

Review article

## Early research of shape memory polymer vascular stents

Lama Hewage Janitha Jeewantha<sup>1,2</sup>, Jayantha Ananda Epaarachchi<sup>1,2\*</sup>, Elizabeth Forster<sup>3</sup>,  
Mainul Islam<sup>1,2</sup>, Jinsong Leng<sup>1,2,4</sup>

<sup>1</sup>Centre for Future Materials, Institute for Advanced Engineering and Space Sciences, University of Southern Queensland, Toowoomba, Australia.

<sup>2</sup>School of Engineering, Faculty of Health Engineering and Sciences, University of Southern Queensland, Toowoomba, Australia.

<sup>3</sup>Health Sciences (N48) Nathan Campus, Griffith University, 4111 Queensland, Australia.

<sup>4</sup>Centre of Composite Materials and Structures, Harbin Institute of Technology, Harbin, China

Received 24 January 2022; accepted in revised form 11 April 2022

**Abstract.** Shape memory polymers (SMPs) are smart materials that can alter their shape from a temporary shape to a permanent shape upon external stimuli. This unique property of SMP has shown enormous potential and breakthrough discoveries in biomedical engineering. A significant number of SMP based research publications in the recent past have demonstrated the researchers' thirst to introduce SMP based devices to the biomedical fields. SMPs favour minimally invasive surgeries and showed enormous potential in vascular stents. However, SMP-based vascular stents have not been realised to date, indicating a lack of high-quality research outputs. Most studies either repeat or provide only minor extensions to ongoing research.

This article reviews SMP materials used for vascular stent development and briefly introduces SMP, architecture, and history. Followed by SMP vascular stent fabrication methods, responsive physical behaviour, the performance of the vascular stents, limitations, and *vitro* and *vivo* clinical results. Finally, challenges and future directions of SMP vascular stents are identified and highlighted.

**Keywords:** smart polymer, shape memory polymers, vascular stents, *in vitro*, *in vivo*

### 1. Background

This review article is primarily focused on the research and development of shape memory polymer (SMP) vascular stent from its initial invention to date. Several databases were included in the search strategy, including Scopus and Web of Science, to ensure a comprehensive literature review. According to the literature, most SMP researchers are focused on stents and a higher number of publications were seen between 2000–2010. However, the most published journal papers predominantly exchange views about synthesis, specific advantages, and the future impact on society. Moreover, consistent continuity of SMP vascular stents was not observed. Therefore,

this comprehensive review focused on SMP vascular stents, and this will be highly beneficial to those who emerge and want to use the potential of SMP in vascular stenting.

### 2. SMP analysis

The researchers were frequently used shape fixity ratio ( $R_f$ ) and shape recovery ratio ( $R_r$ ) to quantify and compare shape memory properties of SMPs and shape memory polymer composites (SMPCs) [1] (Equations (1) and (2)):

$$R_f = \frac{\theta_{\text{fixed}}(N)}{\theta_{\text{max}}} \cdot 100\% \quad (1)$$

\*Corresponding author, e-mail: [jayantha.epaarachchi@usq.edu.au](mailto:jayantha.epaarachchi@usq.edu.au)

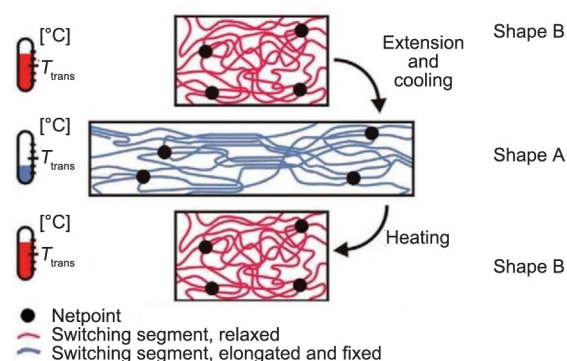
© BME-PT

$$R_r = \frac{\theta_{\max} - \theta_i(N)}{\theta_{\max} - \theta_i(N-1)} \cdot 100\% \quad (2)$$

For the  $N^{\text{th}}$  shape memory cycle ( $N > 1$ ) with respect to previous ( $N - 1$ ) cycle, where  $\theta_{\text{fixed}}$  is SMP/SMPC measured bending angle,  $\theta_{\max}$  is SMP/SMPC programmed bending angle,  $\theta_i$  is SMP/SMPC recovered bending angle.

### 3. General principle and architecture of SMP

SMPs are polymeric materials that keep a memory of their permanent shape and turn its physical shape from a temporary shape to a permanent shape with the application of external stimuli [2]. Most commonly, SMP is triggered by heat, electricity, magnetic field, moisture, chemical stimulus, microwave or light but frequently seen thermal activation, either directly or indirectly [3–15]. Among them, thermal and solvent stimuli are prominent in biomedical applications [16–18]. Energy and molecular structure theories are utilised to explain the shape memory mechanism of SMPs [19]. The permanent shape has low energy; thus, it is more stable. Above activation temperature, SMP molecules are transformed into an unstable high energy state and locked upon cooling [16]. However, the molecular chains are preferred to be in low-energy stable, permanent shape. Therefore, the polymer network releases energy and returns a stable, permanent shape beyond the glass transition temperature ( $T_g$ ) or melting temperature ( $T_m$ ). From a molecular point of view, the SMP network is incorporated with net-points and switchers (Figure 1) [20]. The net-points associated with chain segments that govern the permanent shape and switchers are



**Figure 1.** Molecular mechanism of the thermally induced shape-memory effect  $T_{\text{trans}}$  = thermal transition temperature related to the switching phase. Reproduced from [20] Copyright 2002. Wiley-VCH.

more sensitive to external triggering, which allows the material to deform temporarily [21]. The temporary shape fixation is obtained by solidifying the switching domain, which can be achieved through chemical or physical changes.

SMPs usually exhibit one-way shape memory effect (SME) and their composites might have two-way or multiple-way SME [22]. When exposed to external stimuli, one-way SMPs can switch from temporary shape to permanent shape and the process is irreversible. Besides that, researchers were synthesised one-way SMPs with more than one temporally shape [23, 24]. At triple SME, the material can memorise two temporary shapes and exhibit subsequent recovery at stimulus [25]. SME is a thermo-mechanical function of material and it requires careful coordination of chemical synthesis and macroscale processing [26]. After SMP has reached the transition stage, from the glassy stage to the rubbery phase, SMPs failed to keep the same mechanical properties. In order to better utilise SMPs' potential, this dilemma ought to be addressed. Plenty of literature is available on two way and multi-way SMPs, although many applications have been developed with bidirectional reversibility [7, 10, 27].

SMPs reinforced with fibres to improve mechanical and structural integrity [28, 29]. Carbon and glass are the most commonly used reinforcement fabrics in SMPs. Besides that, fillers and additives have been used to enhance the modulus, strength, toughness and flexibility of the SMP [30]. The addition of photo-thermal fillers and ferromagnetic particles ( $\text{Fe}_3\text{O}_4$ ) can trigger the activation mechanism of SMPs [5, 10, 31, 32].

Only a few SMP researchers have studied the reproducibility of their fabricated devices [33]. The variations of specific properties with time, particularly activation temperature and geometrical changes, could raise questions about the performance of the devices in the long term [34]. Another important factor is actuation time, which could depend on the application and the material parameters [35]. To assure the product functionality, scientists defined several thermo-mechanical parameters such as  $R_f$  Equation (1),  $R_r$  Equation (2) and many more [36]. However, research into long-term behaviours, *i.e.* fatigue and creep of SMPs was limited in the existing literature.

The rapid development in chemical engineering synthesis technologies improved the biodegradability and biocompatibility of SMPs. The novel SMPs

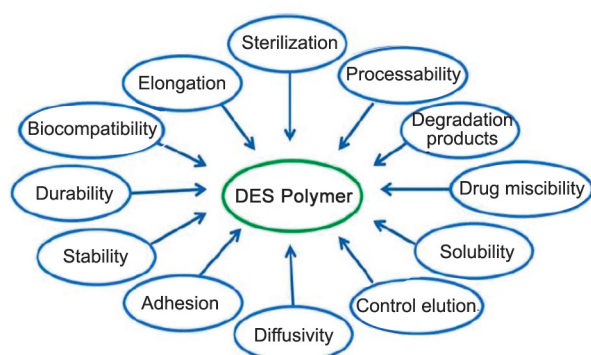
have desirable features, such as large recovery ability, lightweight, superior processability and low cost, which have generated enthusiasm among scientists [37–39]. Researchers have made SMPs with activation  $T_g$  near body temperature by synthesising chemicals in different ratios. Additionally, many investigators have proposed SMP medical devices that can be self-activated at body temperature [39–44]. Those components can be used in minimally invasive surgeries (MIS), which can be deployed into the body with a temporary shape. Upon the body thermal energy, the permanent shape can be recovered [45–47]. However, the biocompatibility and biodegradability of the SMP should be proved before use in the human body. The device functionality within the living organism, material stability, cytotoxicity, physiological and cellular function was studied during biocompatibility assessments [36, 48]. *Vitro* and *vivo* are the most popular and reliable two test procedures used to assure biocompatibility [46]. In *vitro*, biological properties are tested in a controlled environment (test tubes) and primarily determine general toxicity. In *vivo* testing, the materials are exposed to repetitive mechanical loads and biochemical attacks, including moisture, proteins, and enzymes. The test considers the whole living organism; hence, it is a particularly expensive and extremely time-consuming experimental approach. The key terminology and the most recent terms related to biomaterials and biocompatibility were published in the 2018 Chengdu conferences [49]. Biodegradable and biocompatible SMP fabricated implants reported less inflammation and good healing response; hence, they remarkably reduced the severity of the patient's life-threatening risks, such as implant rejection [50]. The ISO 9001 and ISO 13485 biomedical product realisation guidelines were followed during the SMP product design methodology introduced by the International Organization for Standards (ISO) [51]. The American Society for Testing and Materials (ASTM) has also introduced the F1635-11 protocol to assess biological degradation, especially in polymeric surgical implants in medical parts [52]. The potential SMP applications can be seen in general surgical, intravascular and vascular, urogenital, ophthalmic orthopaedic and orthodontic biomedical fields [1, 28, 30, 51, 53–56]. However, the Food and Drug Administration (FDA) or European standard (EN 556-1) approved devices are hardly found in the biomedical field [55].

Since the 1990s, shape memory alloy (SMA) based biomedical devices have been commercially available. However, after 2003 widespread enthusiasm regarding SMP occurred due to breakthrough product innovations such as surgical splints, soft-tissue fasteners and suture-anchor systems [46]. MedShape, USA is one of the leading companies which produce these products [51]. Many scientists committed their immeasurable time to accomplish these achievements. At present, around the globe, research groups and medical or healthcare industries are collaborating to develop modern biomedical devices with maturing SMPs technologies. The first successful polyurethane-based biodegradable and biocompatible material was developed by Mitsubishi Heavy Industry (MHI) Japan [57]. Following this, more research groups developed fundamental devices and models from SMPs closely related to the biomedical field. Professor Andreas Lendlein is one of the SMP pioneers and developed different applications, including self-tightening sutures and ureteral stents. However, a limited number of SMP vascular stents have been experimented with, and none of the devices has passed the extensive certification process.

#### 4. History and importance of SMP vascular stents

Heart disease remains a leading cause of death and morbidity among people worldwide. Myocardial infarction is due to insufficient blood supply or permanent blockage of coronary arteries in the heart, mainly due to atherosclerosis. Biomedical engineers found a solution to overcome this life-threatening issue by discovering an object called a stent. A stent is a tiny tube or circular lattice widely used to prevent narrowing of the arteries and maintain the required blood flow to the myocardium [36]. The stents were most commonly deployed through balloon-expansion or self-expansion [58]. At balloon expansion, the stent was plastically deformed due to inflation of the balloon, on the other hand, during self-expansion, the stent is deployed due to removed constrain forces or stimulus [59]. The first successful vascular stent was invented by Sigwart and his research team in 1986, which was made of stainless-steel mesh and later, it became a standard protocol and was approved by the FDA in 1994 [53, 60]. Commonly stainless steel 316L, Tantalum, Nitinol (Ni-Ti), Cobalt, Platinum and Iridium are used to manufacture

vascular stents [35, 61, 62]. Despite the fact that metal stents have been widely used in clinical practise, there are many drawbacks associated with metallic vascular stents, such as strong immune rejection, due to high stiffness damage to vessel walls, high cost of fabrication and clinical issues such as allergies with Nickel and some impurities in the metal [63–68]. Several studies have been published to address the existing issues due to rigidity mismatch in bare-metal stents such as blood vessel damage, inflammation, delayed endothelialisation, restenosis and thrombosis [35, 69–72]. As an immediate solution, Carbon, Titanium–nitride–oxide, polymer-coated drug-eluting vascular stents were tested, and results were not intensely satisfying the requirements [39, 73]. Due to this reason, researchers have been continuously working to find a permanent solution to replace traditional vascular stents. As a result, Duke University of North Carolina launched a special cardiac catheterisation program in 1983 [74]. Most importantly, biodegradability avoids the second surgery to remove implanted components and avoid undesired pain and discomfort [37]. Moreover, the urgent development of biodegradable vascular stents is more critical because permanent stents can cause more complications in future, such as neointimal hyperplasia and restenosis [73, 75]. The important properties of drug-eluting polymers are shown in Figure 2. Furthermore,



**Figure 2.** Polymer properties affecting drug-eluting stent (DES) performance. Reproduced from [81] Copyright 2010. Current pharmaceutical design.

developed stents were given the most attention because Venkatraman *et al.* [42] successfully experimented self-expandable bi-layered biodegradable vascular stent using elastic memory polymers [51]. Following this discovery, the cardiovascular SMP stents have developed rapidly, and surgical procedures were changed dramatically, with percutaneous rather than open procedures. During this period, stainless steel and polymer-coated metallic stents have been replaced with biodegradable polymeric stents [73, 76, 77]. The stents were fabricated with top-down and bottom-up fabrication strategies including etching, micro-electro discharge machining, electroforming, solvent casting, die-casting and nowadays 3D-printing, laser cutting and braiding [78–80]. However, on most occasions, SMP stents were fabricated while wrapping strips around circular mandrels [42, 44, 69]. Moreover, most SMP stents were fabricated with thermoplastic polymers, and they were mostly self-deployed due to stimulus and the performances were mainly governed by  $T_g$ , cross-links, triggering mechanism, geometry and programming conditions.

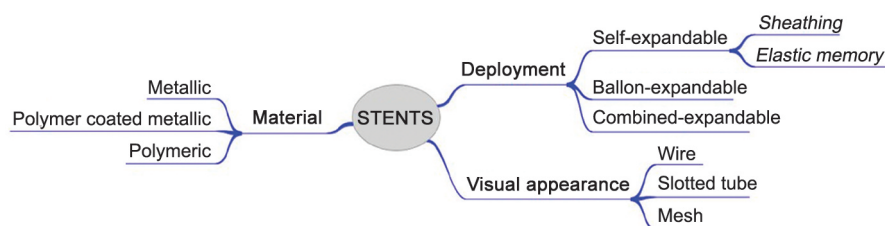
The overall summary of vascular stents can be classified by material, visual appearance and deployment methods which is shown in Figure 3.

In this structured review, the authors identified four SMP families that have thoroughly researched vascular stents as illustrated in Table 1 with their activation method. Section 5, detailed SMP synthesis process, vascular stent fabrication method, activation method, thermo-mechanical properties, *vitro* and *vivo* results and finally, progression to date. The majority of SMP vascular stents will expand on SME. However, expansion may influence by the elastic deformation if the material has a higher elastic modulus.

## 5. SMP vascular stents

### 5.1. PLA, PLLA and PGLA SMP vascular stents

Tamai *et al.* [43] made poly-L-lactic acid (PLLA) self-expandable helical springy coronary stents, which

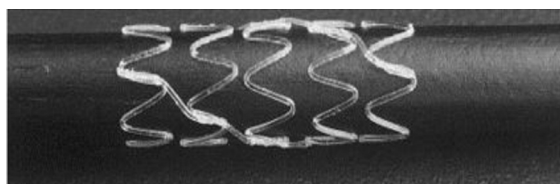


**Figure 3.** SMP vascular stent classification.

**Table 1.** SMP vascular stent materials and activation methods.

Material classification	Author	Material	Activation method*
PLA, PLLA and PGLA based SMP stents	Tamai <i>et al.</i> [43]	Poly-L-lactic acid (PLLA)	Thermal
	Venkatraman <i>et al.</i> [42]	PLLA, poly glycolic acid (PGLA) and polyethylene glycol (PEG)	
	Sonawane <i>et al.</i> [69]	PLGA and poly-lactic acid (PLA)	
Chitosan based SMP stents	Lauto <i>et al.</i> [82]	Deacetylated chitosan	Thermal
	Chen <i>et al.</i> [83]	Chitosan epoxy blend	
	Chen <i>et al.</i> [84]	Chitosan-based epoxy, Sirolimus, and heparin	
Polyurethane based SMP stents	Wache <i>et al.</i> [61]	Polyurethane (PU)	Thermal
	Baer <i>et al.</i> [85]	TPU, hexamethylene diisocyanate (HDI), <i>N,N,N'</i> -tetrakis(2-hydroxypropyl)ethylenediamine (HPED), triethanolamine (TEA) and Epolight 4121	Light
	Duarah <i>et al.</i> [30]	Hyperbranched polyurethane (HPU), carbon dot (CD) and Ag nanohybrids	Thermal
	Zou <i>et al.</i> [86]	Polyurethane and Fe <sub>3</sub> O <sub>4</sub> nanoparticles	Magnetic
	Gu and coworker [31, 87]	polylactide-base polyurethane (PLAU), Fe <sub>3</sub> O <sub>4</sub> nanoparticles and oleic acid	Thermal/Magnetic
	Zeng <i>et al.</i> [44]	Poly(propylene carbonate) (PPC) with TPU	Thermal
PCL based SMP stents	Yu <i>et al.</i> [35]	Poly( $\epsilon$ -caprolactone- <i>co</i> -DL-lactide)/ PCLA	Thermal
	Xue <i>et al.</i> [47]	Poly( $\epsilon$ -caprolactone) (PCL), methylene diphenyl diisocyanate (MDI) and crystallisable microbial polyester, poly[(R)-hydroxybutyrate- <i>co</i> -(R)-3-hydroxyvalerate] (PHBV)	
	Yang <i>et al.</i> [88]	PEG and PCL	
	Boire <i>et al.</i> [89]	PCL and poly( $\alpha$ -allyl carboxylate- $\epsilon$ -caprolactone) (ACPCL)	
	Zheng <i>et al.</i> [44]	Poly(propylene carbonate) (PPC) and PCL	
	Ajili <i>et al.</i> [90]	PCL and PU	
	Ansari <i>et al.</i> [91]	PCL and PU	
	Chasse <i>et al.</i> [52]	PCL	
	Park <i>et al.</i> [92]	PCL and poly(glycidyl methacrylate) (PGMA)	

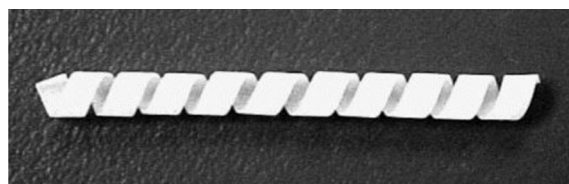
had 0.17 mm strut thickness. The study aimed to investigate the feasibility, safety and efficiency of this PLLA biodegradable stent. The stent was deployed with a hot liquid balloon and the *vivo* preliminary study showed that the stent took 0.2 seconds at 70 °C temperature and 20 min at 37 °C. This higher temperature can affect and damage surrounding tissues. The research group installed 25 stents in selected 15 patients and closely observed them clinically for approximately around six months and no deaths were reported. According to the literature, this was the



**Figure 4.** Image of the Igaki-Tamai self-expandable stent made of PLLA. Reproduced from [43] Copyright 2000. American heart association.

first successful polymeric stent installed in a human body (Figure 4). It was believed that this breakthrough could be a better replacement for metallic stents due to its biodegradability and delivery of locally administrated drugs.

To overcome higher activation temperature of the Igaki-Tamai stent, Venkatraman *et al.* [42] developed a bi-layered, self-responsive near body temperature vascular stent with well-accepted biodegradable PLLA and poly glycolic acid (PGLA), polyethylene glycol (PEG) plasticiser (Figure 5) were



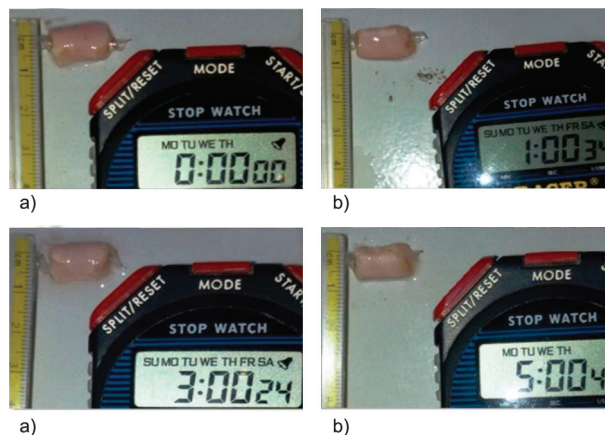
**Figure 5.** Image of the bi-layer self-activation stent prototype. Reproduced from [42] Copyright 2005. Biomaterials.

also developed. The overall thickness of the vascular stent was maintained at 0.15 mm. The different configurations of polymer strips were cast with an automatic film applicator and dried in a vacuum chamber around one week before it was cut into  $2 \times 70$  mm<sup>2</sup> strips. Then the strips were warped around different diameter mandrels.

Venkatraman *et al.* [42] stated that the self-expansion process is unique for helical design and it is proportional to the tensile modulus of the material. The outer layer should have low  $T_g$  and lesser thickness than the inner layer to increase the expansion rate. Furthermore, Venkatraman *et al.* [42] studied bi-layered vascular stents consisting of higher  $T_g$  PLLA and low mechanical strength PLGA 53/47. It was found that the minimum recoil (36%) percentage occurred at thickness ratios of PLLA/PLGA:0.08/0.07 within 8 min at 37°C water. More importantly, PLLA single layer stent took 15 days to fully recover at 37°C, suggesting an extremely higher temperature such as 70°C for efficient use of the Igaki-Tamai stent.

As a further improvement, Sonawane *et al.* [69] used the solvent casting method to fabricate SMP self-expandable stent with PLGA and poly-lactic acid (PLA) biomaterials. The polymers, PLA and PLGA were mixed and the synthesised SMP helically wrapped around the glass mandrel and the drug was loaded. Based on self-expandability, a 3:2 ratio of PLA to PLGA polymer composition with 0.25 mm thick stent was given an optimum. Differential scanning calorimetry (DSC) results verified no physio-chemical reaction between drug and stent polymer materials. Additionally, field emission scanning electron microscopy (FESEM) images have confirmed the presence of drugs on the vascular stent surface. Furthermore, platelet and bacterial adhesion tests were conducted to assure biocompatibility and biodegradability of the stent hemocompatibility.

A Goat vessel was used to conduct a *vivo* experiment, and a considerable increase of stent diameter was observed while decreasing stent length (Figure 6). During the demonstration, the goat vessel was kept at 37°C, which is closer to body temperature. Additionally, Sonawane *et al.* [69] evaluated the tensile behaviour of bare and drug-loaded stents and found that both stents can hold up to 500 g and did not show a significant variation of mechanical strength. The vascular stent was specially checked for bacterial

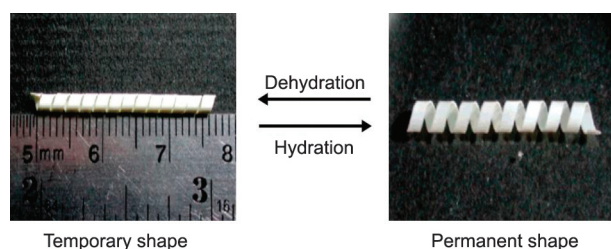


**Figure 6.** Images of (a) original diameter of polymeric stent after insertion into Goat vessel, Ex-*vivo* behaviour of stent (b) 1 min, (c) 3 min, (d) 5 min. Reproduced from [69] Copyright 2017. Artificial Cells Nanomedicine and Biotechnology.

adhesion and found that the antibacterial effect is essential to avoid biofilm formation.

## 5.2. Chitosan vascular stents

Researchers have conducted experiments on chitosan-based self-expandable stents over the last decades. Thermal, elasticity, biocompatible, antimicrobial and homeostatic properties have increased researches interest on chitosan stents [83, 93]. As a preliminary experiment, Lauto *et al.* [82] helically wound a  $0.055 \pm 0.005$  mm thick film around a cylindrical rod and made a deacetylated chitosan (4% w/v) vascular stent. The stent was tested with Wistar rats and no granuloma was observed after two weeks *in situ*. The stent self-expanded over 50% for its initial diameter within 3 min, and the expansion process was irreversible. The implantation should be quick to avoid expansion prior to appropriate positioning. Further, Chen *et al.* [83] developed an epoxy eluted blended chitosan novel SMP vascular stent. During the mechanical testing, the Chitosan stent endured up to 30% of deformation and, most importantly, managed to recover its initial shape simultaneously compared to a metallic stent that tolerated only 10% before becoming irreversible. Another key challenge was maintaining collapsed pressure in the stent. The measured collapsed pressure  $2.18 \pm 0.16$  bars, and it was significantly higher than the required level. The stent swells in an aqueous environment and recovered permanent shape within 150 seconds in 37°C, which was significantly fast shape recovery compared to findings of Tamai *et al.* [43] and Venkatraman *et*

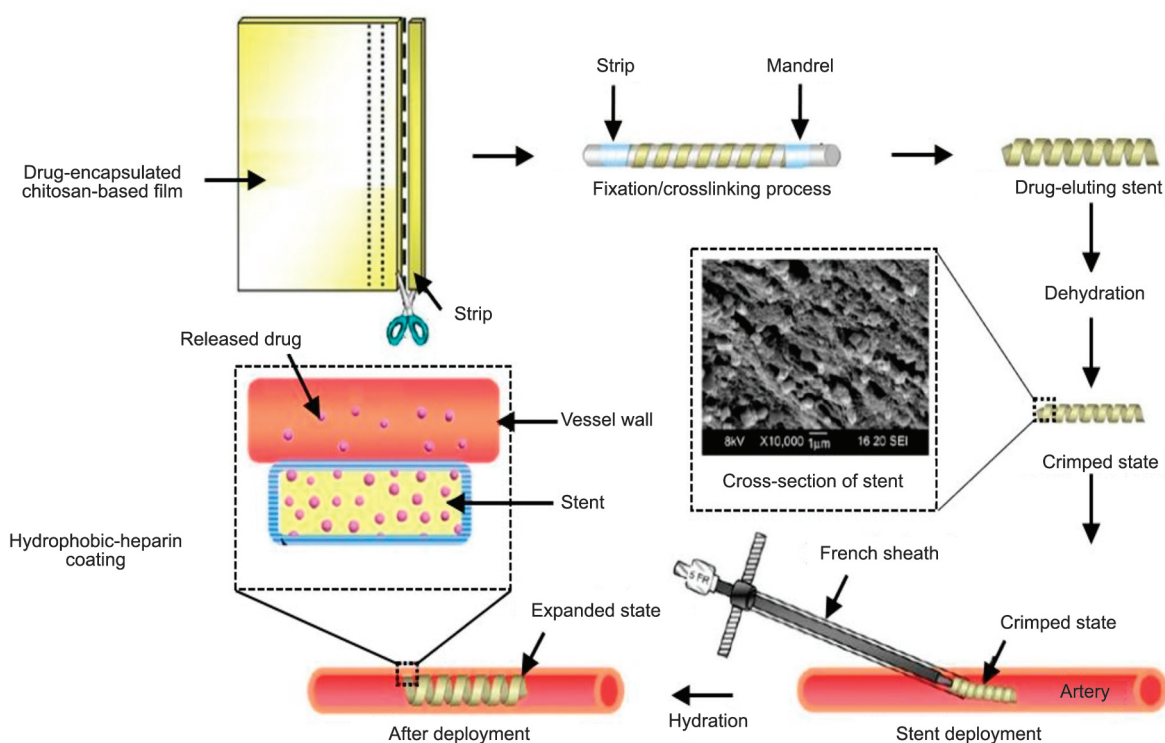


**Figure 7.** Images of the shape-switching process, which is reversible and controlled by hydration or dehydration of the cross-linked stent. Reproduced from [83] Copyright 2007. American Chemical Society.

*al.* [42] It has been seen that ideally, self-expansion time should not be more than one minutes to prevent migration after implantation [30, 42]. Moreover, Chen *et al.* [83] made their stent of a hydrophilic material; hence, the recovery process was enhanced in the aqueous environment while PLLA, PLGA based stents needed to be stimulated thermally (Figure 7). The vascular stent was fabricated to fit the rabbit artery and the stent to artery diameter ratio was maintained about 1.1~1.2 to prevent unnecessary migration after deployment. Before implantation, the vascular stent was coated with heparin to avoid thrombogenicity and sterilised with ethylene oxide. The developers successfully implanted the vascular stent

in a rabbit's abdominal aorta and did not observe any thrombus formation during the clinical examinations. However, frequent exposure of blood cells to foreign body parts can cause thrombus formation, restenosis, platelet adhesion and blood coagulation [94]. Hence the neutral behaviour of implanted body parts is essential for effective functioning. Therefore chiton-based materials are better and suited for implanted artificial devices. Additionally, *vitro* results have proved that heparin coating reduced the risk of platelet adhesion and blood coagulation.

As a continuation of previous research, in 2008, Chen *et al.* [84] developed a chitosan-based epoxy cross-linked sirolimus-eluting hydrophobic heparin-coated biodegradable SMP vascular stent. The uniformly  $0.10 \pm 0.001$  mm thickened chitosan film was cut into  $100 \times 1.2$  mm<sup>2</sup> and wound on 3 mm diameter mandrel before immersed in an epoxy compound, as shown in Figure 8. Finally, it was rinsed with Phosphate Buffered Saline (PBS) solution to remove residual cross-linking and was allowed to dry. During the drying process, the stent was shrunk and returned to a temporary shape and this allowed it to be implanted through small incisions in narrow arteries and recover its permanent shape by hydration. Mechanical stability of the stent was verified, and Chen *et al.*



**Figure 8.** Schematic illustrations of the development process of the sirolimus-eluting polymeric stent and the deployment of the stent in an artery via a French sheath in the animal study. Reproduced from [84] Copyright 2008. Biomaterials.

[84] revealed that 85% of the cross-linked sample had achieved the best mechanical properties. According to *vivo* results, the drug release rate remained constant and can be controlled with the outer heparin layer. The reduction of drug release rate delayed arterial healing and may lead to late thrombosis development. Hence the optimum drug release rate needs to be investigated in future experiments. After one month of clinical examinations, results verified that none of the stents migrated in animals' infrarenal abdominal aorta. Hence it was concluded that proper expansion had taken place during the deployment. According to literature, sirolimus reduced migration, delaying healing and reduced neointimal formation as well [95]. Therefore, during SMP vascular stent fabrication, careful consideration of other affected parameters are more crucial to improve reliability. Further, Chen *et al.* [84] discussed the optimum time span for biodegradable vascular stents before totally deterioration with enzymatic reaction. It was found that the strength should remain more than six months in humans' bodies to achieve optimum medical requirements [96]. Chen *et al.* [84] identified cross-linking may affect the degradation of the stent and further investigation will be required before applying it to humans. The proposed stent has shown a greater potential and one of the best vascular stent designs found in the literature; however, after Chen and co-worker [83, 84], researchers' interest in chitosan stents has sharply declined.

### 5.3. Polyurethane vascular stents

Polyurethane materials are being used frequently for biocompatible SMP material development. A thermoplastic polyurethane (TPU) based SMP vascular stent was developed by Wache *et al.* [61] and it was a turning point in the history of SMP based stent development. Injection moulding and extrusion processes were used in the manufacturing process. In pre-trials, axial deformation of the polyurethane stent was more than 200%. During the activation, the diameter of the vascular stent was increased by approximately 100%. Hence, Wache *et al.* [61] mentioned the importance of finding optimum elongation and recoil parameters. The activation temperature of the stent was between (80–120) °C, which was a significantly high range for biomedical applications. The drugs were loaded into the vascular stent during the manufacturing process. The study revealed a 10% drug loaded stent's release rate with three different active agents

over 150 hours. Maximally the vascular stent can be loaded with drugs 35% by weight in a 30 mm long stent. There is more room for improvements in Wache *et al.*'s [61] vascular stent, such as lowering activation temperature, quick expansion, and improving its mechanical properties.

The next generation polyurethane vascular stent was fabricated Baer *et al.* [85], which can trigger using laser light. TPU pellets were obtained from MHI Japan and dissolved in organic tetrahydrofuran (THF) liquid. The stainless-steel pin was dip-coated multiple times in THF solution to obtain uniform thickness and vacuum dried at 50 °C for 24 hours to avoid air bubbles forming while curing. Before removing the SMP stent from the stainless-steel pin Baer *et al.* [85] have used laser light to etch the required pattern. After that SMP stent was doped in Epolight 4121 platinum dye to embed laser active monomers to deliver thermal energy uniformly. An optical fibre was covered with 5 mm coaxial SMP foam and inserted into the SMP stent. The foam was synthesised with hexamethylene diisocyanate (HDI), *N,N,N'*-tetrakis(2-hydroxypropyl)ethylenediamine (HPED), triethanolamine (TEA) and Epolight 4121 to make sure activation upon laser energy [97]. During thermo-mechanical analysis, DSC results showed that foam  $T_g$  was around 40 °C and can be tailored from 30 to 86 °C by varying the HDI, HPED and TEA chemical composition or changing the stent fabrication methodology [98, 99]. Nevertheless, foam  $T_g$  should not setup below body temperature to avoid self-activation during implantation. The foam was encompassed inside the vascular stent to center the laser diffuser, improve light scattering and to reduce the convective cooling. In zero flow conditions, laser power varied from 0 to 8.6 W with 1 W increments and the stent took 6.3 min to activate. However, the stent managed to expand about 60% during the physiological flow at 180 ml/min. The time taken for activation was too long compared to other SMP vascular stents, and continuous supply of laser power was required during the activation. Another drawback was laying fibre optic cable up to the surgical location. Therefore, the authors have suggested that to use this stent in larger vessels such as internal carotid. On the other hand, unlike other SMP vascular stents, this can expand upon external triggering. Rapidly increased laser power or a constant power over a period of time reduces the actuation time. A rapid expansion of the stent was observed at the tip of the fibre optic due to higher

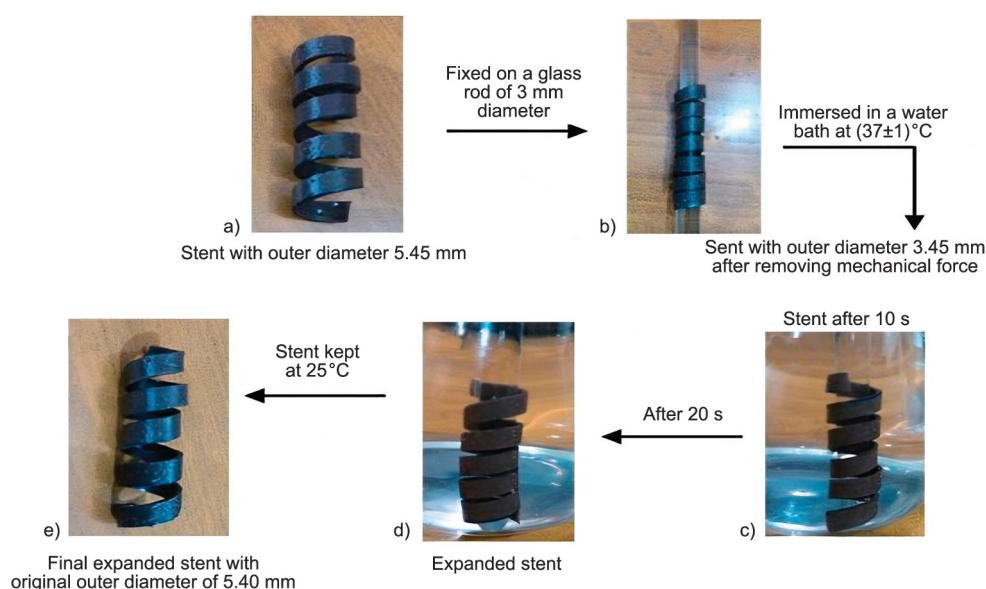
concentrated light density. The surrounding fluid temperature remained relatively constant during the actuation, which reduced severe damage to surrounding tissues. Despite that Baer *et al.* [85] proposed further optimisation and performance enhancement can be carried out by adjusting the laser power and dye concentration which is more favourable to blood. The vascular stent was not tested for biocompatibility even though the study claimed it was a breakthrough of the laser-triggered stent development process.

Hyperbranched polyurethane (HPU) was synthesised with biocompatible carbon dot-silver (CD-Ag) nanohybrid in different weight ratios by Duarah *et al.* [30] and fabricated a rapid self-expanding infection resistant vascular stent. The carbon dots rich with polarised carbon particles can make strong polymeric bonds within a matrix. Therefore, the resultant polymer has excellent mechanical properties [100]. Further, carbon dots are extensively researched in many biomedical applications due to their unique properties, such as solubility and optical properties [101]. In contrast, the carbon dot has been seldomly used in vascular stent applications even though it has higher degradation and fatigue failure time. During their experiments, Duarah *et al.* [30] have made three primary stents, particularly HPU/CD, HPU/AG and HPU/CD-Ag and thoroughly investigated them using experimental methods (Figure 9). A significant performance enhancement was observed in mechanical

and thermal stability with 5% nanohybrid materials in the HPU/CD-Ag stent due to strong H-bonding, covalent bonding and polar-polar interactions between nanohybrids and HPU chains.

Further, SMP behaviour was measured with time and has shown a higher percentage of  $R_f$  and faster  $R_r$  within a tiny  $(37 \pm 1)^\circ\text{C}$  temperature range due to crystallinity improvement in the nanohybrid matrix. It has been shown that the self-expandability was increased with CD nanohybrid. In addition, CD nanohybrid added HPU/CD-Ag vascular stent has expanded to maximum limits within 20 seconds due to the higher heat absorption capacity of the material. More specifically, the vascular stent has self-expanded ( $>99\%$ ) diameter from 3.45 to 5.40 mm within 20 seconds.

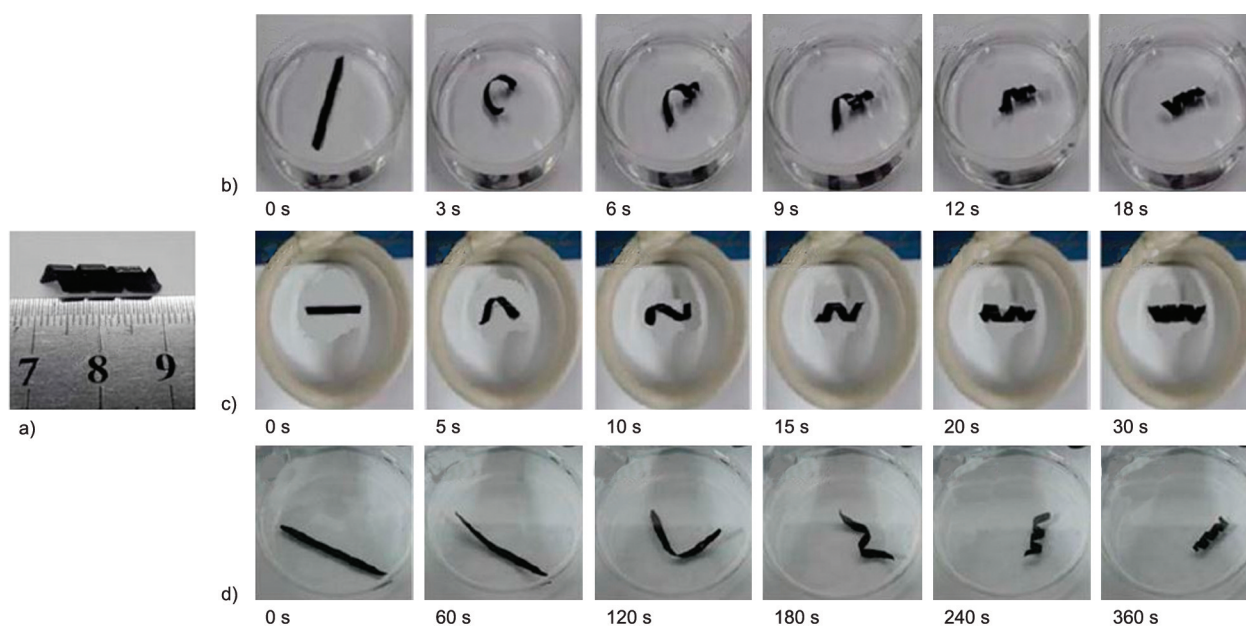
HPU/CD-Ag is used with an optical density method to investigate bacterial resistance and results have proved maximum antibacterial activity and reduction in the growth of *E.coli* and *S.aureus*. It may form biofilm inside the stent during antibacterial growth; However, a limited number of researchers reported biofilm formation. Gwon *et al.*'s [102] antibacterial cefotaxime-eluting vascular stent could not avoid biofilm formation. However, silver nanoparticles (AgNPs) successfully resist and address biofilm formation in vascular stents, but only a few papers were found on this. Because of the limited evidence, HPU/CD-Ag based vascular stents have higher demand among the SMP researchers.



**Figure 9.** Images of Self-expansion behaviour of HPU/CD-Ag5 stent. (a) Original shape, (b) shape after removal of the mechanical force; shapes after (c) 10 s and (d) 20 s at  $37 \pm 1^\circ\text{C}$ ; and (e) final expanded shape after 20 s at  $25^\circ\text{C}$ . Reproduced from [30] Copyright 2016. Biofabrication.

Duarah *et al.* [30] has developed a polyurethane vascular stent which can be activated by a thermal stimulus. However, polyurethane with  $\text{Fe}_3\text{O}_4$  nanoparticles (NPs) allows scientists to activate stents magnetically. Recently, Zou *et al.* [5] have developed an SMP material that can trigger electromagnetic radiation. This new material was prepared using the solution casting method and composed of commercially available PU and  $\text{Fe}_3\text{O}_4$  NPs and oleic acid (OA).  $\text{Fe}_3\text{O}_4$  was used to improve the dispersibility of the compound material. According to DSC results, the transition temperature was matched with  $T_m$ , 42 °C.  $R_f$  and  $R_r$  were not affected with added NPs until they reached 20% by volume fraction and after that, a slight variation was noticed. Further, 1% NPs contained film recovered its permanent shape within 30 seconds and 20% NPs took only 9 seconds. Recovery time and amount of required magnetic radiation decrease with increasing NPs. Yakacki and coworker [40, 103] have synthesised methyl methacrylate (MMA) and poly(ethylene glycol)<sub>n</sub> dimethacrylate (PEGDMA) reinforced with  $\text{Fe}_3\text{O}_4$ , which can activate remotely by magnetic radiation. However, the  $T_g$  was significantly higher for biomedical applications. In a separate study, Gu *et al.* [31] have enhanced magnetic responsiveness by synthesising polylactide-base polyurethane (PLAU) with  $\text{Fe}_3\text{O}_4$  reinforcement and additionally, OA was used to assure homogeneity of  $\text{Fe}_3\text{O}_4$  NPs. Gu *et al.* [31] made samples with different  $\text{Fe}_3\text{O}_4$  ratios 0, 3, 6 and 9% and found

those samples took 15, 19, 14 and 11 seconds to recover permanent shape, respectively, at 55 °C. Moreover, in the same study Gu *et al.* [31] recovered samples within 60, 35 and 24 seconds by altering the magnetic field alone. In addition to that, Gu *et al.* [87] obtained poly(lactide-co-caprolactone) based polyurethane (PCLAU) synthesising PCLA with polytetramethylene ether (PTMEG) and hexamethylene diisocyanate (HDI). The PCLAU/ $\text{Fe}_3\text{O}_4$  dual-induced self-expandable shape memory polyurethane nanocomposites was obtained by adding an appropriate amount of  $\text{Fe}_3\text{O}_4$  into the reactant mixture that operated at 40 °C just above body temperature (Figure 10) [87]. The  $T_g$  was adjusted using lactic acid (LA)/caprolactone (CL) molar ratio and the ratio of 9:1 to set  $T_g$  around body temperatures [23]. To activate two methods, nano- $\text{Fe}_3\text{O}_4$  was introduced to the PCLAU matrix; it has been found that the inclusion of nano- $\text{Fe}_3\text{O}_4$  has not influenced the  $T_g$  of the composite. The amorphous PCLA showed an onset of  $T_g$  around 33 °C. However, nanocomposite has shown slow recovery around 37 °C, and a faster dual-responsive recovery was observed around 40 °C body temperature. Moreover,  $\text{Fe}_3\text{O}_4$  embed nanocomposite showed almost 100%  $R_f$  and more than 82%  $R_r$  during thermomechanical cycles. In *vitro* degradation, cracks were initiated after 3 weeks, and 67% weight loss was recorded after 13 weeks of implantation. Zeng *et al.* [104] fabricated the most recent body temperature triggered SMP vascular stent by melt blending



**Figure 10.** Images of the (a) permanent shape of the stent, (b) shape recovery procedures of the stent in a 40 °C water bath, (c) in an alternating magnetic field, (d) in a 37 °C water bath. Reproduced from [87] Copyright 2017. Applied polymer science.

amorphous poly(propylene carbonate) (PPC) with elastic TPU. The tailored PPC has improved ductility, recovery speed,  $R_r$  but low  $R_f$ . The most optimal properties were obtained with 50% of each PPC and TPU and named PT50. The 1 mm, thick PT50 film was wrapped around a cylindrical glass rod, and a spiral-shaped vascular stent was fabricated. The *vitro* experiment was performed inserting 2.8 mm diameter vascular stent into a 7 mm diameter silicone tube. The silicone tube was immersed in a hot water bath. The temperature of hot water was maintained around 37 °C and within 20 seconds, the vascular stent obtained a silicone tube diameter. The quick response of PT50 has proved suitable as a stenting material. Further, Zeng *et al.* [44] conducted blood and cell compatibility of the designed stent and have shown a great potential of PT50 in stent applications.

Schreiber *et al.* [80] used novel braided technology to fabricate polyurethane SMP stent. The  $T_g$  of the SMP was between 70 to 150 °C and the radial resistance force of 0.08 N at a compression of 1 mm. The experimental compression results revealed heat-treated braids improved the radial stiffness and were competitive with other SMP stents. Additionally, Kim *et al.* [76] analysed polyurethane braided SMP stents' mechanical properties and showed gradual expansion respect to body temperature.

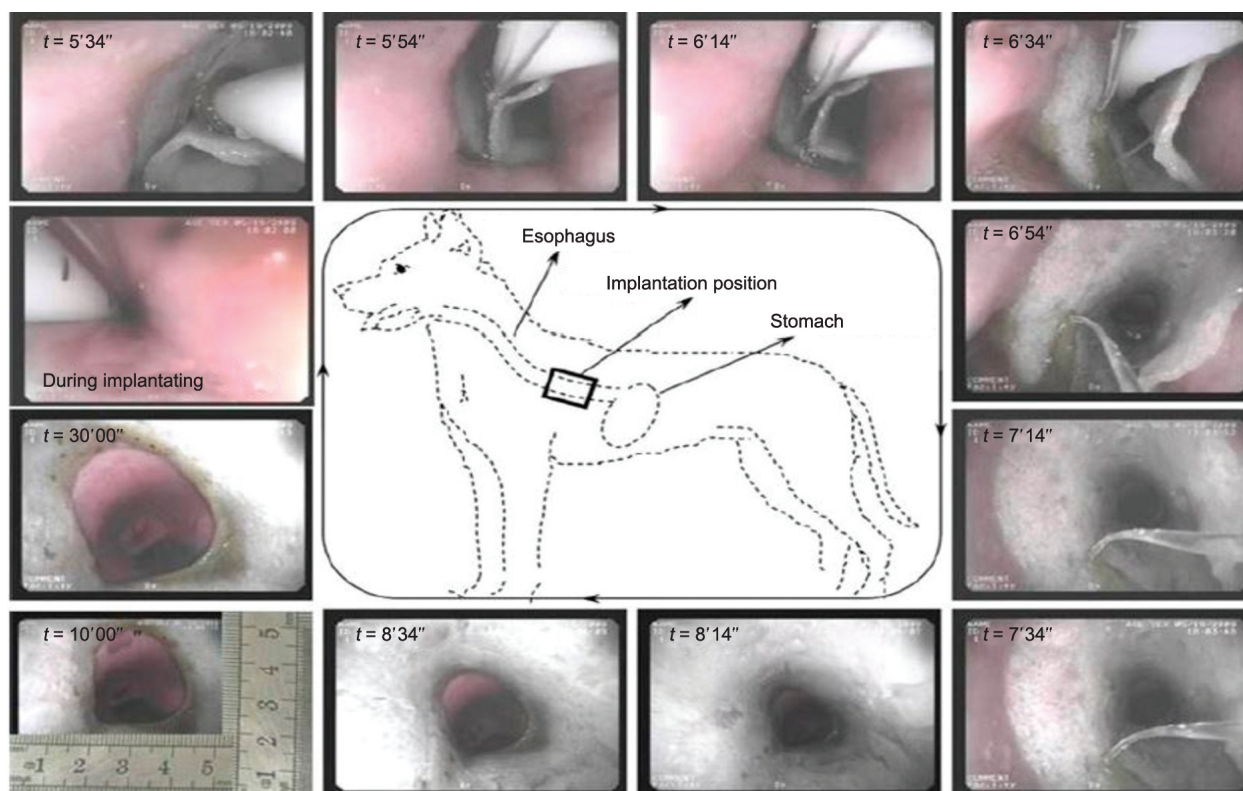
#### 5.4. PCL vascular stents

PCL is the most widely used SMP in biomedical vascular stent applications. However, the PCL  $T_m$  is around 60 °C and difficult for human tissues to withstand that temperature. The PCL's relatively low strength was increased by mixing different co-polymers and hard segments such as PLA, methylene diisocyanate (MDI), PCPCL, and PPC. Bellin *et al.* [24] reported that triple SME stent initial shape to second shape transition was fixed by physical cross-links and second shape to third shape transition derived by covalent cross-links. The initial two shapes were defined through thermomechanical programming and net-points defined the final shape of the stent. Three triggering temperatures are as follows 20, 40 and 60 °C. The polymer structure is divided into two networks. The first polymer network named MACL consists of poly( $\epsilon$ -caprolactone) (PCL) and poly(cyclohexyl methacrylate) (PCHMA). The second network, called (CLEG), consists of PCL and Polyethylene glycol (PEG). The triple SME, allowed to insert vascular stent in a compressed shape and

followed by expansion into the second position and third contracted shape that facilitates the removal of the stent later. Bellin *et al.* [24] analysed the thermomechanical properties; however, essential biomedical requirements were not discussed. Hager *et al.* [2] presented a list of triple and multiple SMPs but further studies on triple SME polymeric stents are limited in the existing literature.

Yu *et al.* [35] further studied the properties of PCL and found that the degradation was exceedingly over 14 weeks and therefore limited in clinical applications. Due to this reason, Yu *et al.* [35] have chosen biocompatibility and biodegradability verified popular poly-DL-lactide (PLA) homopolymer to improve properties of PCL. The proposed new polymeric material was synthesised with  $\epsilon$ -caprolactone ( $\epsilon$ -CL) and DL-lactide (DL-LA) with 10/90 weight ratios. The ring-opening copolymerisation technique was used to synthesise, and the new material was named poly( $\epsilon$ -caprolactone-*co*-DL-lactide)/PCLA. The PCLA material properties were compared with commercially available metallic (Ni-Ti alloy) stents, and nearly equal mechanical strength was observed. Further, Yu *et al.* [35] noted that compressive strength unexpectedly increased with the degradation. The vascular stent was triggered at 37 °C and the geometrical length was 13 mm and the inner and outer diameters were 28 and 30 mm, respectively. During *vivo* experiments, the vascular stent was implanted in a dog's oesophagus through a minimally invasive technique (Figure 11). Before the procedure, the dogs were kept inside (20–22) °C environmental temperature for around 12 hours to lower the body temperature. The surgical team managed to finish implantation within 5 min and 30 seconds. The frequently observed endoscopic results were verified that the stent prevented restenosis before slipping into the stomach. The smoother stent surface and bigger oesophagus were identified as the main reason for the slippage.

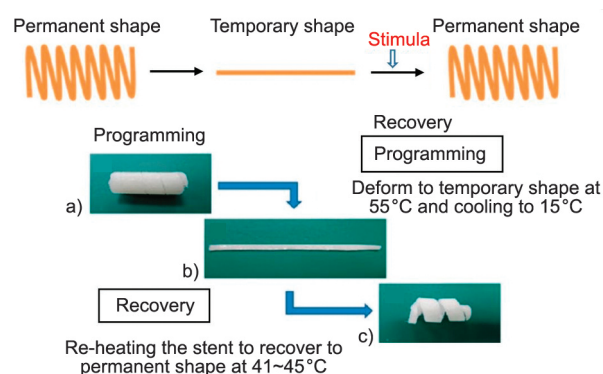
Xue *et al.* [47] has proposed a faster self-expandable vascular stent containing three arms of poly( $\epsilon$ -caprolactone) (PCL) as switching segments together with MDI and crystallisable microbial polyester, poly[(R)-hydroxybutyrate-*co*-(R)-3-hydroxyvalerate] (PHBV) as hard segments]. The PHBV offered adjustable mechanical and thermal properties in the hard segments, while three PCL arms improved chain flexibility to stand switching temperature around (39–40) °C [105]. The stent actuated with  $T_m$  was rapid



**Figure 11.** The schematic illustration and the endoscopic photos showing the process of shape memory recovery of PCLA stent implanted in the dog's esophagus. Reproduced from [35] Copyright 2011. Science+Business Media.

and sharp, unlike the Tamai, Venkatraman stent models with a broad  $T_g$  range. Hence Yang's stent has shown a fast response and completed self-activation within 25 seconds in the body fluid environment. The cyclic thermo-mechanical tensile test results have concluded that hyperbranched PCL stent had higher  $R_f$  due to higher chain fixing ability against slippery. Similarly, the  $R_r$  i.e. proofed self-expansion ability of the vascular stent was experimentally evaluated. Finally, Xue *et al.* [47] have concluded that the hyperbranched PCL has shown remarkable switching ability and low stent triggering temperature.

In further development, Yang *et al.* [88] manufactured a thermally induced shape memory dual drug-eluting biodegradable stent by cross-linking PEG and PCL co-polymers (Figure 12). PCL is highly durable in a neutral environment which has a life time around two years; however, PEG incorporated polymer chains improved the degradation properties and improved the usability of co-polymer [106]. In addition, the reduction of crystallinity extensively has caused a significant drop in the melting temperature of the co-polymer. The proposed stent by Yang *et al.* [88] has addressed two major drawbacks of the current stents. The Yang's stent underwent a major geometrical transformation with external thermal stimuli.



**Figure 12.** Images of shape-memory effect of a cross-linked PEG-PCL<sub>6040</sub> stent. (a) Permanent shape of the stent, (b) temporary shape of the stent, (c) recovery to the permanent shape. Reproduced from [88] Copyright 2013. Applied Materials & Interfaces.

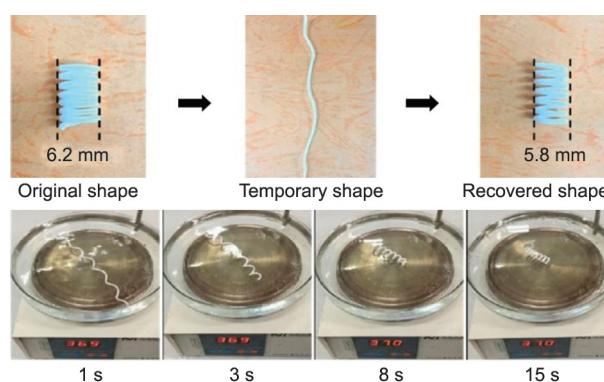
The stent can provide two different drugs to the bloodstream, unlike well-known Igaki-Tamai, Venkatraman, Chen and Yang vascular stents. This may prevent initial platelet adhesion and thrombosis formation. Interestingly, a temporary flat-shaped strip will secure a permanent spiral shape upon exposure to thermal energy from the body at 40 °C within 10 seconds and has shown  $R_f$  and  $R_r > 99\%$  and  $> 90\%$ , respectively. Yang *et al.* [88] have concluded that the increased

PCL portion in the composite result higher  $R_r$  and faster recovery time. The vascular stent degradation studies were carried out experimentally for both stationary and under flow conditions and observed a weight loss of the specimen. After two months, the stent has lost 47% weight without showing any corrosion or structural deformations. Hence Yang's vascular stent has shown greater potential for application in minimally invasive treatments.

New thermo responsive SMP was synthesised by Boire *et al.* [89] in 2015 and has demonstrated robust, facile and readily tuneable SMP properties. The SMP was mainly PCL and poly( $\alpha$ -allyl carboxylate- $\epsilon$ -caprolactone) (ACPCL). PCL has shown comparable properties for vascular stenting applications. Most researchers frequently used this material as a base for new material synthesis. However,  $T_m$  of the material is ( $>50^\circ\text{C}$ ), which is unsafe for body cells, but Boire *et al.* [89] have managed to reduce the triggering temperature down to  $37^\circ\text{C}$ . It has been shown that triggering temperature can be fine-tuned by varying molecular weight and gel content. A *vivo* study was conducted with A/J mice thigh muscle closer to femoral artery vein with one end closed. A 89% PCL and 11% ACPCL co-polymer with a 1.5 cm length and 1.2 cm outer diameter SMP circular stent was implanted and kept for two weeks before further investigations. Boire *et al.* [89] observed successful cellularisation and fibrotic tissue formation during the investigation. However, they have concluded that long term *vivo* studies and further experiments will be required before human use.

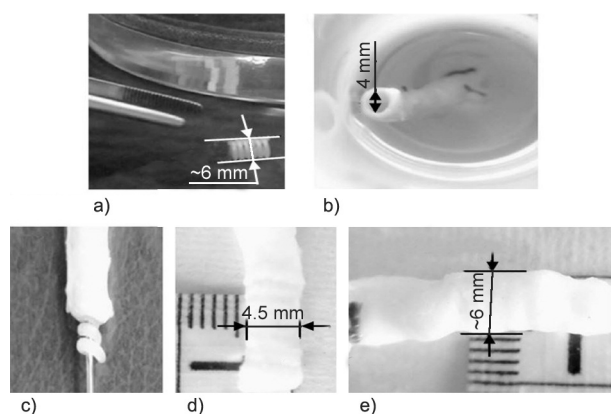
The research data from the recent past clearly have shown that scientists have overcome the remaining problems of control activation temperature of PCL based SMP vascular stents. Most of the developed vascular stents are triggered near body temperatures. Another biocompatible and biodegradable self-expandable vascular stent was developed by Zheng *et al.* [44] using the melt blending method. Most popular biomaterials, poly (propylene carbonate) (PPC) and polycaprolactone (PCL), were used during the fabrication process [107]. The PPC was melt blended with required PCL proportions through a twin-screw extruder and hot-pressed into a 0.8 mm thin film. Additionally, 5% by weight of antihypertensive drug Metoprolol tartrate (MPT) was also introduced to the melt blend before it went through the extruder. DSC results have verified semicrystalline PCL and amorphous PPC having a single melting peak at  $57^\circ\text{C}$  and

single  $T_g$   $33^\circ\text{C}$ , respectively. The blended composite recorded both  $T_m$  and  $T_g$  although increasing PCL concentration slightly increased  $T_g$  while  $T_m$  remained constant. On the other hand, the addition of PCL declined the  $R_f$  and increased  $R_r$ . Hence shape memory performance of the material depended on PCL volume percentages. The optimal  $R_r$  and  $R_f$  achieved at 25% volume percentage of PCL and new material was named as PL-25. Most importantly  $T_g$  of PL-25 was closer to body temperature which makes PL-25 was a successful candidate for fabricate vascular stent. According to Zheng *et al.* [44] experimental results, the vascular stent was submerged in  $37^\circ\text{C}$  water and original shape was recovered within 15 seconds (Figure 13). The recent results have improved considerably in triggering time and temperature compared to the initial stent properties. Blood compatibility was analysed during the *vitro* experiment for platelet adhesion, clotting time, and platelet activation tests with human blood plasma. The MPT drug release rate of the vascular stent was impressive. The initial eluting rate was approximately 22% within the first five days, which is a relatively high value. Afterwards the stent showed a linear slower rate and almost 60% of the drug was eluted within two months. Comparatively, the vascular stent made by Guo *et al.* [108] had a long-term drug release period. According to the *vitro* results, PPC degrades faster than PCL due to PPC's amorphous structure and low  $T_g$ . Zheng *et al.* [44] found that the mass loss of the vascular stent was controlled by the ratio of PCL to PPC by weight. Hence increasing the ratio of PLC, it was possible to enhance the degradation time, which can provide excellent structural stability for an extended period.



**Figure 13.** Images of the shape memory cycle of the PL-25 stent recorded by a digital camera. Reproduced from [44] Copyright 2017. Applied materials & Interfaces.

SMP vascular stent development moved into a new direction after PCL and PU based stents were made with the melt blending method by Ajili *et al.* [90]. Due to PCL and PU, biocompatibility and biodegradability Ajili *et al.* [90] had a higher expectation to avoid most of the drawbacks remaining in vascular stent development. For example, human tissues are not able to withstand PCL's higher melting temperature ( $\sim 60^\circ\text{C}$ ); however, they have found that the transition temperature decreased with the PU content increased [109]. Due to this reason, PCL was introduced to PU melted chamber before it passed through the compression mould. According to thermomechanical results, the optimum properties were obtained with PU/PCL:70/30 by weight ratios and  $T_g$  was found around  $37^\circ\text{C}$ , which proved the stent's functional temperature closer to body temperature (Figure 14). The performance of the vascular stent was evaluated in the human femoral vein, which has a reasonably large nominal diameter. The stent had sufficient hoop strength to expand the vessel, as shown in Figure 14. The study conducted *in vitro* in human bone marrow mesenchymal stem cells (HBMSCs) has proved the biocompatibility of vascular stents by supporting cell adhesion and proliferation. Further, Ajili *et al.* [90] have shown much confidence that their research would be ground-breaking discoveries for the future generation of SMP vascular stents.



**Figure 14.** Images of the proposed screw stent made of PU/PCL (70/30) blend (a) in a segment of femoral vein removed from the leg of a patient (b) stent in temporary shape (a screw with very small diameter) over a delivery instrument (c) recovery step of the SMP stent delivered into the segment of 4 mm ID human femoral vein containing body temperature water at  $37^\circ\text{C}$  (d, e). The filament diameter of the stent is  $800\ \mu\text{m}$ . Reproduced from [90] Copyright 2009. Acta Biomaterialia.

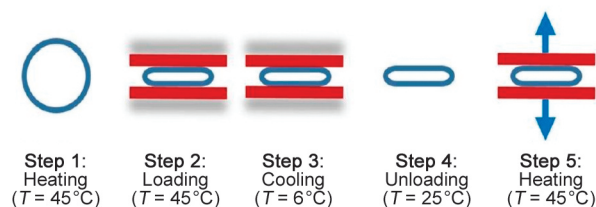
Most researchers experimentally validated their specimens by subjecting them to simple axial loading, but behaviour under torsional loading was rarely observed [39, 44, 110, 111]. In early 2018, Ansari *et al.* [91] experimentally studied PCL and PU combined material properties with combined tension-torsional loading. In contrast to Zheng *et al.* [44] and Ajili *et al.* [90] Ansari's group mixed and blended the materials. In-depth thermo-mechanical experimental data analysis revealed that the blend mixing method has caused higher storage and loss modulus than melt mixing techniques [44, 91]. The Dimethylformamide (DMF) was heated to  $70^\circ\text{C}$  and mixed with PU for around 12 hours before adding PCL. The homogeneous solution was kept in the vacuum oven at the same temperature for 8 hours until DMF evaporated and cast into a 0.4 mm thick sheet. To investigate the effect of moisture during biomedical application and maintain a uniform heat transfer rate, SME was experimented with inside a water bath [112]. The material was pre-stretched up to 25% at a 10 mm/min rate at  $60^\circ\text{C}$  temperature. A torsion load was applied in the combined loading experiment until the material strip achieved  $720^\circ$  rotation and pre-stretched to 25%. The torsional and tension load rates were  $45^\circ/\text{s}$  and 5 mm/s at temperature  $60^\circ\text{C}$  and both loads were applied subsequently. The heating rate greatly affected the recovery finish temperature (RFT) [113]. Further, Ansari *et al.* [91] observed that the pre-stretch and recovery start temperature (RST) have an inverse relationship and the RFT was relatively similar in all the samples. Higher pre-stretch affected Angle recovery and normal strain recovery was affected by higher pre-torsion in combined load cases. The measured transition temperature was dominated by PCL crystals in the blended sample and was observed in the range of  $(42\text{--}68)^\circ\text{C}$ . This temperature range was significantly higher for biomedical applications. Moreover, by altering pre-stretch and deformation temperature ( $T_d$ ), it was possible to control RST and maximum recovery of the strip. Hence, Ansari *et al.* [91] concluded that setting up  $T_d$  at  $40^\circ\text{C}$  can be used to manufacture sensitive biomedical devices such as vascular stents. Ajili *et al.* [90] and Ansari *et al.*'s [91] research were based on PCL and PU materials. The research outcomes are not providing in-depth knowledge about the performance of the materials as Ansari *et al.* [91] focused more on pre-stretch and pre-torsion recovery behaviour. Continuing with the original

work, Ansari *et al.* [114] further studied PU/PCL SMP tubular stent under radial compression and radial force recovery (Figure 15). Ajili *et al.* [110] studied a melt-mixing and solution mixing method to fabricate vascular stents with the weight ratio of PU/PCL:7/3. In the previous experiment, Ansari *et al.* [91] has not provided the details of PU type. However, in the current analysis, two types of PU, LARIPUR 107-93A and LARIPUR 2102-85AE have been used as hard segments with Young's modulus 9.0, 4.3 MPa, respectively. During the analysis, shape memory behaviour was not observed in LARIPUR 107-93A due to the high elastic modulus. In the angle recovery test, melt-mixed samples have better performances due to their lower storage and loss modulus than other solution-mixing samples. During thermomechanical testing such as dynamic mechanical analyser (DMA), Ansari *et al.* [114] found that relaxation was highly dependent on the diameter/thickness ratio of the samples. By increasing the diameter/thickness ratio, the force required to recover has decreased, but RST increased. Moreover, minimum RST was achieved from minimum diameter/thickness ratios. Hence, Ansari *et al.* [114] concluded that the geometrical changes greatly affected on recovery response. Therefore, performance of the vascular stent mainly depended on both geometry and loading condition and the biological environment.

The use of PCL for bio-medical devices was a turning point in bio-material development research. The latest PCL literature focused on degradable properties. Chasse *et al.* [52] have 3D printed a helical flow device and studied the degradation of PCL polymeric samples under a helical fluid flow pattern in 2019. In three flat PCL  $1.000 \times 2.500 \times 0.125$  cm<sup>3</sup> samples, one face was subjected to an artificially generated fluid flow. The helical flow device was 3D printed with acrylonitrile-butadiene-styrene (ABS), and the device's structural integrity was enhanced with polydimethylsiloxane (PDMS). The flow device was kept

inside the testing chamber and immersed in a 37 °C static water bath. The phosphate-buffered saline (PBS) was circulated approximately 1 cm<sup>3</sup>/s and volume flow rate, pH and temperature were controlled. PCL samples were tested for mass loss, molecular weight loss and crystallinity yield were given linear degradation, which can estimate total degradation time. Further, the molecular weight has decreased by 22 and 5% and crystallinity has increased by 16 and 6% in the generated flow conditions. This experimental degradation property needs further study before setting fabricating parameters of vascular stents. The work that has been completed to date has verified that PCL activation temperature can be varied upon different co-polymers, which is a breakthrough in the development of bio-compatible material research.

Park *et al.* [92] applied SMP stenting to treat nasolacrimal duct obstruction (epiphora). A silicone tube was used in the conventional treatment method, but inserting through a narrow complex-shaped nasolacrimal duct was challenging. The silicone products were used in the biomedical field for a long time, but hydrophobicity restrained their use in water-absorbing applications [115]. Hence, Park *et al.* [92] proposed a self-expandable SMP vascular stent with a diameter of 0.42 mm to solve this problem. The stent material was synthesised with PCL and poly(glycidyl methacrylate) PGMA ratios of 94 and 6%, respectively. Park *et al.* [92] simulated nasolacrimal, assuming a straight tube with a CFX module of ANSYS V.18. In addition to that Park *et al.* [92] managed to adjust triggering temperature (32–44) °C controlling the crystallinity and net-point spacing by varying PCL/PGMA molar and cross-linking ratios. The vascular stent was easy to insert and manoeuvre. In addition to the enhanced draining capacity, the vascular stent is resistant to bacterial infection. The performance was observed in a short period with rabbits and Park *et al.* [92] intended to conduct a proper animal model in the future and verify the feasibility for human use.



**Figure 15.** Images of force recovery test of the tubular stent. Reproduced from [114] Copyright 2019. Smart Materials and Structures.

## 6. Challenges and future directions

Stenting, a well-known clinically proven medical technique, has been used for three decades and has saved many human lives. The development of minimally invasive surgical procedures challenged existing stents due to their geometrical dimensions. Therefore, researchers were enthusiastic, focused

and committed to the develop biocompatible and biodegradable SMP stents [116].

Most of the researchers were focused on PLA, PCL, and PU SMPs. The dip coating, solvent casting, injection moulding, melt and blending, blend and melting fabrication techniques have been widely used. Even though SMPs can be activated with different triggering mechanisms, the most developed vascular stents were triggered on body thermal energy. One major challenge was that polymeric stents are prone to fail due to their intrinsic weak mechanical strength. However, their softness will ensure easier maneuvering within arterial walls and cause minor arterial injuries [87]. Hence stents mechanical properties needed to be optimised between the required strength and the softness of the material. The current studies have validated SMP vascular stents performances under static conditions. However, vascular stents will be installed in blood vessels and subjected to physiological reactions with combined loads and stress unconditionally [52]. Hence, critical investigation of SMP vascular stents under actual operational working conditions is needed. Nevertheless, no thermo-mechanical standards for stent materials have been published by any governing body to date [88].

SMP degradation properties can be optimised by varying chemical composition, applying polymer coatings and better material processing techniques. During the degradation, the strength of the stent weakens. Consequently, a sudden collapse of the stent may cause blood-stream blockage and cause life-threatening risks to the patient. Biofilm formation is another major hurdle in the SMP vascular stent clearance process. The fabrication of the inside surface of the stent to better mimic the actual human vein conditions is a challenge as the incompatible surface conditions might adversely affect the blood flow. The lack of knowledge in biodegradability, durability, reliability, availability, and cost adversely affected SMP vascular stents development [43].

It has been found that none of the recently developed stents have been certified by global governing medical regulatory authorities. On the other hand, it is difficult to evaluate and predict the behaviour of foreign implanted synthetic parts inside the human body. Therefore, researchers often conduct *vitro* and *vivo* animal models to validate performance under specific biological conditions. Moreover, long-term

functional analysis has not been done to the satisfaction of approval bodies. Therefore, the durability of synthesised vascular stents was challenged during the FDA certification process. Due to these reasons, many successful and breakthrough research outcomes are commercially not available to date.

However, the authors believe SMP can offer a cost-effective, sustainable replacement for existing metallic and polymeric stents. The recently developed SMP vascular stents have demonstrated an excellent capability to overcome the potential complications in treatment. A more systematic approach and well-defined framework is warranted in SMP-based vascular stent development to provide reliable and cost-effective medical solutions for human wellbeing.

## References

- [1] Lendlein A., Kelch S.: Shape-memory polymers. *Angewandte Chemie*, **41**, 2034–2057 (2002).  
[https://doi.org/10.1002/1521-3773\(20020617\)41:12<2034::AID-ANIE2034>3.0.CO;2-M](https://doi.org/10.1002/1521-3773(20020617)41:12<2034::AID-ANIE2034>3.0.CO;2-M)
- [2] Hager M. D., Bode S., Weber C., Schubert U. S.: Shape memory polymers: Past, present and future developments. *Progress in Polymer Science*, **49–50**, 3–33 (2015).  
<https://doi.org/10.1016/j.progpolymsci.2015.04.002>
- [3] Glock S., Canal L. P., Grize C. M., Michaud V.: Magneto-mechanical actuation of ferromagnetic shape memory alloy/epoxy composites. *Composites Science and Technology*, **114**, 110–118 (2015).  
<https://doi.org/10.1016/j.compscitech.2015.04.009>
- [4] Yang D., Huang W., He X., Xie M.: Electromagnetic activation of a shape memory copolymer matrix incorporating ferromagnetic nanoparticles. *Polymer International*, **61**, 38–42 (2012).  
<https://doi.org/10.1002/pi.3188>
- [5] Zou H., Weder C., Simon Y. C.: Shape-memory polyurethane nanocomposites with single layer or bilayer oleic acid-coated Fe<sub>3</sub>O<sub>4</sub> nanoparticles. *Macromolecular Materials and Engineering*, **300**, 885–892 (2015).  
<https://doi.org/10.1002/mame.201500079>
- [6] Sahoo N. G., Jung Y. C., Cho J. W.: Electroactive shape memory effect of polyurethane composites filled with carbon nanotubes and conducting polymer. *Materials and Manufacturing Processes*, **22**, 419–423 (2007).  
<https://doi.org/10.1080/10426910701232857>
- [7] Xie T.: Tunable polymer multi-shape memory effect. *Nature*, **464**, 267–270 (2010).  
<https://doi.org/10.1038/nature08863>
- [8] Xu J., Song J.: High performance shape memory polymer networks based on rigid nanoparticle cores. *Proceedings of the National Academy of Sciences*, **107**, 7652–7657 (2010).  
<https://doi.org/10.1073/pnas.0912481107>

- [9] Mohr R., Kratz K., Weigel T., Lucka-Gabor M., Moneke M., Lendlein A.: Initiation of shape-memory effect by inductive heating of magnetic nanoparticles in thermoplastic polymers. *Proceedings of the National Academy of Sciences*, **103**, 3540–3545 (2006).  
<https://doi.org/10.1073/pnas.0600079103>
- [10] Li W., Liu Y., Leng J.: Selectively actuated multi-shape memory effect of a polymer multicomposite. *Journal of Materials Chemistry A*, **3**, 24532–24539 (2015).  
<https://doi.org/10.1039/c5ta08513f>
- [11] Leng J., Lan X., Liu Y., Du S.: Shape-memory polymers and their composites: Stimulus methods and applications. *Progress in Materials Science*, **56**, 1077–1135 (2011).  
<https://doi.org/10.1016/j.pmatsci.2011.03.001>
- [12] Rose A., Zhu Z., Madigan C. F., Swager T. M., Bulović V.: Sensitivity gains in chemosensing by lasing action in organic polymers. *Nature*, **434**, 876–879 (2005).  
<https://doi.org/10.1038/nature03438>
- [13] Herath H. M. C. M., Epaarachchi J. A., Islam M. M., Al-Azzawi W., Leng J., Zhang F.: Structural performance and photothermal recovery of carbon fibre reinforced shape memory polymer. *Composites Science and Technology*, **167**, 206–214 (2018).  
<https://doi.org/10.1016/j.compscitech.2018.07.042>
- [14] Yu K., Liu Y., Leng J.: Conductive shape memory polymer composite incorporated with hybrid fillers: Electrical, mechanical, and shape memory properties. *Journal of Intelligent Material Systems and Structures*, **22**, 369–379 (2011).  
<https://doi.org/10.1177/1045389x11401452>
- [15] Jiang H. Y., Kelch S., Lendlein A.: Polymers move in response to light. *Advanced Materials*, **18**, 1471–1475 (2006).  
<https://doi.org/10.1002/adma.200502266>
- [16] Safranski D. L., Smith K. E., Gall K.: Mechanical requirements of shape-memory polymers in biomedical devices. *Polymer Reviews*, **53**, 76–91 (2013).  
<https://doi.org/10.1080/15583724.2012.752385>
- [17] Correia C. O., Mano J. F.: Chitosan scaffolds with a shape memory effect induced by hydration. *Journal of Materials Chemistry B*, **2**, 3315–3323 (2014).  
<https://doi.org/10.1039/c4tb00226a>
- [18] Zhao G., Wang Z., Wang H., Zhao H., Fu Y., Yang J.: Effect of doping nanoparticles on the output force performance of chitosan-based nanocomposite gel actuator. *Polymer-Plastics Technology and Materials*, **58**, 967–977 (2019).  
<https://doi.org/10.1080/03602559.2018.1520258>
- [19] Sokolowski W. M.: Shape memory polymer foam and applications. in ‘Shape-memory polymers and multifunctional composites’ (eds.: Leng J., Du S.) CRC Press, Boca Raton, 277–302 (2010).
- [20] Behl M., Lendlein A.: Shape-memory polymers. *Materials Today*, **10**, 20–28 (2007).  
[https://doi.org/10.1016/s1369-7021\(07\)70047-0](https://doi.org/10.1016/s1369-7021(07)70047-0)
- [21] Behl M., Razzaq M. Y., Lendlein A.: Multifunctional shape-memory polymers. *Advanced Materials*, **22**, 3388–3410 (2010).  
<https://doi.org/10.1002/adma.200904447>
- [22] Wang Z., Song W., Ke L., Wang Y.: Shape memory polymer composite structures with two-way shape memory effects. *Materials Letters*, **89**, 216–218 (2012).  
<https://doi.org/10.1016/j.matlet.2012.08.112>
- [23] Jia H., Chang K., Gu S.-Y.: Synthesis and properties of reversible disulfide bond-based self-healing polyurethane with triple shape memory properties. *Chinese Journal of Polymer Science*, **37**, 1119–1129 (2019).  
<https://doi.org/10.1007/s10118-019-2268-2>
- [24] Bellin I., Kelch S., Langer R., Lendlein A.: Polymeric triple-shape materials. *Proceedings of the National Academy of Sciences*, **103**, 18043–18047 (2006).  
<https://doi.org/10.1073/pnas.0608586103>
- [25] Yang X., Wang L., Wang W., Chen H., Yang G., Zhou S.: Triple shape memory effect of star-shaped polyurethane. *ACS Applied Materials and Interfaces*, **6**, 6545–6554 (2014).  
<https://doi.org/10.1021/am5001344>
- [26] Lendlein A., Gould O. E. C.: Reprogrammable recovery and actuation behaviour of shape-memory polymers. *Nature Reviews Materials*, **4**, 116–133 (2019).  
<https://doi.org/10.1038/s41578-018-0078-8>
- [27] Meng H., Li G.: A review of stimuli-responsive shape memory polymer composites. *Polymer*, **54**, 2199–2221 (2013).  
<https://doi.org/10.1016/j.polymer.2013.02.023>
- [28] Ohki T., Ni Q.-Q., Ohsako N., Iwamoto M.: Mechanical and shape memory behavior of composites with shape memory polymer. *Composites Part A: Applied Science and Manufacturing*, **35**, 1065–1073 (2004).  
<https://doi.org/10.1016/j.compositesa.2004.03.001>
- [29] Lan X., Liu Y., Lv H., Wang X., Leng J., Du S.: Fiber reinforced shape-memory polymer composite and its application in a deployable hinge. *Smart Materials and Structures*, **18**, 024002 (2009).  
<https://doi.org/10.1088/0964-1726/18/2/024002>
- [30] Duarah R., Singh Y. P., Gupta P., Mandal B. B., Karak N.: High performance bio-based hyperbranched polyurethane/carbon dot-silver nanocomposite: A rapid self-expandable stent. *Biofabrication*, **8**, 45013 (2016).  
<https://doi.org/10.1088/1758-5090/8/4/045013>
- [31] Gu S.-Y., Jin S.-P., Gao X.-F., Mu J.: Polylactide-based polyurethane shape memory nanocomposites (Fe<sub>3</sub>O<sub>4</sub>/PLAUs) with fast magnetic responsiveness. *Smart Materials and Structures*, **25**, 055036 (2016).  
<https://doi.org/10.1088/0964-1726/25/5/055036>
- [32] Wang H., Fang L., Zhang Z., Epaarachchi J., Li L., Hu X., Lu C., Xu Z.: Light-induced rare earth organic complex/shape-memory polymer composites with high strength and luminescence based on hydrogen bonding. *Composites Part A: Applied Science and Manufacturing*, **125**, 105525 (2019).  
<https://doi.org/10.1016/j.compositesa.2019.105525>

- [33] Ahmad M., Luo J., MirafTAB M.: Feasibility study of polyurethane shape-memory polymer actuators for pressure bandage application. *Science and Technology of Advanced Materials*, **13**, 15006 (2012).  
<https://doi.org/10.1088/1468-6996/13/1/015006>
- [34] Lorenzo V., Díaz-Lantada A., Lafont P., Lorenzo-Yustos H., Fonseca C., Acosta J.: Physical ageing of a PU-based shape memory polymer: Influence on their applicability to the development of medical devices. *Materials and Design*, **30**, 2431–2434 (2009).  
<https://doi.org/10.1016/j.matdes.2008.10.023>
- [35] Yu X., Wang L., Huang M., Gong T., Li W., Cao Y., Ji D., Wang P., Wang J., Zhou S.: A shape memory stent of poly( $\epsilon$ -caprolactone-co-DL-lactide) copolymer for potential treatment of esophageal stenosis. *Journal of Materials Science: Materials in Medicine*, **23**, 581–589 (2012).  
<https://doi.org/10.1007/s10856-011-4475-4>
- [36] Dyamenahalli K., Famili A., Shandas R.: Characterization of shape-memory polymers for biomedical applications. in ‘Shape memory polymers for biomedical applications’ (ed.: L’Hocine Y.) Woodhead, Cambridge, 35–63 (2015).  
<https://doi.org/10.1016/B978-0-85709-698-2.00003-9>
- [37] Neffe A. T., Hanh B. D., Steuer S., Lendlein A.: Polymer networks combining controlled drug release, biodegradation, and shape memory capability. *Advanced Materials*, **21**, 3394–398 (2009).  
<https://doi.org/10.1002/adma.200802333>
- [38] Wischke C., Neffe A. T., Steuer S., Lendlein A.: Evaluation of a degradable shape-memory polymer network as matrix for controlled drug release. *Journal of Controlled Release*, **138**, 243–250 (2009).  
<https://doi.org/10.1016/j.jconrel.2009.05.027>
- [39] Yakacki C. M., Shandas R., Lanning C., Rech B., Eckstein A., Gall K.: Unconstrained recovery characterization of shape-memory polymer networks for cardiovascular applications. *Biomaterials*, **28**, 2255–2263 (2007).  
<https://doi.org/10.1016/j.biomaterials.2007.01.030>
- [40] Yakacki C. M., Shandas R., Safranski D., Ortega A. M., Sassaman K., Gall K.: Strong, tailored, biocompatible shape-memory polymer networks. *Advanced Functional Materials*, **18**, 2428–2435 (2008).  
<https://doi.org/10.1002/adfm.200701049>
- [41] Sharp A. A., Panchawagh H. V., Ortega A., Artale R., Richardson-Burns S., Finch D. S., Gall K., Mahajan R. L., Restrepo D.: Toward a self-deploying shape memory polymer neuronal electrode. *Journal of Neural Engineering*, **3**, L23–L30 (2006).  
<https://doi.org/10.1088/1741-2560/3/4/L02>
- [42] Venkatraman S. S., Tan L. P., Joso J. F., Boey Y. C., Wang X.: Biodegradable stents with elastic memory. *Biomaterials*, **27**, 1573–1578 (2006).  
<https://doi.org/10.1016/j.biomaterials.2005.09.002>
- [43] Tamai H., Igaki K., Kyo E., Kosuga K., Kawashima A., Matsui S., Komori H., Tsuji T., Motohara S., Uehata H.: Initial and 6-month results of biodegradable poly-L-lactic acid coronary stents in humans. *Circulation*, **102**, 399–404 (2000).  
<https://doi.org/10.1161/01.CIR.102.4.399>
- [44] Zheng Y., Li Y., Hu X., Shen J., Guo S.: Biocompatible shape memory blend for self-expandable stents with potential biomedical applications. *ACS Applied Materials and Interfaces*, **9**, 13988–13998 (2017).  
<https://doi.org/10.1021/acsami.7b04808>
- [45] Sokolowski W., Metcalfe A., Hayashi S., Yahia L., Raymond J.: Medical applications of shape memory polymers. *Biomedical Materials*, **2**, S23 (2007).  
<https://doi.org/10.1088/1748-6041/2/1/S04>
- [46] Wong Y., Kong J., Widjaja L. K., Venkatraman S. S.: Biomedical applications of shape-memory polymers: How practically useful are they? *Science China Chemistry*, **57**, 476–489 (2014).  
<https://doi.org/10.1007/s11426-013-5061-z>
- [47] Xue L., Dai S., Li Z.: Biodegradable shape-memory block co-polymers for fast self-expandable stents. *Biomaterials*, **31**, 8132–8140 (2010).  
<https://doi.org/10.1016/j.biomaterials.2010.07.043>
- [48] Muzaffar A., Deshmukh K., Basheer Ahamed M., Khadheer Pasha S. K.: Shape memory polymer composites in biomedical field. in ‘Polymer nanocomposites in biomedical engineering’ (eds.: Sadasivuni K. K., Ponnammma D., Rajan M., Ahmed B., Al-Maadeed M. A. S. A.) Springer, Cham, 299–329 (2019).
- [49] Ghasemi-Mobarakeh L., Kolahreza D., Ramakrishna S., Williams D.: Key terminology in biomaterials and biocompatibility. *Current Opinion in Biomedical Engineering*, **10**, 45–50 (2019).  
<https://doi.org/10.1016/j.cobme.2019.02.004>
- [50] Hardy J. G., Palma M., Wind S. J., Biggs M. J.: Responsive biomaterials: Advances in materials based on shape-memory polymers. *Advanced Materials*, **28**, 5717–5724 (2016).  
<https://doi.org/10.1002/adma.201505417>
- [51] Safranski D. L., Griffis J. C.: Shape-memory polymer device design. William Andrew, Cambridge (2017).
- [52] Chasse B., Xu H., Budhlall B. M.: *In-vitro* biodegradation study of poly( $\epsilon$ -caprolactone) films using a 3D printed helical flow prototype to simulate the physiological conditions for cardiovascular implanted devices. *Biomedical Physics and Engineering Express*, **5**, 65021 (2019).  
<https://doi.org/10.1088/2057-1976/ab4e2b>
- [53] Madbouly S., Alauzen T., Ross S.: Biodegradable shape-memory polymers and composites. in ‘Biopolymers and composites’ (eds.: Madbouly S. A., Zhang C.) Springer, Singapore, 331–352 (2020).  
<https://doi.org/10.1515/9781501521942-012>

- [54] Du L., Yang S., Li W., Li H., Feng S., Zeng R., Yu B., Xiao L., Nie H-Y., Tu M.: Scaffold composed of porous vancomycin-loaded poly(lactide-co-glycolide) microspheres: A controlled-release drug delivery system with shape-memory effect. *Materials Science and Engineering: C*, **78**, 1172–1178 (2017).  
<https://doi.org/10.1016/j.msec.2017.04.099>
- [55] Lendlein A., Behl M., Hiebl B., Wischke C.: Shape-memory polymers as a technology platform for biomedical applications. *Expert Review of Medical Devices*, **7**, 357–379 (2010).  
<https://doi.org/10.1586/erd.10.8>
- [56] Safranski D. L., Griffis J. C.: Applications of shape-memory polymers. in ‘Shape-memory polymer device design’ (eds.: Safranski D. L., Griffis J. C.) Elsevier, Oxford, 189–222 (2017).  
<https://doi.org/10.1016/B978-0-323-37797-3.00006-3>
- [57] Fu Y. Q., Huang W. M., Luo J., Lu H.: Polyurethane shape-memory polymers for biomedical applications. in ‘Shape memory polymers for biomedical applications’ (ed.: L’Hocine Y.) Woodhead, Cambridge, 167–195 (2015).  
<https://doi.org/10.1016/B978-0-85709-698-2.00009-X>
- [58] Boddu A. R., Balmuri A. R., Kamalesh M.: Clinical utility of self-expanding stents in coronary artery disease. *Research Reports in Clinical Cardiology*, **6**, 117–122 (2015).  
<https://doi.org/10.2147/RRCC.S51653>
- [59] Duerig T., Wholey M.: A comparison of balloon-and self-expanding stents. *Minimally Invasive Therapy and Allied Technologies*, **11**, 173–178 (2002).  
<https://doi.org/10.1080/136457002760273386>
- [60] Sigwart U., Puel J., Mirkovitch V., Joffre F., Kappenberg L.: Intravascular stents to prevent occlusion and re-stenosis after transluminal angioplasty. *New England Journal of Medicine*, **316**, 701–706 (1987).  
<https://doi.org/10.1056/nejm198703193161201>
- [61] Wache H. M., Tartakowska D. J., Hentrich A., Wagner M. H.: Development of a polymer stent with shape memory effect as a drug delivery system. *Journal of Materials Science: Materials in Medicine*, **14**, 109–112 (2003).  
<https://doi.org/10.1023/A:1022007510352>
- [62] Wu T., Chen X., Fan D., Pang X.: Development and application of metal materials in terms of vascular stents. *Bio-medical Materials and Engineering*, **25**, 435–441 (2015).  
<https://doi.org/10.3233/BME-151542>
- [63] Tsimikas S.: Drug-eluting stents and late adverse clinical outcomes: Lessons learned, lessons awaited. *Journal of the American College of Cardiology*, **47**, 2112–2115 (2006).  
<https://doi.org/10.1016/j.jacc.2006.03.019>
- [64] Köster R., Vieluf D., Kiehn M., Sommerauer M., Kähler J., Baldus S., Meinertz T., Hamm C. W.: Nickel and molybdenum contact allergies in patients with coronary in-stent restenosis. *The Lancet*, **356**, 1895–1897 (2000).  
[https://doi.org/10.1016/S0140-6736\(00\)03262-1](https://doi.org/10.1016/S0140-6736(00)03262-1)
- [65] Novitzke J.: A patient guide to brain stent placement. *Journal of Vascular and Interventional Neurology*, **2**, 177–179 (2009).
- [66] Forbes T. J., Kim D. W., Du W., Turner D. R., Holzer R., Amin Z., Hijazi Z., Ghasemi A., Rome J. J., Nykanen D., Zahn E., Cowley C., Hoyer M., Waight D., Gruenstein D., Javois A., Foerster S., Kreutzer J., Sullivan N., Khan A., Owada C., Hagler D., Lim S., Canter J., Zellers T., Investigators C.: Comparison of surgical, stent, and balloon angioplasty treatment of native coarctation of the aorta: An observational study by the CCISC (congenital cardiovascular interventional study consortium). *Journal of the American College of Cardiology*, **58**, 2664–2674 (2011).  
<https://doi.org/10.1016/j.jacc.2011.08.053>
- [67] Jeewandara T. M., Wise S. G., Ng M. K. C.: Biocompatibility of coronary stents. *Materials*, **7**, 769–786 (2014).  
<https://doi.org/10.3390/ma7020769>
- [68] Liu R., McGinty S., Cui F., Luo X., Liu Z.: Modelling and simulation of the expansion of a shape memory polymer stent. *Engineering Computations*, **36**, 2726–2746 (2019).  
<https://doi.org/10.1108/ec-10-2018-0462>
- [69] Sonawane V. C., More M. P., Pandey A. P., Patil P. O., Deshmukh P. K.: Fabrication and characterization of shape memory polymers based bioabsorbable biomedical drug eluting stent. *Artificial Cells Nanomedicine and Biotechnology*, **45**, 1740–1750 (2017).  
<https://doi.org/10.1080/21691401.2017.1282867>
- [70] Lee Y-K., Park J. H., Moon H. T., Lee D. Y., Yun J. H., Byun Y.: The short-term effects on restenosis and thrombosis of echinomycin-eluting stents topcoated with a hydrophobic heparin-containing polymer. *Bio-materials*, **28**, 1523–1530 (2007).  
<https://doi.org/10.1016/j.biomaterials.2006.11.020>
- [71] Grabow N., Bünger C. M., Schultze C., Schmohl K., Martin D. P., Williams S. F., Sternberg K., Schmitz K. P.: A biodegradable slotted tube stent based on poly(L-lactide) and poly(4-hydroxybutyrate) for rapid balloon-expansion. *Annals of Biomedical Engineering*, **35**, 2031–2038 (2007).  
<https://doi.org/10.1007/s10439-007-9376-9>
- [72] Uurto I., Mikkonen J., Parkkinen J., Keski-Nisula L., Nevalainen T., Kellomäki M., Törmälä P., Salenius J-P.: Drug-eluting biodegradable poly-D/L-lactic acid vascular stents: An experimental pilot study. *Journal of Endovascular Therapy*, **12**, 371–379 (2005).  
<https://doi.org/10.1583/05-1525.1>

- [73] O'Brien B., Carroll W.: The evolution of cardiovascular stent materials and surfaces in response to clinical drivers: A review. *Acta Biomaterialia*, **5**, 945–958 (2009).  
<https://doi.org/10.1016/j.actbio.2008.11.012>
- [74] Stack R., Califf R., Phillips H., Pryor D., Quigley P., Bauman R., Tchong J., Greenfield J.: Interventional cardiac catheterization at duke medical center-the duke interventional cardiac catheterization program. *American Journal of Cardiology*, **62**, 3F–4F (1988).
- [75] Zilberman M., Schwade N. D., Eberhart R. C.: Protein□loaded bioresorbable fibers and expandable stents: Mechanical properties and protein release. *Journal of Biomedical Materials Research Part B*, **69**, 1–10 (2004).  
<https://doi.org/10.1002/jbm.b.20026>
- [76] Kim J. H., Kang T. J., Yu W.-R.: Simulation of mechanical behavior of temperature-responsive braided stents made of shape memory polyurethanes. *Journal of Biomechanics*, **43**, 632–643 (2010).  
<https://doi.org/10.1016/j.jbiomech.2009.10.032>
- [77] Wei X., Gong C., Gou M., Fu S., Guo Q., Shi S., Luo F., Guo G., Qiu L., Qian Z.: Biodegradable poly(ε-caprolactone)–poly(ethylene glycol) copolymers as drug delivery system. *International Journal of Pharmaceutics*, **381**, 1–18 (2009).  
<https://doi.org/10.1016/j.ijpharm.2009.07.033>
- [78] Guerra A. J., Ciurana J.: Stent's manufacturing field: Past, present, and future prospects. in 'Angiography' (ed.: Pamukcu B.) IntechOpen, London, 41–60 (2018).  
<https://doi.org/10.5772/intechopen.81668>
- [79] Martinez A. W., Chaikof E. L.: Microfabrication and nanotechnology in stent design. *Wiley Interdisciplinary Reviews: Nanomedicine and Nanobiotechnology*, **3**, 256–268 (2011).  
<https://doi.org/10.1002/wnan.123>
- [80] Schreiber F., Schuster P., Borinski M., Vogt F., Blindt R., Gries T.: Improving the mechanical properties of braided shape memory polymer stents by heat setting. *Autex Research Journal*, **10**, 73–76 (2010).
- [81] Parker T., Dave V., Falotico R.: Polymers for drug eluting stents. *Current Pharmaceutical Design*, **16**, 3978–3988 (2010).  
<https://doi.org/10.2174/138161210794454897>
- [82] Lauto A., Ohebshalom M., Esposito M., Mingin J., Li P., Felsen D., Goldstein M., Poppas D.: Self-expandable chitosan stent: Design and preparation. *Biomaterials*, **22**, 1869–1874 (2001).  
[https://doi.org/10.1016/S0142-9612\(00\)00371-9](https://doi.org/10.1016/S0142-9612(00)00371-9)
- [83] Chen M.-C., Tsai H.-W., Chang Y., Lai W.-Y., Mi F.-L., Liu C.-T., Wong H.-S., Sung H.-W.: Rapidly self-expandable polymeric stents with a shape-memory property. *Biomacromolecules*, **8**, 2774–2780 (2007).  
<https://doi.org/10.1021/bm7004615>
- [84] Chen M.-C., Chang Y., Liu C.-T., Lai W.-Y., Peng S.-F., Hung Y.-W., Tsai H.-W., Sung H.-W.: The characteristics and *in vivo* suppression of neointimal formation with sirolimus-eluting polymeric stents. *Biomaterials*, **30**, 79–88 (2009).  
<https://doi.org/10.1016/j.biomaterials.2008.09.006>
- [85] Baer G. M., Small W., Wilson T. S., Bennett W. J., Matthews D. L., Hartman J., Maitland D. J.: Fabrication and *in vitro* deployment of a laser-activated shape memory polymer vascular stent. *BioMedical Engineering OnLine*, **6**, 43 (2007).  
<https://doi.org/10.1186/1475-925X-6-43>
- [86] Zhou B., Liu Y., Wang Z., Leng J.-S.: Modeling the shape memory effect of shape memory polymer. in 'Second International Conference on Smart Materials and Nanotechnology in Engineering. Weihai, China' Vol. **7493**, 749340 (2009).  
<https://doi.org/10.1117/12.837350>
- [87] Gu S.-Y., Chang K., Jin S.-P.: A dual-induced self-expandable stent based on biodegradable shape memory polyurethane nanocomposites (PCLAU/Fe<sub>3</sub>O<sub>4</sub>) triggered around body temperature. *Journal of Applied Polymer Science*, **135**, 45686 (2018).  
<https://doi.org/10.1002/app.45686>
- [88] Yang C.-S., Wu H.-C., Sun J.-S., Hsiao H.-M., Wang T.-W.: Thermo-induced shape-memory PEG-PCL copolymer as a dual-drug-eluting biodegradable stent. *ACS Applied Materials and Interfaces*, **5**, 10985–10994 (2013).  
<https://doi.org/10.1021/am4032295>
- [89] Boire T. C., Gupta M. K., Zachman A. L., Lee S. H., Balikov D. A., Kim K., Bellan L. M., Sung H.-J.: Pendant allyl crosslinking as a tunable shape memory actuator for vascular applications. *Acta Biomaterialia*, **24**, 53–63 (2015).  
<https://doi.org/10.1016/j.actbio.2015.06.004>
- [90] Ajili S. H., Ebrahimi N. G., Soleimani M.: Polyurethane/polycaprolactane blend with shape memory effect as a proposed material for cardiovascular implants. *Acta Biomaterialia*, **5**, 1519–1530 (2009).  
<https://doi.org/10.1016/j.actbio.2008.12.014>
- [91] Ansari M., Golzar M., Baghani M., Soleimani M.: Shape memory characterization of poly(ε-caprolactone) (PCL)/polyurethane (PU) in combined torsion-tension loading with potential applications in cardiovascular stent. *Polymer Testing*, **68**, 424–432 (2018).  
<https://doi.org/10.1016/j.polymertesting.2018.04.032>
- [92] Park J. Y., Lee J. B., Shin W. B., Kang M.-L., Shin Y. C., Son D. H., Yi S. W., Yoon J.-K., Kim J. Y., Ko J., Kim C.-S., Yoon J. S., Sung H.-J.: Nasolacrimal stent with shape memory as an advanced alternative to silicone products. *Acta Biomaterialia*, **101**, 273–284 (2019).  
<https://doi.org/10.1016/j.actbio.2019.11.001>

- [93] Schiffman J. D., Schauer C. L.: Cross-linking chitosan nanofibers. *Biomacromolecules*, **8**, 594–601 (2007).  
<https://doi.org/10.1021/bm060804s>
- [94] Kim S. M., Park K.-S., Lih E., Hong Y. J., Kang J. H., Kim I. H., Jeong M. H., Joung Y. K., Han D. K.: Fabrication and characteristics of dual functionalized vascular stent by spatio-temporal coating. *Acta Biomaterialia*, **38**, 143–152 (2016).  
<https://doi.org/10.1016/j.actbio.2016.04.029>
- [95] Joner M., Finn A. V., Farb A., Mont E. K., Kolodgie F. D., Ladich E., Kutys R., Skoriya K., Gold H. K., Virmani R.: Pathology of drug-eluting stents in humans: Delayed healing and late thrombotic risk. *Journal of the American College of Cardiology*, **48**, 193–202 (2006).  
<https://doi.org/10.1016/j.jacc.2006.03.042>
- [96] Hirano S., Tsuchida H., Nagao N.: *N*-acetylation in chitosan and the rate of its enzymic hydrolysis. *Biomaterials*, **10**, 574–576 (1989).  
[https://doi.org/10.1016/0142-9612\(89\)90066-5](https://doi.org/10.1016/0142-9612(89)90066-5)
- [97] Small W., Buckley P. R., Wilson T. S., Bennett W. J., Hartman J., Saloner D., Maitland D. J.: Shape memory polymer stent with expandable foam: A new concept for endovascular embolization of fusiform aneurysms. *IEEE Transactions on Biomedical Engineering*, **54**, 1157–1160 (2007).  
<https://doi.org/10.1109/TBME.2006.889771>
- [98] Gall K., Yakacki C. M., Liu Y., Shandas R., Willett N., Anseth K. S.: Thermomechanics of the shape memory effect in polymers for biomedical applications. *Journal of Biomedical Materials Research Part A*, **73**, 339–348 (2005).  
<https://doi.org/10.1002/jbm.a.30296>
- [99] Baer G., Wilson T. S., Matthews D. L., Maitland D. J.: Shape-memory behavior of thermally stimulated polyurethane for medical applications. *Journal of Applied Polymer Science*, **103**, 3882–3892 (2007).  
<https://doi.org/10.1002/app.25567>
- [100] Li S., Guo Z., Zhang Y., Xue W., Liu Z.: Blood compatibility evaluations of fluorescent carbon dots. *ACS Applied Materials and Interfaces*, **7**, 19153–19162 (2015).  
<https://doi.org/10.1021/acsami.5b04866>
- [101] Du F., Zhang M., Li X., Li J., Jiang X., Li Z., Hua Y., Shao G., Jin J., Shao Q., Zhou M., Gong A.: Economical and green synthesis of bagasse-derived fluorescent carbon dots for biomedical applications. *Nanotechnology*, **25**, 315702 (2014).  
<https://doi.org/10.1088/0957-4484/25/31/315702>
- [102] Gwon D. I., Lee S. S., Kim E.-Y.: Cefotaxime-eluting covered self-expandable stents in a canine biliary model: Scanning electron microscopic study of biofilm formation. *Acta Radiologica*, **53**, 1127–1132 (2012).  
<https://doi.org/10.1258/ar.2012.120220>
- [103] Yakacki C. M., Satarkar N. S., Gall K., Likos R., Hilt J. Z.: Shape-memory polymer networks with Fe<sub>3</sub>O<sub>4</sub> nanoparticles for remote activation. *Journal of Applied Polymer Science*, **112**, 3166–3176 (2009).  
<https://doi.org/10.1002/app.29845>
- [104] Zeng B., Li Y., Wang L., Zheng Y., Shen J., Guo S.: Body temperature-triggered shape-memory effect *via* toughening sustainable poly(propylene carbonate) with thermoplastic polyurethane: Toward potential application of biomedical stents. *American Chemical Society Sustainable Chemistry and Engineering*, **8**, 1538–1547 (2020).  
<https://doi.org/10.1021/acssuschemeng.9b06080>
- [105] Xue L., Dai S., Li Z.: Synthesis and characterization of three-arm poly( $\epsilon$ -caprolactone)-based poly(ester-urethanes) with shape-memory effect at body temperature. *Macromolecules*, **42**, 964–972 (2009).  
<https://doi.org/10.1021/ma802437f>
- [106] Wagner D. D., Burger P. C.: Platelets in inflammation and thrombosis. *Arteriosclerosis, Thrombosis, and Vascular Biology*, **23**, 2131–2137 (2003).  
<https://doi.org/10.1161/01.ATV.0000095974.95122.EC>
- [107] Sun H., Mei L., Song C., Cui X., Wang P.: The *in vivo* degradation, absorption and excretion of PCL-based implant. *Biomaterials*, **27**, 1735–1740 (2006).  
<https://doi.org/10.1016/j.biomaterials.2005.09.019>
- [108] Guo Q., Knight P. T., Mather P. T.: Tailored drug release from biodegradable stent coatings based on hybrid polyurethanes. *Journal of Controlled Release*, **137**, 224–233 (2009).  
<https://doi.org/10.1016/j.jconrel.2009.04.016>
- [109] Ajili S. H., Ebrahimi N. G., Ansari M.: Rheological study of segmented polyurethane and polycaprolactone blends. *Rheologica Acta*, **47**, 81–87 (2007).  
<https://doi.org/10.1007/s00397-007-0213-8>
- [110] Hu J., Chen W., Fan P., Gao J., Fang G., Cao Z., Peng F.: Uniaxial tensile tests and dynamic mechanical analysis of satin weave reinforced epoxy shape memory polymer composite. *Polymer Testing*, **64**, 235–241 (2017).  
<https://doi.org/10.1016/j.polymertesting.2017.09.038>
- [111] Westbrook K. K., Castro F., Long K. N., Slifka A. J., Qi H. J.: Improved testing system for thermomechanical experiments on polymers using uniaxial compression equipment. *Polymer Testing*, **29**, 503–512 (2010).  
<https://doi.org/10.1016/j.polymertesting.2010.02.011>
- [112] Yang B., Huang W. M., Li C., Li L.: Effects of moisture on the thermomechanical properties of a polyurethane shape memory polymer. *Polymer*, **47**, 1348–1356 (2006).  
<https://doi.org/10.1016/j.polymer.2005.12.051>
- [113] Baghani M.: Analytical study on torsion of shape-memory-polymer prismatic bars with rectangular cross-sections. *International Journal of Engineering Science*, **76**, 1–11 (2014).  
<https://doi.org/10.1016/j.ijengsci.2013.11.016>

- [114] Ansari M., Golzar M., Baghani M., Abbasishirsavar M., Taghavimehr M.: Force recovery evaluation of thermo-induced shape-memory polymer stent: Material, process and thermo-viscoelastic characterization. *Smart Materials and Structures*, **28**, 95022 (2019).  
<https://doi.org/10.1088/1361-665X/ab28fc>
- [115] Li M., Neoh K. G., Xu L. Q., Wang R., Kang E. T., Lau T., Olszyna D. P., Chiong E.: Surface modification of silicone for biomedical applications requiring long-term antibacterial, antifouling, and hemocompatible properties. *Langmuir*, **28**, 16408–16422 (2012).  
<https://doi.org/10.1021/la303438t>
- [116] Langer R., Tirrell D. A.: Designing materials for biology and medicine. *Nature*, **428**, 487–492 (2004).  
<https://doi.org/10.1038/nature02388>