Function and Performance of Nitinol Vascular Implants

Andreas Melzer[1,*] and Dieter Stoeckel[2]

[1]*University of Dundee, Institute for Medical Science and Technology IMSaT, Scotland*

[2]*NDC, Nitinol Devices & Components, Inc., Fremont CA, USA*

Abstract: Nitinol, an intermetallic compound of Titanium and Nickel has found widespread use as material for vascular implants. Its unusual stress hysteresis and superelasticity makes these devices biomechanically compatible with the body's structures. Nitinol also exhibits excellent corrosion resistance, biocompatibility and MR compatibility. The paper focuses on the material specific properties of self expanding stents to treat peripheral vascular disease, like biased and temperature dependent stiffness, kink resistance, stent durability and fatigue, and describes the design aspects of different Nitinol stents. The function and performance of other Nitinol implants is also explained, like vena cava filters, cardiac septal repair devices, and self-expanding heart valve prosthesis. A separate section describes the performance of endovascular Nitinol implants in Magnetic Resonance Imaging, MRI.

Keywords: Nitinol, thermal shape memory, superelasticity, self expanding stents, MR compatibility, vascular implants.

1. NITINOL BASED CARDIOVASCULAR IMPLANTS

1.1. Introduction

Nitinol (Nickel-Titanium) alloys have become the materials of choice for self-expanding stents, vascular filters, graft support systems, heart valve frames, occlusion devices, and various other devices for minimally invasive interventional and endoscopic procedures. Most medical device companies now offer products, the performance of which is based on the highly unusual properties of these materials.

Nitinol alloys are most commonly known for their superelasticity and thermal shape memory. While the term shape memory is used to describe the phenomenon of restoring a predetermined shape through heating, after having "plastically" deformed that shape, the term superelasticity refers to the enormous elasticity of the alloys. It can be ten times more than the elasticity of the best stainless steels used in medicine today, and follows a non-linear path, characterized by a pronounced hysteresis. Shape Memory as well as superelasticity are the results of a solid state transformation, which can be triggered thermally or mechanically, and is dependent on the composition and processing history of the material. An in-depth description of the atomistic mechanisms of this transformation and the unique properties derived from it can be found in [1]. In this section we will describe important device characteristics that can be attributed to the specific properties of Nitinol, particularly as they relate to self-expanding implants like stents and filters.

1.2. Superelasticity and Shape Memory in Nitinol

Conventional implant materials, like stainless steel or cobalt based alloys, exhibit a distinctly different elastic deformation behavior from that of the structural materials of the living body. The elastic deformation of these metals and alloys is limited to approximately 1% strain, and elongation typically increases and decreases linearly (proportionally) with the applied force. In contrast, natural materials, like hair, tendon and bone can be elastically deformed, in some cases, up to 10% strain in a non-linear way [2]. When the deforming stress is released, the strain is recovered at lower stresses. As shown in Fig. (**1**), the loading/unloading cycle is characterized by a pronounced hysteresis.

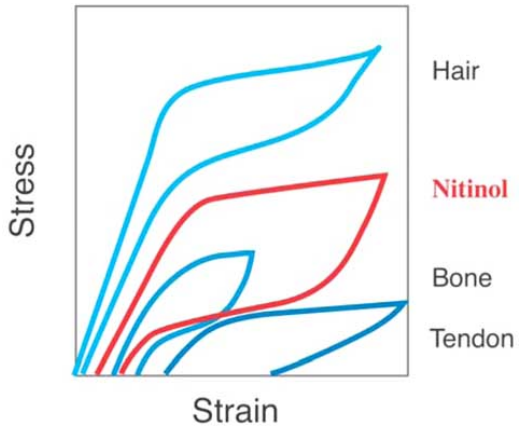

Fig. (1). Natural materials, like hair, tendon and bone can be elastically deformed, in some cases, up to 10% strain in a non-linear way. When the deforming stress is released, the strain is recovered at lower stresses and the loading/unloading cycle is characterized by a pronounced hysteresis. (Modified from Shabalovskaya et al. [2]).

A similar behavior is found with Nitinol alloys, equiatomic or near-equiatomic intermetallic compounds of titanium and nickel. Fig. (**2**) shows a characteristic stress/strain curve for a Nitinol alloy wire at body temperature. As with natural materials, the loading and unloading curves show plateaus, along which large deflections (strains) can be accumulated on loading, or recovered on unloading, without significant increase, or decrease, respectively, in loads (stress). Because deformation of more than 10% strain can

*Address correspondence to this author at the University of Dundee, Institute for Medical Science and Technology IMSaT, 1 Wuerzburg Loan, Dundee DD2 1FH, UK; Tel: +441382388355; E-mail: a.melzer@dundee.ac.uk

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be elastically recovered, this behavior is called "superelasticity".

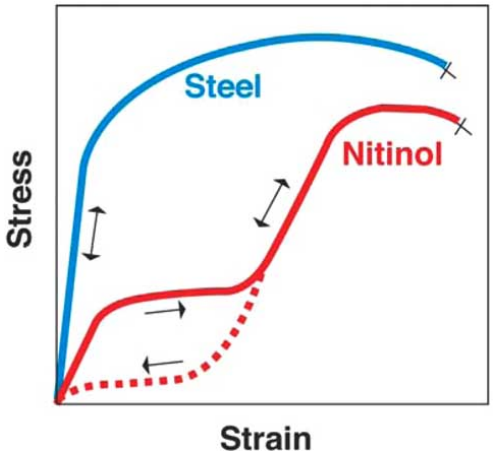


Fig. (2). Characteristic stress/strain curve for a Nitinol alloy wire at body temperature in comparison to steel. Deformation of more than 10% strain can be elastically recovered. This behavior is called "superelasticity".

Superelastic Nitinol appears macroscopically to be simply very elastic. However, the mechanism of deformation is quite different from conventional elasticity, or simply stretching of atomic bonds. When a stress is applied to Nitinol, and after a rather modest elastic deformation, the material yields to the applied stress by changing its crystal structure. This "stress induced" phase transformation allows the material to change shape as a direct response to the applied stress. When the stress is removed, the material reverts to the original structure and recovers its original shape. While superelasticity is the result of a stress induced phase transformation, shape memory is the result of a thermal phase transformation. In fact, when superelastic Nitinol is cooled to below a critical temperature (the transformation temperature, which is dependent on alloy composition and processing history), it also changes its crystal structure. If no force is applied, this phase change is not accompanied by a shape change. The material can be plastically deformed in the "low temperature phase", but the original shape can be restored by heating above the transformation temperature.

The most unusual property of Nitinol alloys is the stress hysteresis mentioned earlier. While in most engineering materials stress increases linearly with strain upon loading and decreases along the same path upon unloading (as shown in Fig. (**2**) with steel for comparison), Nitinol exhibits a distinctly different behavior. After an initial linear increase in stress with strain, large strains can be obtained with only a small further stress increase. This is called the loading plateau. The end of this plateau is reached at about 8% strain. Unloading from the end of the plateau region, causes the stress to decrease rapidly until a lower plateau ("unloading plateau") is reached. Strain is recovered in this region with only a small decrease in stress. The last portion of the deforming strain is finally recovered in a linear fashion.

The stress hysteresis or path dependence of Nitinol results in a device feature termed biased stiffness [3], which is particularly interesting for the function and performance of self-expanding stents. It allows a deployed stent to exert a low outward force against the vessel wall, termed chronic outward force, COF, but to resist recoil pressures or any other external compression forces with forces that are much higher. These forces are called *radial resistive forces* or RRF [4]. The radial resistive force is determined by the loading section of the stress/strain curve, while the chronic outward force is given by the unloading section. Both forces influence the clinical performance of stents.

1.3. Self-Expanding Stents

Self-expanding Nitinol stents have improved the treatment of peripheral vascular disease. They represent one of the largest volume users of superelastic/shape memory alloys. Nitinol stents are manufactured to a size slightly above the vessel diameter. They are crimped and constrained in a delivery catheter. At the treatment site the constraint is removed and the stent expands elastically (Fig. **3**) until it hits the vessel wall, where it continues to exert a gentle outward force, COF, as described above. The unusual plateau-like elastic behavior of Nitinol (almost constant stress over a large deflection) allows the continuing opening force of the stent acting on the vessel wall to remain very low even through large deflections and oversizing of the stent (it is typically recommended to use a size of a stent that is 1-2 mm larger than the inner diameter of the target vessel). Meanwhile the forces generated by the stent to resist compression, RRF, increase rapidly with deflection until the plateau stress is reached. Although most self-expanding stent placements are preceded by a percutaneous transluminal balloon angioplasty, there are indications that the chronic outward force of a Nitinol stent placed without previous PTA causes the vessel to remodel with less intimal hyperplasia than if PTA is performed prior to stenting [5].

Fig. (3). Self expanding Nitinol stents are implanted in arteries with stenosis to reestablish blood flow. The stents are Laser cut form tube, expanded and then crimped and constrained in a delivery catheter.

Another unusual feature of Nitinol stents is their *temperature dependent stiffness*. Most stents are produced with a transition temperature of 30°C (to allow for full expansion at body temperature). They feel quite weak when squeezed or crushed at room or lower temperature. In contrast, they feel much stiffer when squeezed at temperatures above 30°C.

Quantitative data show that the chronic outward force actually doubles when the temperature is increased from 20 to 37°C. As mentioned before, the manufacturer can adjust the transition temperature of the stent to a certain extent during processing. This gives the designer another option to increase or decrease the radial forces of the stent without changing the design or physical dimensions, as for each degree that the transition temperature is below body temperature, the loading and unloading forces increase by approximately 4 N/mm^2.

The unusual elasticity of Nitinol allows stents to be kink resistant. This is an important feature of Nitinol for stents in superficial vessels that could be deformed through outside forces. The carotid artery is a prime example. There is a perceived risk for balloon-expandable stents in carotid arteries to be permanently deformed through outside pressure resulting in a partially or completely blocked vessel, once the buckling strength of the stent is exceeded. Although Nitinol stents typically do not have the buckling strength of stainless steel or cobalt-chrome stents, they cannot be permanently deformed through outside forces. Nitinol stents can be completely compressed (crushed) flat and will return to their original diameter when the deforming force is removed (Fig. **4**). A quantitative analysis of the forces relevant to the performance of superelastic stents can be found in [4].

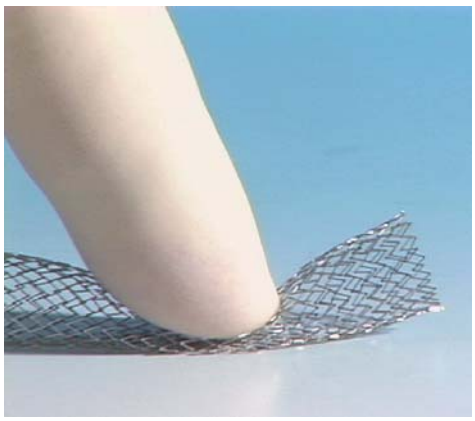


Fig. (4). Nitinol stents can be completely compressed (crushed) flat and will return to their original diameter when the deforming force is removed.

The goal of stenting is clearly to maintain patency of the vessel lumen for as long as possible. To accomplish this goal, a multitude of design parameters have to be considered that may affect the clinical performance of the stent. Amongst those are strength and stiffness (radial resistive and chronic outward forces), scaffolding (window size and shape), conformability, flexibility during delivery as well as after deployment, fatigue durability in pusatile and non-pulsatile deformation modes, crush recoverability, delivery profile, expansion ratio, oversizing strategy, forshortening, radiopacity, and last but not least biocompatibility and corrosion resistance. There is no clear indication that any one of those parameters has a more pronounced impact on clinical performance than the others. In the end, it is the combination of a many aspects that influences stent performance. As all design parameters are interrelated, a change in one input results in changes in many outputs, requiring compromises in optimization [6].

It is probably fair to state that until recently, stents were designed mostly by engineers with only limited understanding of the anatomical environment and the physiological and biomechanical peculiarities of the vasculature using anecdotal input from physicians. Only after stent fractures had been observed in certain clinical trials [7], did the medical device industry change its approach to stent design and now tries to understand the dynamics of the vasculature [8]. The Superficial Femoral Artery (SFA), for example, is a particulary "dynamic" artery, running from the hip to the knee, through muscle and joints. It undergoes severe changes in geometry associated with leg movement. Duda, *et al.* in the Sirocco Trial observed fracture rates of 17% at 6 months and 26% at 18 months in SFAs stented with the Cordis SMART Stent. The clinical results of stenting still were considered excellent, despite the fractures. Scheinert, et al subsequently confirmed that the SMART Stent maintained patency even with broken struts. These authors, however, found that other stent designs did not fare as well as the SMART Stent with respect to patency or restenosis rates [9]. Thus, it can be concluded that a fracture does not necessarily equate to device failure. A fracture in some designs leads to a high probability of clinical failure, while fractures in others do not. Nevertheless, stent durability and fatigue are major considerations in stent development today.

Worldwide, there are probably close to 100 different Nitinol Stent designs being marketed or in evaluation. Table **1** shows a list of self-expanding Nitinol stents currently available in the US and their approved indications. Basically all stents are FDA approved for the treatment of biliary blockages or obstructions. Although this is a non-vascular application, most stents have been and are used off-label to treat peripheral vascular disease while the device companies try to get approvals for specific indications. So far, only a limited number of Nitinol stents are approved in the US for specific vascular indications, like the carotid, iliac, or superficial femoral arteries [10].

The evolution of Nitinol stent designs is clearly linked to the development of the material itself. Early on, Nitinol was only available in wire form. The first to experiment with Nitinol wires was Charles Dotter in the early nineteen eighties [11]. He used shape memory Nitinol wire coils as intra-arterial scaffolds. A coil wound to a small diameter and delivered through a catheter into the vessel, would expand to a larger diameter, e.g. the diameter of the vessel lumen, upon warming with 60°C saline solution. Although the shape memory effect looked like ideally suited for the scaffolding of vessels, it took many more years for Nitinol stents to appear in the market. Dotter clearly was ahead of his time. The melting and processing of Nitinol had not been fully developed with consistent quality, nor had the properties of this material been fully understood. Today, 25 years after Dotter's experiments, all Nitinol stents are self-expanding without the need for post-deployment heating.

Since Nitinol tubing became available in the early nineteen nineties, most stents are produced by laser cutting a pattern from a seamless Nitinol tube with a diameter close to the delivery size. Subsequently, the cut part is expanded to the final stent diameter (slightly larger than the target vessel diameter) in a sequence of forming and heat-treating steps,

Table 1. Approval Status of Some Nitinol Self-Expanding Stents in the USA

Company	Stent	Indication				
		Biliary	Carotid	Iliac	SFA	Other
Abbott	Xact	X	X			
Abbott	Xceed	X				
Abbott	Xpert	X				
Abbott	JoStent SelfX	X				
Bard	Fluency	X				
Bard	Luminexx	X				
Bard	Conformexx	X				
Bard	Vivexx		Trial			
Bard	LifeStent NT	X			X	
BSC	Sentinol	X				
BSC	Symphony	X				
BSC	NexStent		X			
Cook	Zilver	X			Trial	
Cordis	SMART	X		X	Trial	
Cordis	Precise	X	X			
ev3	Intracoil	X			X	
ev3	Protégé	X	X			
Gore	Viabahn	X			X	
Gore	Viatorr	X				TIPS
Guidant	Absolute	X				
Guidant	Acculink		X			
Guidant	Dynalink	X				
Medtronik	Aurora	X				
Medtronik	Bridge SE Aurora	X				
Medtronik	Exponent		pending			

which also control the transformation temperature. In a final step, the stent is electropolished to create a smooth and biocompatible surface. For delivery, it is crimped back to the original tubing size and constrained in a delivery system. This is typically done at temperatures below the transformation temperature of the stent.

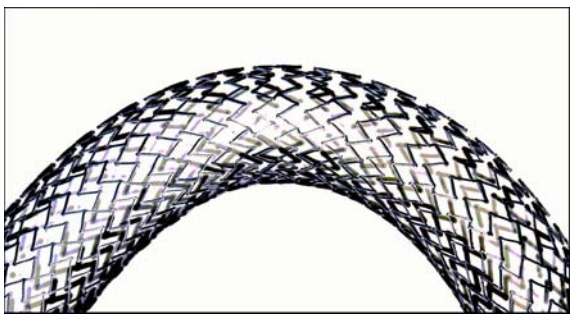


Fig. (5). The Cordis SMART stent was the first self-expanding stent that employed a "micromesh" design for improved scaffolding and adaption to curved vessels.

Early examples of tube-based laser-cut stents are the Angiomed (Bard) Memotherm and the Scimed (BSC) Radius stents. The Memotherm was a rigid, closed-cell design with a diamond shaped pattern similar to the original Palmaz balloon expandable stent. The Radius, on the other hand, is a flexible open-cell design with sequential rings and periodic peak-to-peak non-flex bridges. The advancement in laser cutting techniques and other processing technologies during the last few years made possible the production of Nitinol stents with small feature sizes, and Nitinol tubes with diameters as small as 0.3 mm can be processed. The Cordis SMART (Fig. **5**) stent was the first self-expanding stent that employed a "micromesh" design for improved scaffolding and prevention of tissue prolapse. Most of today's Nitinol stents employ variations of these basic design features (Fig. **6**). However, despite the similarities in design, different stents can yield different clinical outcomes, and it is still hard to tell what exactly makes one stent better than the other in clinical use. The most widely used Nitinol stent is the SMART stent (Cordis) with more than one million patients treated. Despite reported fractures, this stent clinically per-

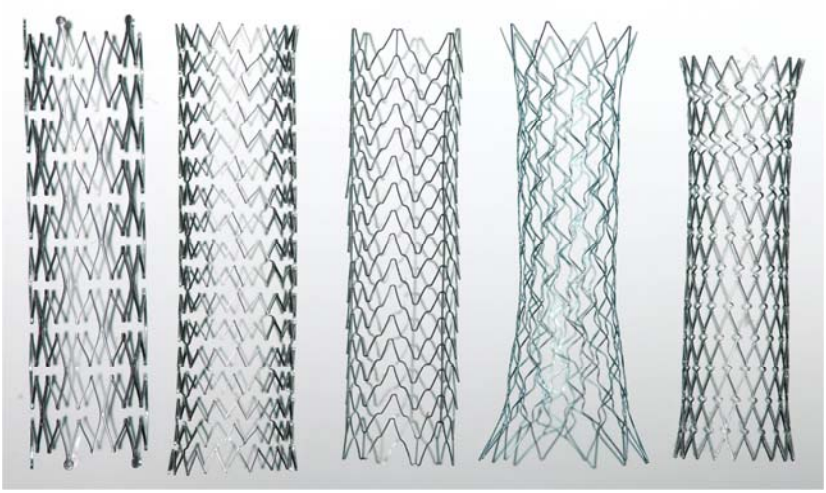


Fig. (6). Today's Nitinol stents employ variations of the basic design features introduced by the SMART stent in 1998.

forms well even in challenging environments. A more fracture resistant stent is the LifeStent Flexar (Bard), which was recently approved by the FDA for the treatment of occlusive disease in native superficial femoral arteries (SFA) and proximal popliteal arteries. Positive data were also recently reported from a trial using the drug (paclitaxel) coated Cook Zilver stent in the superficial femoral artery [12].

Laser-cut stents from Nitinol tubing are being produced in a wide range of sizes. Neurovascular stents like the Wing-Span Stent (Boston Scientific) are indicated for use in improving cerebral artery lumen diameter in patients with intracranial atherosclerotic disease [13]. "Vascular reconstruction devices" like the Enterprise device (Cordis) are intended for use with embolic coils for the treatment of wide-neck, intracranial, saccular or fusiform aneurisms in vessels of 3 to 4 mm diameter [14]. At the other end of the spectrum are aortic stents stents and stent grafts with diameters of > 30 mm.

Although most self-expanding Nitinol stents are cut from tubing, there are a few exceptions. In 2006, for example, the FDA approved Boston Scientific's NexStent for use in patients with carotid artery disease [15]. The NexStent is made from a laser cut Nitinol sheet that, for delivery in a catheter, is tightly rolled up. After being released from the delivery system, it "un-coils" and can adapt to multiple diameters in tapered or non-tapered vessel configurations. In 2002, the FDA had approved a simple wire coil stent, the IntraCoil (then Sulzer IntraTherapeutics) for the treatment of patients with superficial femoral artery and popliteal artery lesions [16]. The concept is similar to Dotter's early Nitinol experimental stent.

1.4. Vena Cava Filters

Vena Cava filters are placed in patients at risk of pulminary embolism (PE). They are designed to capture hazardous blood clots caused by trauma, surgery or other medical conditions before they can reach the lungs, causing a pulmonary embolism. The first Nitinol vena cava filter, quite possibly the first Nitinol vascular device, implanted in a patient was the Simon Nitinol filter. This filter used the thermal shape memory effect of Nitinol for expansion and had to becooled prior and during implantation [17]. The function of newer versions of the filter, however, is based on the superelastic expansion.

One of the market leaders for permanent filters is the TrapEase filter (Cordis), a "double basket" design that can be delivered using the jugular, femoral or antecubital approach [18]. It is an example of the simplicity of design afforded by the use of Nitinol. The filter consists of a monolythic structure laser cut from a Nitinol tube and expanded to its operating shape. It can be delivered through a 6 Fr catheter and is indicated for caval diameters up to 30 millimeters (Fig. **7**). A retrievable version of this filter is the Optease Filter (Cordis).

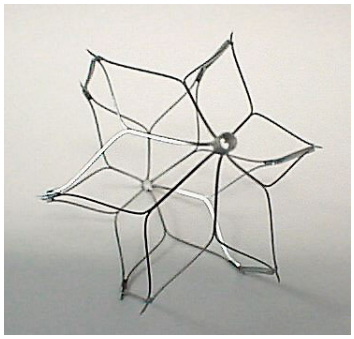


Fig. (7). TheTrapEase vena cava filter (Cordis) consists of a monolythic structure laser cut from a Nitinol tube and expanded to its operating shape.

Most recently approved as a permanent filter was the SafeFlo (Rafael Medical) [19], a Nitinol wire based double ring anchor and spiral filter element, which is delivered through a 6 Fr catheter, too.

1.5. Cardiac Septal Defect Repair

An interesting area of application, well-suited for the use of Nitinol, is the area of cardiac septal defect repair. The septum is the wall of tissue that divides the left and right sides of the heart, i.e., the right atrium from the left atrium and the right ventricle from the left ventricle. If a defect (e.g., a hole) exists in this septum, it allows blood to travel

directly from the right side of the heart to the left side of the heart, or vice versa, allowing mixing of venous and arterial blood. In some cases, this "shunt" may not be clinically significant; however, in many cases it can result in complications, including reduced oxygen content in the arterial blood, stroke, abnormal enlargement of part of the heart, pulmonary hypertension, and possibly even eventual heart failure.

The three primary types of cardiac septal defects are atrial septal defect (ASD), ventricular septal defect (VSD), and patent foramen ovale (PFO). ASDs and VSDs are congenital defects in the form of holes that exist in the atrial and ventricular septa, respectively. They are also quite rare, with a prevalence of less than 0.1% for both ASDs and VSDs. The *patent foramen ovale* has a different origin and a much higher prevalence. As the chambers of the heart form in a human fetus, a tunnel is created between the newly-formed right a left atria (the *foramen ovale*) that allows blood to pass directly from the venous circulation to the arterial circulation, circumventing the non-functioning fetal lungs. After birth, the pressure differential between the right and left atria changes with the introduction of blood flow to the now-functioning lungs. As a result the tunnel collapses and eventually closes completely within the first few months. In about 25% of people, however, the *foramen ovale* does not completely seal and remains "patent." In patients with a *patent foramen ovale* (PFO), the tunnel can reopen under circumstances of elevated right atrial pressure, such as coughing, straining or during sport exercises. One of the primary concerns with PFO is that it provides a pathway for blood clots (that normally form in the venous circulation) to pass directly to the arterial circulation without being filtered out by the capillary bed of the lungs. As a result, the presence of a PFO has been linked to a number of clinical issues, especially to cryptogenic stroke. Clinical trials are underway to prove the link between PFO to severe migraine [20]. The results of these trials will influence the ultimate size of PFO closure device market, which today is quite small at about $50 million worldwide. By some estimates, the potential market for PFO closure devices (if the clinical links to migraine are proven) is as high as $30 billion. There are currently no FDA-approved devices for PFO repair in the U.S., but a number of devices are available in Europe.

PFO closure is generally performed by a percutaneous, catheter-based procedure via a femoral vein access. PFO repair devices generally fall into one of three categories: "umbrella-type" devices (which often have a large umbrella-like structure on each side of the septum with a connecting post), suture-based technologies (which attempt to mimic surgical repair), and "in-tunnel" devices (which close the defect from within and minimize material in the atrial). Currently, only umbrella-type devices are commercially available (in Europe). Nitinol is especially useful for the umbrella-type and in-tunnel PFO repair devices, which require deployment of a large device (up to 35 mm diameter) through a small catheter (typically about 2 to 3 mm diameter). Some examples of Nitinol-based umbrella-type devices include the Amplatzer™ device from AGA Medical (which is made from a formed Nitinol braided wire structure), the STARflex™ from NMT Medical (which primarily uses MP35N for the framework, but has Nitinol centering springs), the Premere™ device from St. Jude Medical (made from Nitinol sheet), and the Helex™ device from W.L. Gore (made from a helical Nitinol wire with a PTFE cover) (Fig. **8**).

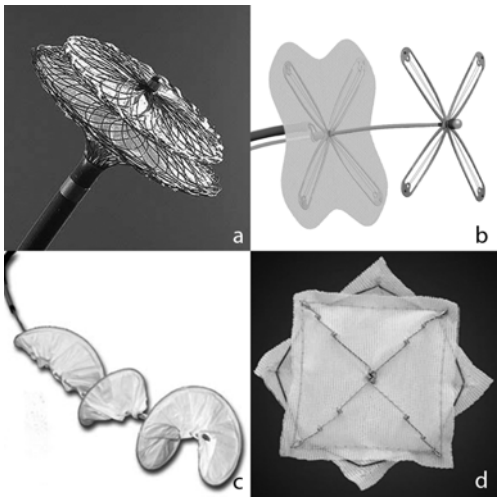


Fig. (8). Examples of "umbrella-type" PFO repair devices (a) the Amplatzer™ from AGA Medical, (b) the Premere™ device from St. Jude Medical, (c) the Helex™ from W.L. Gore, and (d) the STARflex™ from NMT Medical.

The original umbrella-type devices are adapted from closure devices originally developed for ASD and VSD repair. A new class of PFO repair devices has emerged recently that are tailored specifically for the unique anatomy of the PFO. These are the so-called "in-tunnel" devices. These devices recognize that the PFO is a long tunnel that is within and parallel to the septal wall rather than a simple perpendicular hole (like an ASD). In-tunnel devices are designed to conform to this unique PFO anatomy, while avoiding undue distortion of the native anatomy and minimizing the material left behind in the left and right atria. Two in-tunnel PFO repair devices are currently in development: the SeptRx™ device from SeptRx, Inc. and the FlatStent EF™ from Coherex Medical. The SeptRx device is made from a laser-cut Nitinol tube with a Nitinol wire braid, and the FlatStent EF is made from Nitinol sheet with a polymer mesh (Fig. **9**).

1.6. Self expanding Heart Valve Prosthesis

CoreValve has introduced the first self expanding heart valve prosthesis with biological valve leaflets, ReValving™ [21]. Percutaneous aortic valve replacement procedures are performed entirely in the cardiac catheterization laboratory avoiding open heart surgery under general anesthesia. The advancements of this beating heart procedure have eliminated the need for special equipment and incisions required to access and support the heart. The valve scaffolding element is designed similar to laser cut self expanding stents. The leaflets are placed in the stent structure and fixed with surgical threads (Fig. **10**). To avoid permanent deformation and damage of the biological valve the prosthesis is compressed with a special tool and inserted into the delivery system right at the start of the interventional procedure. The delivery system is either introduced into the femoral artery or the valve is delivered via a minor surgical trans apical route through the apex of the heart. The first clinical results are very promising. A variety of new designs for self ex-

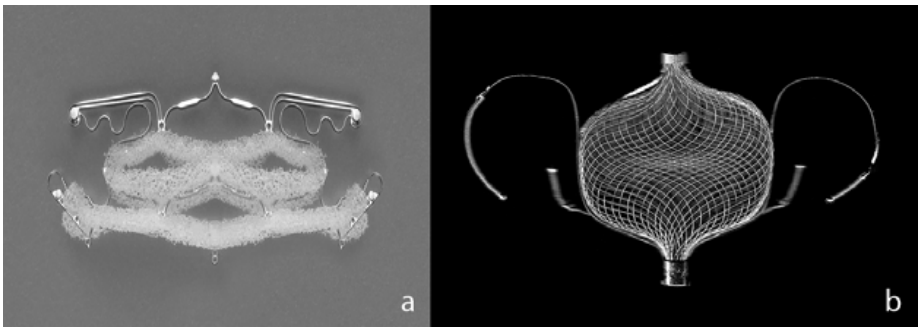


Fig. (9). Examples of "in-tunnel" PFO repair devices (**a**) the FlatStent EF™ from Coherex Medical and (**b**) the SeptRx™ device from SeptRx, Inc.

panding heart valve prosthesis are under development and are competing with balloon expandable systems (i.e Edwards Life sciences) [22].

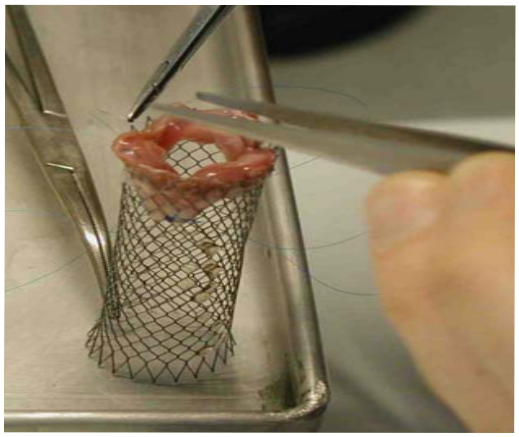


Fig. (10). The scaffolding element of self expanding heart valve prosthesis are designed similar to laser cut self expanding stents. The valve leaflets are fixed in the stent structure with surgical sutures.

2. PERFORMANCE OF ENDOVASCULAR NITINOL IMPLANTS IN MAGNETIC RESONANCE IMAGING, MRI

2.1. Introduction

The unconventional electromagnetic characteristics of Nitinol render these endovascular implants suitable for Magnetic Resonance Imaging. Together with Iron, Cobalt, Samarium etc. Nickel belongs to ferromagnetic metal family. However, in the state of the intermetallic compound, the Titanium and Nickel atoms are loosing their magnetic moment. Nitinol shows a similiar magnetic susceptibility as other medical grade titanium alloys [23]. Implants made of metal can cause significant MR image artifacts due to the material (susceptibility artifact) and/or electromagnetic characteristics (RF artifact). These artifacts are caused by the distortion of the magnetic field and interferences with the radio frequency (RF) waves of the MR imaging process. Complete signal loss occurs in close proximity or inside implants.

Recent years of research in MRI safety have revealed the problem of conductive heating of metallic implants which occurs at about 10 cm length in 3 Tesla MRI imaging and at about 20 cm length in 1.5 Tesla MR systems. Most implant are below the critical length but care has to be taken if stents are implanted with overlap and for the very long stents designed for AAA (Abdominal Aortic Aneurysm) and superficial femoral artery. Heating of more than 4^{o}C has been measured in phantoms models according to ASTM standards [24].

A more indirect risk is the problem of MR signals void in the lumen of the stents leading to diagnostic problems. The signal void is based of the closed cell design of a stents that functions along the line of a Faraday cage. MR signal can not be generated in and received from the substrate in the lumen, thus making MRI diagnosis of instent thrombosis and restenosis impossible. By integrating a MRI antenna (electric resonator tuned to the resonance frequency of the MRI i.e. 64 MHz for 1.5 Tesla) into the stent and other vascular implants i.e. heart valve prosthesis improved imaging is possible, which has been proven in various animal trials.

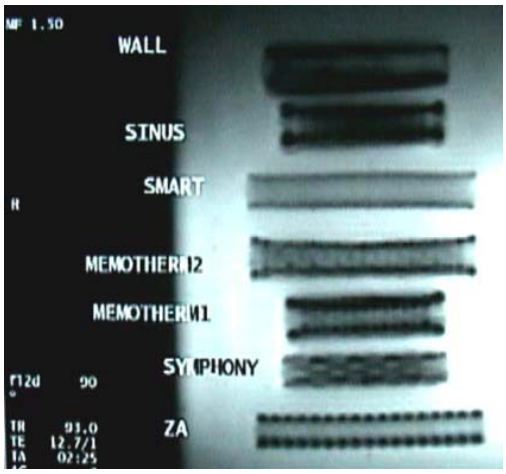
2.2. Nitinol Stents in MRI

A variety of reports on MR imaging behavior of stents have been published in recent years. MRI in the presence of stents have been examined regarding contrast enhancement aspects [25, 26] influence of different materials and different structures [27] and stent grafts [28, 29] under particular MRI settings. Effects of the 0.2 Tesla, 1.0 Tesla and 1.5 Tesla type of the sequence and imaging parameters, the orientation of the stent axis in the static magnetic field, and the orientation of the recorded slice were assessed systematically [23, 30].

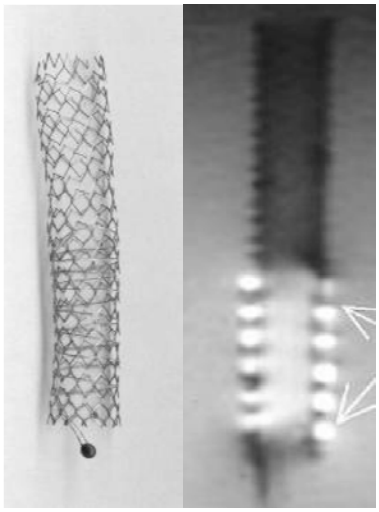
All self expanding stents show artificial increase of the wall thickness in MRI and shielding of the stent lumen (Fig. **11a**). To overcome the RF artifacts the stents have been enabled to function as electric resonators to interact with the MR imaging process (Fig. **11b**) [31, 32].

2.3. Stent Based Heart Valve Prosthesis in MRI

The use of MRI with 3D-orientation and good image quality (superior soft tissue contrast, arbitrary slice orientation, flow visualization) enables the planning and the implementation of the interventions without the side effects of the examination with X-Rays. The clear and artifact free visualization of the heart valve is one condition for a safe and reli-



a



b

Fig. (11). (**a**) MRI imaging of self expanding stents. (**b**) To overcome the RF artifacts, stents have been enabled to function as electric resonator (white arrows) to interact with the MR imaging process [30, 31].

able examination with MRI and MRI guided interventional use.

The artifacts can be minimized by using Nitinol with low magnetic susceptibility and a design of the scuff holding element which avoids electrical conductive loops. Due to the mechanical requirement the designs can not be made loop-free and non conductive. A resonant circuit tuned to the Larmor frequency of the MR tomography can overcome the RF artifacts and thus improve the visualization of the resonant prosthetic heart valve similar to the stents described above. We have developed a first set of prototypes using Bard Memotherm stent and freshly excised porcine aortic valve [33] (Fig. **12**). The valves have been implanted successfully under MRI guidance (1.5 Tesla) in an acute porcine animal model [34].

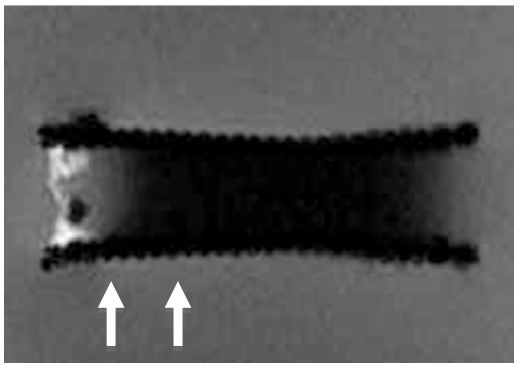


Fig. (12). MR visible self expanding heart valve prototype using a Bard Memotherm 22x 60 mm stent and a freshly excised porcine aortic valve [33] (see Fig. **10**). The section between the white arrows contains the resonant circuit and shows the valve leave lets whereas the non-resonant section is shielding the lumen.

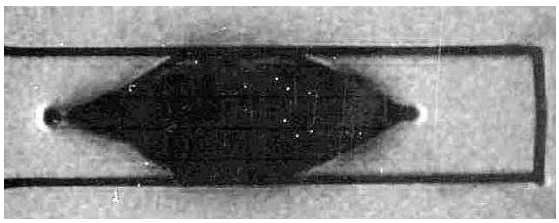
2.3. Vena Cava Filter in MRI

Wire based VCF show MRI artifacts that are depending on both the susceptibility and the RF characteristics of the pattern. Closed loops of wire are subjected to eddy current induction and local magnetization. The VCF TrapEase™ and OptEase™ are cut from Nitinol tube in a structure of interconnected closed cells that show a significant RF shielding [35, 36]. Two types of active MRI VCFs with integrated resonant circuits have been produced: wire based and laser cut from Nitinol tube (Fig. **13**). The active Vena Cava filter have been successfully tested in an acute porcine animal model for evaluation of MRI guided implantation, blood clot imaging and retrieval of the filter [37].

2.4. Closure Devices for Cardiac Septal Defects in MRI

MR Imaging of heart defects is a non invasive option specifically for the pediatric patients but also the follow up of the adult patient that has received a PFO closure profits from MRI [38]. MRI guided delivery of closure devices have been evaluated in animal [39].

As an example of a Nitinol closure device we have examined the Amplatzer™ from AGA medical [23]. The Nitinol mesh shows low artifacts, the clamping tubes that keep both end of the device lead to a larger artifact (Fig. **14 a,b**). Novel



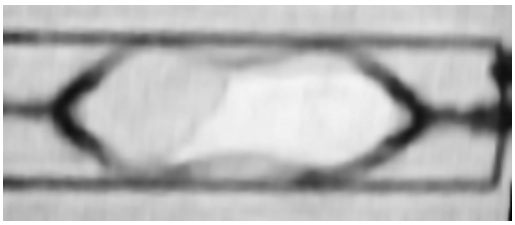


Fig. (13). (**a**) Basket type Nitinol vena cava filter (e.g. TrapEase) demonstrate an almost complete MR signal void because of the RF shielding of the device. (**b**) Integration of resonant circuit provides improved imaging and depiction of clotted blood in the filter (Thrombus is equivalent to the bright section).

intra tunnel devices for PFO closure such as SeptRx and feature smaller artifact mainly because less material is present and no stainless steel components are applied as structural components. PFO compression via two rings connected through the PFO tunnel may provide safe sealing and the integration of resonant circuits is equally helpful to overcome shielding artifacts and facilitates cardiac MR imaging.

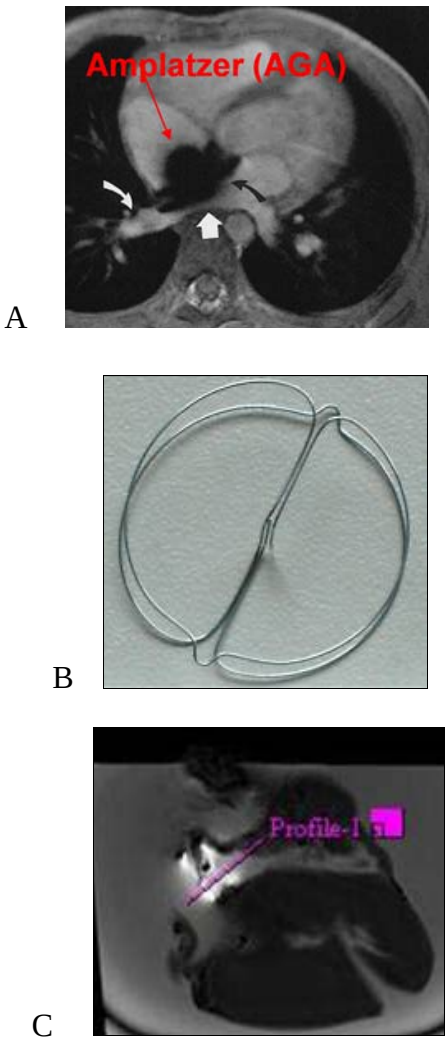


Fig. (14). (**A**) The Nitinol made Amplatzer™ occlusion devices shows low artifacts in MRI mainly due to the two crimped steel tubes that hold the ends of the mesh (source: www.radiographics.rsnajnls.org). (**B**) Resonant circuits of a novel Nitinol dual ring occluder is helpful to overcome shielding artifacts and facilitates cardiac MR imaging (**C**).

3. SUMMARY AND CONCLUSIONS

The paper describes the utilization and specific design criteria of super elastic Nitinol in cardiovascular implants. The feature of superelasticity and shape memory exhibited by advanced Nitinol materials adds multifunctionality to the design of various devices used to treat cardiovascular disease. Self-expansion, dynamic interference and stress hysteresis make Nitinol stents the preferred treatment for peripheral vascular disease with several stent types and brands now approved for indications, like iliac, SFA, carotid a.o.. Large and complex structures like vena cava filters and cardiac septal occlusion devices can be delivered through small catheters or using low profile delivery systems. Corrosion resistance, bio- and biomechanical compatibility, as well as MR compatibility and safety make Nitinol well suited for permanent implants in the vasculature.

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