CHAPTER 12

SMA Cardiovascular Applications and Computer-Based Design

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12.1 INTRODUCTION

The aim of this chapter is to provide a comprehensive overview of the application of shape memory alloys (SMAs) in the field of cardiovascular surgery, which represents the most significant share of the SMA medical market. The unique features of SMAs, and in
particular of Nitinol, which is the most-used SMA in the biomechanical field, fit well with the needs of current trends in cardiovascular surgery: the minimization of the device size and, at the same time, the reduction of the procedural impact on the patient’s recovery time. Such an approach, commonly called minimally invasive surgery, is usually based on operations performed through small artificial incisions or natural body openings. In particular, for cardiovascular applications, minimally invasive surgery is usually referred to as endovascular surgery because the miniaturized devices are delivered by low-profile catheters, introduced through narrow vascular accesses running inside the cardiovascular system.

The outlook of SMA-based cardiovascular devices is promising not only because medical procedures are becoming less and less invasive, but also because the impact of cardiovascular diseases (CVDs, which is the generic name given to the dysfunctions of the cardiovascular system such as atherosclerosis, hypertension, coronary heart disease, heart failure, or stroke) steadily increases. In fact, CVDs are now the leading cause of death in the Western countries. A report from American Heart Association (2011) stated that, on the basis of 2007 mortality rate data, more than 2200 Americans die of CVD each day, an average of 1 death every 39 s. The same trend is also present in Europe, where CVDs are the major cause of death in almost every European country, causing over 4 million deaths. These data, integrated with a constant increase of the average life expectancy, explain the high impact of such pathologies on the economy of Western countries. In Europe, the estimated direct and indirect cost of CVDs is €196 billion per year, whereas the total direct and indirect cost of CVDs and stroke in the United States for 2008 is estimated to be $297.7 billion.

Given these considerations, in this chapter we review the application of SMAs in the field of cardiovascular devices, with a special focus on the device design and, in particular, on how advanced computer-based modeling is used in this context. For this reason, the chapter will first give a brief overview of the different types of SMA cardiovascular devices; then, for each of them, a literature review of the related numerical simulation is proposed.

### 12.2 CARDIOVASCULAR DEVICES: AN OVERVIEW

Historically, vascular diseases have been treated by combining open surgery with medical management. Since the first description of a percutaneous procedure, the use of endovascular approaches has revolutionized the treatment of vascular disease. In recent decades, endoluminal therapy of vascular diseases has expanded from simple dilatation of atherosclerotic lesions to more complex acute lesions, assuming also a significant role in the management of aortic aneurysms or dissections.

As already stated, these broadened indications for endovascular therapy have been supported by technological improvements in the concept, design, and technologic content of endovascular devices. Among this advancement, the contribution of SMAs, and in particular of Nitinol, cannot be ignored.
The goal of this section is to provide an overview of the SMA-based applications in cardiovascular surgery. Given the constant growth of such a type of application, the present review is not exhaustive, but it aims at assessing the main applicative sectors and the clinical tasks targeted by the SMA features embedded in the device. It is worth noting that most of the applications described here can be used for endovascular surgery because, thanks to the pseudoelastic effect, they feature the capability to be highly compressed within a low-profile delivery system and to recover the desired shape after the deployment.

We have categorized the SMA cardiovascular devices into three main families: (1) catheters and guidewires; (2) embolic filters; and (3) stents and stent grafts. The first category represents the basic elements of endovascular approach. Embolic filters were the first SMA-based devices in the cardiovascular field, whereas stents and stent-grafts represent the major share of the SMA cardiovascular market. We have collected all of the remaining applications in a supplementary, generic category (i.e., other cardiovascular devices).

### 12.2.1 Catheters and Guidewires

The pseudo-elastic effect of Nitinol is exploited in two fundamental components of the endovascular procedure toolkit: the guidewire and the catheter. The purpose of a guidewire is to create a pathway in the vascular system up to the target lesion (Fig. 12.1).
The guidewire can be thought of as the rail on which the catheter runs. The catheter is basically a low-profile hollow tube, aimed at delivering a miniaturized device to the target lesion. The operator has an indirect view of the system through real-time X-ray images where the vascular tree visibility is enhanced by the injection of a contrast dye.

In most endovascular procedures, guidewires have to achieve and keep the critical access across the target lesion along the entire procedure. They must be significantly long in order to reach vascular districts, which are far from the endovascular access. For instance, the carotid artery, which is located in the neck, may be reached from a femoral access point, which is instead located in the upper part of the leg. Moreover, the vascular path could be characterized by a severe tortuosity and numerous side branches; this is the reason why wires have to be steerable and torqueable. This issue also clearly explains why even a small permanent deformation is undesired.

To accomplish such requirements, current guidewires are usually made of an inner core wire (mandrel) surrounded by an outer coil wrap. The core wire has a tapered shape in the distal part, which does not extend to the guidewire tip, where a more flexible and ideally atraumatic distal coil tip is instead positioned. Moreover, in order to improve lubricity, the guidewire surface is coated with a hydrophilic polymer.

As pointed out by Duerig and colleagues in 1999, guidewires were the first applications to take advantage of the kink resistance (i.e., resistance of the wire to deformation) of Nitinol. In fact, the inner core of guidewires was originally made of stainless steel. Later on, the flexibility provided by the unique properties of SMAs allowed the development of novel guidewires, which are able to enhance the dexterity of the surgeon. Nowadays, both stainless steel and Nitinol are used for the inner part of guidewires. Stainless steel provides the advantage of a good torque controllability, whereas Nitinol is more stretchable and steerable. Moreover, Nitinol is used also for its "soft" end-user feel, which allows for precise placement and delivery through very tortuous paths without losing the good torqueability and pushability of the device.

Future developments are expected in micro-guidewires (diameter less than 0.5 mm), which can be used for abdominal blood vessels, peripheral arteries, and cerebral arteries, in view of less and less invasive surgical techniques.

Nitinol is also used to produce active catheter shafts: high deformability, kink resistance, and the capability to recover the original shape make them far more controllable than conventional devices and allow easy insertion and inspection inside the human body. One of the appealing applications of SMAs in this sector is the actuation of the catheter, which can be based on different types of NiTi microactuators, providing various motions such as bending, torsion, and extension/contraction, as well as presenting different deformation mechanisms according to the requirements of the device. For example, thin NiTi plates fixed along the side of the catheter, or NiTi wires embedded in the catheter body, can provide the bending motion when heated. NiTi coils that are designed to twist or untwist according to temperature variation can provide torsional...
motion. Extension motion can be instead provided by NiTi coils acting as an extending or contracting spring. Moreover, stiffness control can also be provided by equipping the catheter with further SMA coils, which become stiffer when heated because of the phase change from martensite to austenite. The activation mechanism usually relies on the Joule effect, producing heat by the application of an electric current.

### 12.2.2 Embolic Filters

Embolic filters (Fig. 12.2) are another successful example of the use of the pseudoelastic effect for endovascular surgery. In fact, the basic mechanism exploited in such a type of application is the capability to expand inside the lumen, starting from a low-profile shape.

The device, compressed within the catheter, is delivered inside the vessels and deployed thanks to the pseudoelastic effect. Once the filter is open, it has the aim of capturing small clots floating within the bloodstream. When the basket filter is full, it is usually retrieved by a pull-back mechanism, which reduces the device profile to the compressed configuration, thus allowing the endovascular navigation.

A number of design variants have been proposed for this class of SMA-based devices, but their efficacy is still a matter of clinical debate.

In this context, a special mention should be given to the venous Simon filter, which can be considered to be the first vascular SMA application. This pioneering filter, still used in clinical practice, is used to prevent emboli in patients who are unable to tolerate anticoagulants and exploits the shape memory effect. The device, manufactured in the open configuration with a NiTi alloy having austenite finish temperature (Af) equal to the body temperature, is preloaded into a catheter while in the martensitic state. The

![Figure 12.2 Illustrative sketch of different carotid embolic filters adapted from Mousa et al. (2012):](image-url)

(a) Spider FX (Covidien, Mansfield, Mass); (b) Emboshield Nav6 (Abbott Vascular, Santa Clara, Calif); (c) FiberNet (Lumen Biomedical Inc, Plymouth, Minn); (d) FilterWire EZ (Boston Scientific Corp, Natick, Mass); (e) RX Accunet (Abbott Vascular, Santa Clara, Calif); (f) Angioguard RX (Cordis Corp, Bridgewater, NJ).
device is then easily crimped on the catheter thanks to the residual deformation due to the martensitic transformation, which is kept through a chilled saline solution flowing in the catheter during the insertion of the filter into the body. Once the catheter is in place, the filter is released, the flow of chilled saline solution is stopped, and the device is warmed by the surrounding blood, thus recovering its predefined flower-like shape.

12.2.3 Stents and Stent Grafts

The term *stent* derives from Dr. C.T. Stent, a dentist who developed an orthodontic device in the late 1800s to assist in fanning an impression of teeth. Such a term is nowadays generally given to an expandable tube-like device that is inserted into a natural conduit of the body to mechanically enlarge a disease-induced localized narrowing of the duct (Fig. 12.3). Although some stents are made by polymers, most of them are metallic. In endovascular surgery, stents are usually used to enlarge the so-called stenosis—that is, an abnormal narrowing of arterial lumen due to a local disease-induced thickening of the vessel wall.

The metallic frame of the device can be (partially) covered by a polymeric skirt to form a hybrid vascular prosthesis called a *stent graft*. Such devices are normally used to diverge the arterial blood flow within pathologic vascular anatomies; for instance, stent grafts are used to reduce the presurization of the aneurysmatic sac or to exclude the perfusion of false lumen in a dissected artery.

The real success of stents rose from their use in the treatment of coronary stenosis. In fact, the intravascular stent drew the interest of the medical community in 1994, when the stainless steel Palmaz-Schatz coronary stent was launched to integrate percutaneous transluminal angioplasty (PTA), a procedure exploiting the inflation of low-profile balloon to enforce the stenosis enlargement. The sole PTA provided an immediate gain of the lumen patency, but a significant part of the patients experienced restenosis within a year and need further treatment. For this reason, the placement of a stent immediately after PTA was included in the percutaneous procedure, leading to a significant decrease of the acute reocclusion. This innovative device transformed the practice of interventional cardiology, selling 1 million units in less than 2 years and creating a $700 million annual market nearly overnight.

![Figure 12.3 Example of a Nitinol self-expanding stent for cardiovascular applications. The device is partially deployed during a free-expansion deployment. While the distal part has gained the original expanded configuration, the proximal part is still compressed within the delivery system.](image-url)
It is worth noting that the mechanisms underlining the Palmaz-Schatz stent and its
descendants are based on the local plastic deformation occurring during the balloon
inflation at the hinge points of the design. In this manner, the stent remains in an
open configuration also after the balloon deflation, counteracting the arterial elastic
recoil.

As soon as the use of stenting started to extend to peripheral vessels, characterized by
long lesions, tortuous anatomies, and complex kinematics, such plastic deformations
showed undesired drawbacks. The Nitinol features appeared to be a key factor in
building a new type of device, the so-called self-expanding stents.

The idea of using Nitinol to make a self-expanding cardiovascular device is not
certainly recent; in fact, in the 1980s, Charles Dotter used Nitinol coils as scaffolds for
arteries, showing the promising functional qualities of Nitinol for endovascular treat-
ments. Unfortunately, at that time, the manufacturing procedures were not advanced
enough and the mechanical qualities of Nitinol had not been completely understood
yet, so it took many years before the first Nitinol stents began to be marketed. At
the beginning of the 1990s, Nitinol was just available in wire form; therefore, Nitinol
stents were not very successful.

In the mid-1990s, laser cutting began to be performed on Nitinol tubes. The avail-
ability of Nitinol tubing made this material the first choice for self-expanding stents and
stent-grafts. Bard released the Memotherm in 1997, and Cordis released the SMART
stent in 1998. The SMART stent became the dominant design in the endovascular
marketplace and remains the leader today. The success of the SMART stent design
was largely due to its very fine mesh structure that offered exceptional contouring, flex-
ibility, and apposition characteristics. It is estimated that Nitinol stents now account for
60% of the endovascular stent market. Nitinol stents also dominate nonvascular markets
including urologic, upper and lower gastrointestinal, and trachea-bronchial applica-
tions. Currently, a number of self-expanding Nitinol stents have been marketed or
are under development.

Generally, stents for endovascular applications are made of V-shaped struts forming
circumferential rows, linked together by bridges. This ring configuration offers a good
compromise between the flexibility, strength, and capability of the design to get a small
diameter to accomplish the accommodation within the low-profile catheter. There are
various designs showing different link types between struts. The two main configurations
of ring type stents are the open-cell and the closed-cell ones. Struts can be connected
through peak-to-peak, peak-to-valley, or mid-strut connections. Virtually every combi-
nation of peak, valley, and mid-strut connections has been designed, and even hybrid
connections have been provided. The overall mechanical features of the stents are influ-
enced both by geometrical design details (e.g., type of ring “bridges,” strut width, length,
thickness, shape) and by the material processing. In this context, it is worth
mentioning the appealing initiative of an open-source design tool for a self-expanding
stent as recently proposed by Bonsignore and colleagues, who shared their experience as stent designers in a practical guide and resource for the design and analysis of a generic Nitinol stent.

The majority of self-expanding Nitinol stents are cut from tubing, but there are some exceptions, including wire-based and sheet-based manufacture. For example, NexStent (Boston Scientific), for use in patients with carotid artery disease, is made from a laser cut Nitinol sheet. The sheet is tightly rolled up for delivery into the catheter and can be deployed to adjust to a variety of diameters. Another exception is the IntraCoil (Sulzer IntraTherapeutics), a wire coil stent for the treatment of superficial femoral artery and popliteal artery lesions, based on the same principle of Dotter’s experiments.

Some stents made of wire coil are also available, but they never met with great success in vascular applications because they usually present a larger diameter in the constrained state than ring-type designs. Instead, wire coil stents are generally used in nonvascular applications (e.g., prostate and urethral stenting) because they are usually fully retrievable even weeks or months after implantation.

The main manufacturing method for Nitinol tubes is laser micromachining, but photochemical etching has also been demonstrated to be a viable procedure. It is worth mentioning that nano-manufacturing, a method implying high-vacuum sputtering, may be used to build metallic implants in an additive fashion.

To appreciate the importance of the stent-related market and the corresponding economical and social impact, we report some data about that. Coronary artery disease is a major cause of death around the world, so coronary stenting is a widely used technique. The global market for coronary stent devices reached $7.1 billion in 2011. By 2016, it is expected that total market value will reach $10.6 billion. The Americas region dominated coronary stent device revenues for 2011 with a 40% share. This segment was worth $2.8 billion in 2011 and is projected to reach $4.3 billion by 2016. Europe is the second largest shareholder, with 37% of the market for coronary stent devices. Its market size is expected to increase from $2.6 billion in 2011 to $3.4 billion in 2016.

The peripheral angioplasty procedures are expected to reach 3.3 million procedures by 2016, while coronary angioplasty procedures are estimated to reach 4.5 million procedures by 2016. The major driving factor for these markets would be the increasing demand for minimally invasive percutaneous endovascular treatment. The stents used in interventional procedures represent the largest segment of the market.

The market for stent grafts also appears to be promising. In fact, Medtech Ventures reported that approximately 700,000 new cases of aortic aneurysms are estimated to be diagnosed every year in the developed world. Given this data, the global market for aortic stent grafts was estimated at $1.8 billion in 2012, a growth of 12% over its 2011 level of $1.6 billion. The market has been growing at an average annual growth rate of 15% over the previous 5 years.
12.2.4 Others

12.2.4.1 Percutaneous Valves

Another frontier of cardiovascular surgery is the minimally invasive replacement of the heart valves, particularly of the aortic valve. In this context, Nitinol, with its capability to recover the initial shape after a severe deformation, is playing a key role. In 2002, Cribier performed the first clinical implant of a percutaneous balloon-expandable aortic valve at the level of the native valve; in 2004, Grube implanted for the first time a self-expandable transcatheter valve.

In the last decade, different devices have been designed and submitted for clinical evaluation (Fig. 12.4) confirming that such an innovative technique represents a promising solution for aortic stenosis. However, at present, it is still an immature procedure due to limited follow-up data and durability evaluation.

Two types of transcatheter devices are currently available: cobalt chromium balloon-expandable (Edwards Sapien XT) or nitinol self-expandable stent (Medtronic Corevalve). In the case of balloon-expandable devices, a balloon inflation leads to the valved stent expansion, which excludes and compresses the native diseased leaflets. The self-expandable valve placement procedure is very similar to the previous one, except for the fact that self-expandable prostheses automatically open through a stepwise deployment when gradually extracted from the delivery catheter.29

Figure 12.4 Transcatheter aortic valve prostheses currently used in clinical practice: the Medtronic CoreValve (left) and the Edwards SAPIEN XT (right). For more details, please refer to Ref. 29.
12.2.4.2 Coils
Coils are often used to treat cerebral aneurysms, which are localized dilations of the intracranial arteries. The coils are wires that are positioned into the aneurysm in a ball-shaped fashion, with the aim of inducing a clotting reaction, eliminating the risk of rupture. Traditionally, coils are made of platinum, but recently a coil made with a mixture of platinum and NiTi has been produced (ev3 Neurovascular, Irvine, CA), resulting in less stretching and better resistance to compaction with respect to other types of coils.\(^5\)

12.2.4.3 Clips for Cardiac Surgery
The Nitinol U-Clip (Coalescent Surgical, Inc, Sunnyvale, Calif) is a device that can be used for vascular sutures, exploiting the superelasticity of Nitinol, which provides an automatic closure of the clips. Studies on animal models have been performed. It has been demonstrated that Nitinol clips have the same effectiveness of conventional sutures when used for anastomosis, with the advantage that the procedure is faster because there is no necessity to tie knots.\(^30,31\)

Another type of SMA clip used in cardiovascular surgery is represented by the thermoreactive Nitinol clips used for sternal closure after cardiac operations. Because healing complications after cardiac surgery, such as dehiscence, osteomyelitis, mediastinitis, and superficial wound infection often occur and produce a significant mortality, it is apparent that the technique of sternal closure is an important factor to analyze. Thermoreactive Nitinol clips offer an improved method for sternal closure, which results in less postoperative complications and improves osteosynthesis due to the lighter forces produced on the sternum with respect to conventional closures.\(^32,33,34\)

12.2.4.4 Occlusion Devices
In order to treat patent ductus arteriosus or atrial and ventricular septal defects, which are anomalies that can reduce life expectancy, self-expanding occlusion devices made of Nitinol have been developed (e.g., AGA Medical Corp., LEPU medical). These devices have single or multiple lobes made of a fine Nitinol mesh that can be introduced in vessels through a catheter by a minimally invasive procedure.

The SMA occlusion device, made of SMA wires and a waterproof film of polyurethane, represents an alternative to open surgery, which is an extremely invasive treatment. The device is initially inserted through the hole in the septum by means of a catheter. It is partially deployed so that it covers the opening in the septum. Then, it is pulled against the opening and is further deployed from the catheter, creating a second cover on the other side of the opening.\(^35\)

12.2.4.5 Heart Surgery Instruments
Another cardiovascular application of Nitinol is an open-heart stabilizer, used to prevent regional heart movement while performing surgery. Superelasticity and constant force
plateau presented by SMAs are exploited also in the tissue spreader, used to spread fatty tissue on the heart.\textsuperscript{36}

\subsection*{12.2.4.6 Clamps}
The clamp technique, used to arrest bleeding and to control blood flow from the arteries, is essential in any surgical operation. SMA features allow a constant stress response over a large range of strains, thus allowing an innovative concept of clamping that is able to guarantee better control of occlusive pressure on the vessel wall.\textsuperscript{37}

\subsection*{12.2.4.7 Annuloplasty Band}
Mitral valve regurgitation is often due to structural defects of the valve, treatable with surgical intervention. Mitral valve repair, when possible, is preferred to mitral valve replacement because it is more effective, more durable, and does not require anticoagulant drug therapy. The main mitral valve repair techniques involve securing a full ring or partial band around the mitral annulus. This type of intervention also can be performed through minimally invasive approaches. Purser and colleagues\textsuperscript{38} developed a prototype ring composed of a NiTi SMA, a silicone sheath, and a polyester sewing cuff, which meets the requirement of being flexible enough to be deployed through an 8-mm trocar but stiff enough to provide benefits to the damaged mitral valve. The tradeoff between flexibility and stiffness is fulfilled by the NiTi core of the prototype, which is flexible if maintained at lower temperatures (in the martensitic phase), allowing for simpler deployment, placement, and suturing; also, it is semirigid over 37 °C (i.e., at normal human body temperature), thus providing the desired support to the mitral valve. Ex vivo trials performed by Purser and colleagues clearly demonstrated that the device could be easily implanted through robot-assisted surgery, showing a good durability and a positive reduction effect on mitral valve regurgitation.\textsuperscript{38,39}

\subsection*{12.2.4.8 Prosthetic Pump}
An early patent\textsuperscript{40} reported the design of a prosthetic pump intended for application as an artificial heart, which was able to reproduce the natural movement of contracting and relaxing muscles and whose contractile elements are formed of NiTi.

\subsection*{12.2.4.9 Nitinol Blades for Resecting Calcified Aortic Heart Valves}
Calcified aortic heart valves are often replaced by equine or porcine pericardial valves on a metal stent structure. However, the calcified layer may be noncircular, causing leakage, regurgitation, and even distortion of the installed artificial valve, resulting in reduction of effectiveness and durability of the implant. A device showing foldable Nitinol blades is presented by Hauck and colleagues,\textsuperscript{41} with the aim of resecting the degenerated aortic valve, in order to leave a proper circular geometry ideal for installation of the artificial
valve. The objective is not only the creation of better conditions for the artificial valve to operate, but also a reduction of the surgery intervention time and complications.

### 12.2.4.10 Snare

Conceptually simple SMA devices, called snares, were discussed in the previous chapter, as they are used in various surgical fields. In cardiovascular surgery, snares are used to remove foreign bodies in a minimally invasive fashion. Because of the flexibility and radiopacity of the material, Nitinol snares have been demonstrated to offer good torque control and grasping capacity, being able to remove various foreign bodies of different size. Such a device allows undesired obstructions to be eliminated from vascular conduits in a quick and safe manner.  

### 12.3 EXAMPLES OF COMPUTER-BASED DESIGN

Finite element analysis (FEA) was developed more than 70 years ago to solve complex elasticity and structural analysis problems in civil and aeronautical engineering. Thanks to the significant advances of computational facilities, the use of FEA in biomechanics has expanded. The application of various modeling and simulation tools in medicine and clinical translational research has been proposed by a diverse group of investigators, posing the problem of the homogenization of methods and result reporting. In this scenario, the FEA-based simulation of SMA-based cardiovascular implants to assess the structural performance of various types of devices is steadily increasing. In particular, the simulations, in addition to traditional in-vitro techniques, are used to improve the design and testing of medical devices, such as cardiovascular stents. In fact, to date, these devices have been mostly developed using a trial-and-error approach: a first prototype is manufactured and physical tests are performed to check whether the design criteria are fulfilled. If this is not the case, the design is adapted and a new prototype is manufactured and tested. Unfortunately, this approach is time-consuming, expensive, and often not able to fully address the product’s performance and (bio)mechanical requirements. Moreover, performing physical tests on these generally small devices is challenging. A promising strategy to design medical devices is virtual product development, which enables the development and optimization of novel designs and consequently reduces the costs and the time to market.

Starting from such considerations, the goal of this chapter is to provide a comprehensive overview of the use of structural FEA to analyze and optimize SMA-based cardiovascular devices. We have organized the chapter by categorizing the literature review with respect to the vascular region treated by the analyzed devices. A preliminary section is dedicated to the collection of studies addressing the mechanical response of a given SMA-based device in a general environment. In conclusion, we discuss the research perspective suggested by the review.
12.3.1 General-Purpose Studies

Most of the studies addressing the use of FEA to predict or optimize SMA-based cardiovascular devices are focused on self-expanding stents. It is easy to understand that the steady increase of minimally invasive approaches has led to an important commercial interest in this sector, calling for engineering tools supporting the design of new and competing devices.

One of the pioneering works about Nitinol stent simulation is the 1997 study by Whitcher,\(^46\) who proposed the use of FEA to estimate the structural behavior of Boston Scientific’s Symphony stent under in-vivo loading conditions. The study presents a lot of simplified assumptions. For example, a von Mises-yield criterion elastoplastic model was adopted for the constitutive modeling of Nitinol, whereas the in-vivo loading conditions are replicated through the application of a pressure load on a stent portion, thus neglecting the inclusion of the arterial model.

A few years later, in 2000, Rebelo and Perry\(^47\) discussed the use of FEA to simulate the expansion of Nitinol stent implementing for pseudo-elasticity, the constitutive model originally proposed by Auricchio and Taylor,\(^48,49\) based on the concept of generalized plasticity.\(^50\) In 2002, the study was extended to evaluate the fatigue resistance of the stent by Perry and colleagues.\(^51\)

A further insight on the fatigue of self-expanding stent components was provided in 2003 by Pelton and colleagues,\(^52\) who quantified the cyclic deformation behavior of superelastic Nitinol in order to calculate design safety factors for medical devices. In particular, the study combines experiments and numerical analysis. The authors performed displacement-controlled fatigue testing on laser-cut stent-like devices, while the fatigue strains were calculated from displacements with nonlinear FEA methods. Surprisingly, the results demonstrate that the oscillating strain amplitude is the main contributor to fatigue behavior.

More fundamental work regarding the constitutive modeling of SMA was done in 2004 by Jung et al.\(^53\) and Auricchio and Petrini,\(^54\) who developed constitutive models suitable for numerical analysis, such as FEA, with particular attention to the robustness of the algorithms in order to facilitate the analysis of real industrial applications, such as the design of medical stents.\(^55\)

A significant attempt to tailor the design of a novel self-expandable stent by the use of FEA was proposed in 2006 by Theriault and colleagues,\(^4\) who discussed the development of a Nitinol stent with a progressive expansion device made of polyethylene, allowing smooth and gradual contact between the stent and the artery’s wall by creep effect. The Nitinol is modeled with a superelastic law and the study presents two simulations. The first simulation determines the final geometry of the stent laser cut from a small tube, whereas the second simulation examines the behavior of the prosthesis during surgery and over the 4 weeks following the operation.
In 2008, Kim et al. discussed the mechanical modeling of self-expandable stents fabricated using braiding technology. For this purpose, they proposed a finite element model by coupling a preprocessing program for the three-dimensional geometrical modeling of the braiding structure of the stent. The Nitinol wires of the stent were assumed to be superelastic and their mechanical behavior was incorporated into the finite element software through a user material subroutine employing a one-dimensional super elastic model.

In 2010, given the growing interest toward an exhaustive modeling of SMA macroscopic behavior due to their extensive use in a number of applications in many fields of engineering, particularly in biomedical engineering, Auricchio et al. reviewed the properties of a robust three-dimensional model able to reproduce both pseudo-elastic and shape-memory effect. In particular, the model parameters are calibrated with respect to experimental data. The model was used to perform the finite element analysis of pseudo-elastic Nitinol stent deployment in a simplified atherosclerotic artery model (Fig. 12.5).

In 2011, Rebelo and colleagues proposed a comparative study that grounds its motivation on the statement that FEA of Nitinol medical devices has become...
prevalent in the industry. Starting from this idea, the authors presented a study in which some commonly made assumptions in FEA of Nitinol devices were verified. The base model pertains to the simulation of the fabrication of a diamond-shaped stent specimen, followed by cyclic loading, which is being used by a consortium of several stent manufacturers focused on the assessment of fatigue-life prediction for Nitinol devices.

In 2012, Azaouzi et al. proposed a simplified model of a Nitinol stent in order to perform FEA of the stent deployment as well as the pulsatile loading inside the artery. The aim was to provide a tool to forecast the fatigue life of Nitinol stents and to optimize the design phase.

Similarly, in the same year, Garcia and colleagues studied the influence of the main geometrical parameters on the radial force of a self-expanding stent. Using FEA, they performed a parametric analysis of a commercial stent model (Acculink, Abbot Vascular), which was developed to estimate the influence of geometrical variables on the stent radial expansion force. As result of the optimization, the authors proposed a new stent design with variable radial stiffness, which was virtually implanted on both healthy and atheromatous vessels to evaluate its effectiveness.

Three other studies targeted the stent design optimization using FEA. Khalil-Abad and colleagues proposed a planar lattice free of stress concentrators for the synthesis of a stent with smooth cell shapes. The authors discussed a design optimization to minimize the curvature and reduce the bending strain of the elements defining the lattice cells, resulting in a novel cell geometry with improved fatigue life and radial supportive force suitable for Nitinol self-expandable stent-grafts.

One year later, in 2013, Hsiao and Yin proposed the key idea to shift the highly concentrated stresses/strains away from the stent crown and redistribute them along the stress-free bar arm by tapering its strut width. The authors used FEA to evaluate the mechanical integrity and pulsatile fatigue resistance of the stent to various loading conditions, proving an increase of the fatigue safety factor when compared to the standard stent with constant strut width.

In the same year, Azaouzi et al. targeted the objective to optimize the stent design by reducing the strain amplitude and mean strain over the stent, which are generated by the cyclic pulsating load. They proposed an optimization-based simulation methodology in order to improve the fatigue endurance of the stent. The design optimization approach is based on the response surface method, which is used in conjunction with Kriging interpolation and the sequential quadratic programming algorithm.

A more simplified analysis has been proposed by Nematzadeh and Sadrnezhad, who used FEA to investigate the effect of crimping and $A_t$ of Nitinol on mechanical performance of a Z-shaped open cell. Their study shows that low $A_t$ Nitinol has better mechanical and clinical performance due to small chronic outward force, large radial resistive force, and appropriate superelastic behavior.
12.3.2 Carotid Artery

The use of self-expanding Nitinol stents for the minimally invasive treatment of carotid stenosis represents an important share of the market of SMA-based medical devices. In fact, the world market for carotid stent systems, which was valued at $150 million in 2010, is expected to double by 2015. Abbott Vascular, Cordis, Boston Scientific, Medtronic, and Covidien have transformed the competitive landscape of the market by bundling their carotid stents with their embolic protection devices as complete carotid stent systems.66

Despite such a significant industrial interest in this specific sector, few studies are addressing the use of FEA to design or optimize the carotid stent design. The first study in this direction was done in 2007 by Wu et al.,67 who evaluated biomechanical properties of Nitinol carotid stents and their interactions with carotid arteries through FEA. Wu and colleagues adopted a geometrical idealization of the carotid bifurcation, simulating the implant of a segmented-design Nitinol stent and accounting also for the delivery sheath.

In 2011, Auricchio and colleagues68 further extended the study by Wu et al. to a patient-specific case. The authors used FEA to evaluate the performance of three different self-expanding stent designs in the same carotid artery model, based on computed angiography tomography images. In particular, adopting the simulation strategy presented in Conti et al.,69 the authors defined six stent models considering the three designs in different sizes and configurations (i.e., straight and tapered), evaluating the stress induced in the vessel wall, the lumen gain, and the vessel straightening. The study represents a first step toward a quantitative assessment of the relationship between a given carotid stent design and a given patient-specific carotid artery anatomy.

The same computational framework was subsequently experimentally validated70 and used to assess the impact of the carotid stent design on its capability to scaffold the artery,71 with particular attention to realistic modeling of the artery (Fig. 12.6).72

![Figure 12.6](image-url) Elaboration of computed tomography angiography images resulting in the three-dimensional reconstruction of both lumen of left carotid bifurcation and plaque (on left). Results of simulation of carotid artery stenting; the stent crimped within the catheter is virtually deployed by means of FEA in a patient-specific carotid artery model. Adapted from Ref. 72.
12.3.3 Aorta

The rapid evolution of self-expandable stent-grafts in the last decade also motivates the use of FEA in this emerging sector. One of the first investigations in this area was presented in 2008 by Kleinstreuer et al., who discussed a finite element analysis of tubular, diamond-shaped stent grafts under representative cyclic loading conditions for abdominal aortic aneurysm repair. In particular, the authors studied the mechanical behavior and fatigue performance of different materials found in commercially available stent-graft systems, evaluating and comparing the effects of crimping, deployment, and cyclic pressure loading on stent-graft fatigue life, radial force, and wall compliance through the numerical simulations.

More recently, De Bock and colleagues experimentally validated the use of FEA to virtually deploy a bifurcated stent graft (Medtronic Talent) in a patient-specific model of an abdominal aortic segment. The authors modeled the entire deployment procedure, with the stent graft being crimped and bent according to the vessel geometry and subsequently released. The validation of numerical procedure was performed comparing the simulation outcomes with the in vitro data regarding the placement of the device in a silicone mock aneurysm, imaged by high-resolution computed tomography.

In the same year, Demanget et al. numerically simulated the bending of two manufactured stent grafts (Aorfix by Lombard Medical and Zenith by Cook Medical Europe) using FEA. The authors studied the global behavior of the stent grafts by assessing stent spacing variation and cross-section deformation. The study is motivated by the potential relationship between the clinical complications and the insufficient stent-graft flexibility, especially when devices are deployed in tortuous arteries.

The same authors validated the numerical procedure in a subsequent study, where the two commercially available stent grafts were subjected to severe bending tests. Their three-dimensional geometries in undeformed and bent configurations were imaged from X-ray microtomography. The images were elaborated to set up stent-graft numerical models, subjected to the boundary conditions measured experimentally. The computational framework was further used to numerically assess the flexibility and mechanical stresses undergone by stents and fabric of currently marketed stent-graft limbs (Aorfix, Anaconda, Endurant, Excluder, Talent, Zenith Flex, Zenith LP, and Zenith Spiral-Z).

In 2013, Auricchio and colleagues described the use of a custom-made stent-graft to perform a fully endovascular repair of an asymptomatic ascending aortic pseudoaneurysm in a patient who was a poor candidate for open surgery (Fig. 12.7). The authors also discussed the possible contribution of a dedicated medical images analysis and patient-specific simulation as support to procedural planning. In particular, the authors have compared the simulation prediction based on preoperative images with postoperative outcomes. The agreement between the computer-based analysis and reality demonstrated by this study further encourages the use of FEA-based simulations, not only as a tool for device designers but also as a procedural planning tool for the physicians.
12.3.4 Intracranial Artery

As in the case of other vascular districts, the endovascular treatment of cerebral aneurysms using stents has advanced markedly in the last decade. Intracranial stents must be very flexible longitudinally and have low radial stiffness: although there are a number of designs of intracranial stents, there are very few studies examining the stress distribution and deformation of cerebrovascular stents using FEA and/or experiments.

One of the pioneering works in this field is the study by Shobayashi et al., who investigated the relationship between stent mesh patterns and the mechanical properties of different variants of design units of the stent, the so-called cells, evaluating several mesh

Figure 12.7 Segmentation of pre and postoperative images regarding the implant of self-expanding endograft in the ascending aorta. The preoperative images are used as input for the simulation of the implant, which resemble the reality for both the qualitative and quantitative point of view. For more details, please refer to Ref. 73.
patterns through FEA. It is worth noting that the superelastic behavior of Nitinol was approximated by a homogeneous, isotropic, and elastic—plastic material using the Von Mises plasticity model. Moreover, the adopted loading conditions were idealized, neglecting the actual conformation of the cerebral vascular anatomy.

More recently, De Bock and colleagues\(^\text{80}\) overcame such a limitation using finite element analysis to perform nine virtual stenting procedures—that is, the virtual deployment of three different Nitinol stent designs in three patient-specific cerebral aneurysmatic vessels. The authors evaluated the performance of the devices assessing the percentage of strut area covering the aneurysm neck, the straightening induced on the cerebrovascular tree by the stent placement (i.e., the reduction of vessel tortuosity), and stent apposition to the wall, quantified as the percentage of struts within 0.2 mm of the vessel.

### 12.3.5 Superficial Femoral Artery and Renal Artery

The superficial femoral artery (SFA) runs from the hip to the knee, through muscle and joints. It is the main blood supplier for the leg, so it undergoes severe changes in geometry associated with leg movement.\(^\text{81}\) As in other vascular districts, SFA and its branches, such as the popliteal artery, located below the knee, can suffer atherosclerosis or develop aneurysms. Thanks to the successful use of self-expanding Nitinol stents in other anatomical conduits, SFA has also been targeted as a candidate for revascularization through stenting. In fact, several clinical reports and trials evaluated the effect of stents in the SFA, proving that for long lesions the use of stents is beneficial compared to balloon angioplasty alone. This result has opened the door to a widespread use of stents in the SFA but, unfortunately, there are still concerns about the long-term results of the implants because of their nonnegligible fracture rate (1.8–18%, see Ref. \(^\text{82}\) and references therein). In order to identify the potential fracture risks of currently available or new SFA stent designs, a number of theoretical, numerical, and experimental studies have been performed (Fig. 12.8). In this section, we just mention the most recent literature, with a particular focus on the use of structural FEA as an investigation method.

In 2009, Rebelo and colleagues\(^\text{84}\) performed simulations of peripheral stenting. They obtained realistic artery geometry from magnetic resonance imaging scans of a patient in the fetal and supine positions and comparing deformation of the stent deployed into the two different artery configurations. Despite that the study accounts for the main issues of stenting modeling, the authors consider a small portion of the superficial femoral artery far from the popliteal artery and consequently the configuration change of the artery appears to be fairly limited.

In 2011, Harvey\(^\text{84}\) evaluated through FEA the fatigue performance of two stent geometries. Two vessel models were used: a constant diameter straight tube and a segment of the superficial femoral artery obtained from a computed tomography angiography dataset. The author imposed two types of loading to the stent model through the
vessel model. Although the study proves the utility of combining advanced nonlinear finite element simulations and fatigue predictions for the design of peripheral Nitinol stents, the imposed loading conditions are rather idealized.

In 2012, Petrini and colleagues presented an integrated numerical and experimental approach to foresee Nitinol stent fatigue fracture, also reproducing a wide range of in vivo conditions. This study was further integrated by Meoli and colleagues, who used FEA coupled to fatigue analysis to investigate two in vitro set-ups proposed in the literature and to increase the understanding of fatigue behavior of commercial Nitinol stents. Their results indicate that the two investigated testing conditions produce quite different fatigue behavior both in terms of constant-life diagram and strain distribution in the stents. However, also in these studies, the imposed loading conditions are rather idealized, leaving space for further developments embedding patient-specific loading conditions derived by medical image.

12.3.6 Heart Valves

As discussed before, the recent developments in percutaneous implantation to replace heart valves in a minimally invasive fashion through stent-mounted prosthetic valves have opened another wide field of application for the unique features of Nitinol. Clearly, also in this specific sector, structural FEA has been used to assess and optimize new SMA-based devices, as briefly discussed in this section.

In 2004, DeHerrera and colleagues numerically investigated the mechanical adequacy of a new valve design with a flat-sheet based stent, which could be rolled up to about 20 French (6.66 mm outside diameter) and delivered percutaneously.
The same research group proposed a similar approach to analyze a new percutaneously-delivered device (at that time under development at Edwards Lifesciences) that induces mitral valve reshaping. The device is intended to be placed in the coronary sinus and is basically a spring with anchoring stents at each end. In particular, the FEA study examined the mechanical behavior of one of the anchoring stents, from its forming from a small diameter tube to its deployment into the coronary sinus.

Some years later, in 2013, Tzamtzis et al. presented a numerical study on the different mechanisms responsible for the radial force exerted on the aortic annulus by self-expanding and balloon-expandable prostheses. In particular, the authors simulated and compared the mechanical behavior of the Medtronic CoreValve (self-expanding) and the Edwards SAPIEN (balloon-expandable) devices. In the case of the self-expanding valve, the radial force was essentially dependent on the diameter of the left ventricular outflow tract, which is the heart region where the valve is anchored.

In the same year, Kumar et al. published a study using computational modeling and simulation to design a new Nitinol-based mitral valve stent and to evaluate its crimpability and fatigue behavior. A self-expandable stent with new features, addressing issues of valve migration and paravalvular leaks, was proposed; its expansion, crimpability, deployment patterns, and fatigue behavior was simulated and analyzed. Moreover, the proposed simulations also embed cyclic cardiac muscle loading, cyclic blood pressure loading, and cyclic valve leaflet forces in the fatigue life assessment for mitral valves.

12.4 CONCLUSIONS

The SMA appliances described in this chapter have addressed the urgent need for ever-increasing innovation and reliability posed by the cardiovascular field. As we have seen, in spite of the nonlinearities presented by the peculiarity of SMA behavior, the development of more and more sophisticated cardiovascular devices has been achieved, thanks to the computational help provided by software simulation.

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