final superelastic behavior at body temperature (Xu, Wang, & Liu, 2011). In order to relieve the stresses and preserve the final shape of the nitinol wires, heat treatment should be higher than 450°C and longer than 10 min (Liu, Wang, Yang, & Qi, 2008). The thermal treatment conditions were studied to demonstrate the effect of shape-setting parameters on mechanical response of the metallic backbone. The annealing temperature alters the device's radial force, as the annealing temperature increases, the radial force increases. At higher temperature, more thermal energy is available to allow Ni and Ti atoms to diffuse (Fan et al., 2004; Khalil-Allafi, Eggeler, Dlouhy, Schmahl, & Somsen, 2004), which affects the phase transformation temperatures and mechanical properties of material. For annealing temperature, the results showed that 10 and 15 min had lower radial force compared to other annealing times. These results can be understood as at the high temperatures and atoms diffusion, no enough time was allowed to nuclei to be precipitated (Liu et al., 2008), which affect the radial force. There is no significant effect for the number of cycles as the heat treatment conditions were identical.

The resistance force for deployment and retrievable of the stent graft is important for the device's performance assessment. The resistance forces were measured for bare metallic backbone without ePTFE covering, inside covering ePTFE, and outside covering ePTFE. The metallic backbones definitely required less force for deployment and retrieval within different sheaths sizes, as the metallic backbone radially compressed is the only material that is collapsed within the sheath. Therefore, the resistance force is relatively very low, as the friction is only generated between the internal surface of sheath and the nitinol wire backbone only. There were increased resistance forces for the ePTFE covered backbones, either inside or outside covering, because the crimped ePTFE membranes along with the

nitinol backbone actually increase the total volume of the collapsed device. Additionally, the crimped ePTFE directly contact the luminal side of the delivery sheaths generating a much higher resistance. For these reasons, ePTFE covered backbones could not be deployed or retrieved with a smaller size sheath (i.e., 8 Fr).

The resistance force for deployment and retrieval of the stent grafts decreased with using larger stent size, as the luminal space for the device inside the sheath becomes larger, which allows proceeding the stent graft easier. For inside and outside covering ePTFE attachment to metallic backbone, the outside covering ePTFE shows higher resistance both in deployment and retrieval. This can be understood as the contact area in case of inside covering ePTFE is the contact area of the metallic backbone, which was built up using thin nitinol wires. However, for outside covering ePTFE, the contact area is for the cylinder-shape ePTFE, which is much larger compared to the backbone contact area. Also, ePTFE is glued to the metallic backbone only on both ends in the case of outside covered ePTFE, which permits most of ePTFE to deform freely, so wrinkles formation was more common. These wrinkles increase the resistance significantly during pushing and pulling the stent graft within the sheath.

Leakage quantification is a crucial step to evaluate the device's functionality. In vitro tests with swine aorta were first conducted under several pressures while fixing the flow rate. Then, the device leakage was tested under different flow rates with a fixed pressure level. Physiological data such as pressure levels and flow rates obtained from swine tests were used in these tests. The mean blood pressure range was from 20 to 100 mmHg, and the average flow rate range was between 1 and 10 L/min. The results showed that the leakage increased for higher pressure and flow rate. The results indicated the importance of using oversized stent graft, especially on the ends. We used 20 mm (no oversizing) and 25 mm (25% oversizing) diameter stent grafts, where the aorta maximum diameter is 20 mm. The minimum leakage was found for 25 mm diameter stent graft where, stent graft oversizing ratio is 25%, which is the recommended oversizing ratio (van Prehn et al., 2009). The stent grafts showed good control for the leakage, using the oversized stent graft.

Based on in vitro results for nitinol structure, regarding geometric parameters, thermal conditions, and leakage results, we have designed and fabricated a novel stent graft for in vivo swine tests (Figure 4). There is no commercially available stent graft covers the thoracic, visceral and abdominal aorta, simultaneously. As a result, the cumulative friction across an extremely long stent is a major consideration during deployment. In addition, commercial stents are designed for stepwise wire and catheter exchanges that are not practical for physicians inexperienced in endovascular procedure and certainly in emergency settings. The retrieval feature is also not available on any other commercial stent design. To address these challenging issues, we have studied and developed a single-body stent graft system that contains the integrated guidewire tip, nosecone, and delivery wire. Several unique design/fabrication techniques contribute significantly to the ease of deployment of this ultralong design providing distinct difference compared with commercial permanent stents or stent grafts. In vivo test results have showed the feasibility of using the device for bleeding control, developed in injured vascular system. Gross look for the stent graft inside

an injured artery, demonstrated safe handling for the hemorrhage with minimum leaked blood around the injured artery.

An alternative to fabricate mechanically durable extremely long stent grafts would be with the use of ultraprecision laser cutting process for nitinol stent backbone. A new laser cut nitinol stent backbone has been recently manufactured (Figure 11), which has higher radial force and lower profile. Identical design parameters to control the extremely long stent graft's mechanical performance can also be used to optimize the next generation prototype's mechanical properties.

5 | CONCLUSIONS

A novel extremely long stent graft has been manufactured using an optimized micro-laser-welding technique and a unique attachment technique of the highly stretchable ePTFE tube. The radial force of the metallic backbone was comprehensively studied in terms of geometric parameters and heat treatment conditions. Wire diameter, laser-welding distance, as well as deployed diameter can be manipulated to customize the radial force of the metallic backbone. Heat treatment conditions such as, annealing temperature and annealing time showed significant effect on the backbone radial force. Various forms of configurations for ePTFE covering onto a nitinol backbone was utilized to accommodate the retrieval and deployment resistance level, especially for the long stent graft. In vitro leakage measurements demonstrated the capability of 25% oversized stent graft to control bleeding in swine aorta. Through in vivo tests in swine models, our novel developed stent graft, with retrieval compatibility, could be effective for torso hemorrhage control.

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FIGURE 1.

(a) Mechanical setup for radial force using mechanical testing system by pulling Dacron strap surrounding the stent. (b) Stents with different distance between two consecutive welds, 17, 23, and 30 mm. (c) Stents with different deployed diameters, 15, 20, and 25 mm

FIGURE 2.

(a) Experimental setup for resistance force measurement during deployment and capturing the stent graft in different sheaths. (b) Different configurations of ePTFE with respect to the metallic backbone, bare, internal, and external

FIGURE 3.

(a) Experimental setup for leakage quantification between stent graft and swine aorta. The setup consists of pulsatile pump, water bath, and tank of glycerin-water solution. Pressure sensor is attached to measure the solution pressure and recorded by pressure transducer. (b) Close look for the swine aorta. (c) An image for the stent graft deployed inside the swine aorta

FIGURE 4.

A prototype that contains a nosecone, two PTFE covered regions, bare metal region for perfusion and an extension wire for easy delivery and retrieval process

FIGURE 5.

Radial force measurements for different geometric parameters. (a) Radial force for several wire diameters, 0.012", 0.014", 0.015", and 0.0175" with respect to diameter reduction %. (b) Radial force for different welding distances, 17, 23, and 30 mm with respect to diameter reduction %. (c) Radial force for stents with different deployed diameters, 15, 20, and 25 mm with respect to diameter reduction %

FIGURE 6.

Radial force measurements for stents with different heat treatment conditions. (a) Radial force for different annealing temperatures, 450, 500, and 600°C with respect to diameter reduction %. (b) Radial force for several annealing time, 10, 15, 20, 25, and 30 min with respect to diameter reduction %. (c) Radial force for different number of annealing cycles with respect to diameter reduction %

FIGURE 7.

Resistance force for retrieval and deployment of the stent graft in the sheath. (a) and (b) are graphs that represent the retrieval and deployment resistive forces versus the stent length of bare stent in 8, 9, and 10 Fr sheaths. (c) and (d) are graphs that represent the retrieval and deployment resistive forces versus the stent length of internally attached ePTFE stent in 9 and 10 Fr sheaths. (e) and (f) are graphs that represent the retrieval and deployment resistive forces versus the stent length of externally attached ePTFE stent in 9 and 10 Fr sheaths

FIGURE 8.

Resistance force for retrieval and deployment of stent graft in 9 Fr sheath. (a) and (b) are the resistance forces versus stent length for stent graft with different configurations of ePTFE, bare, inside, and outside. (c) Resistance force versus stent length for different capturing speeds, 250, 500, and 750 mm/min

FIGURE 9.

leakage measurement for swine aorta using different pressures and flow rates for uncontrolled, 20 and 25 mm. (a) Leakage measurements for different pressures, 85, 72.5, 60, and 52.5 mmHg with flow rate 2,500 cc/min. (b) Leakage measurements for different flow rates, 900, 1,200, and 1,500 cc/min with pressure 25 mmHg and (c) qualitative representation of the leakage in the tubes with different diameter, which shows a 15 mm stent provides seal of (c-1) an 8 mm hole against pulsatile flow, (c-2) 75% of the stent, and (c-3) even 50% smaller

FIGURE 10.

In vivo images to a swine subject for hemorrhage control using the stent graft. (a) An image shows the hemorrhage, occurred in the thoracic region of the aorta artery. (b) An image shows the hemorrhage control using the stent graft. (c) Gross image of the stent graft deployed inside an injured artery

FIGURE 11.

A laser cut nitinol stent backbone covered with ePTFE for the extremely long stent graft to control torso hemorrhage