Shape memory alloys in medicine

Materiali z oblikovnim spominom v medicini

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Abstract: The Shape memmory alloys (SMA) are success story in medical applications market with enormous growth in usage. Huge advances from surgical point of view mean great opportunity for new commercial applications. This paper reviews the development of the shape memory alloys (SMA), constitutive behavior and use in medicine. Shape memory effect, pseudoelasticity and other basic properties of SMA are presented. Later many medical devices using shape memory effect and current commercial applications are presented. Because vast majority of current SMA medical devices is made from Nitinol, this paper also consider the factors that impinge on the associated risk analysis of using Nitinol in medical applications.

Izvleček: Materiali z oblikovnim spominom so zgodba o uspehu uporabe v medicini in še vedno pridobivajo na svoji uporabnosti. Velike prednosti s kirurškega stališča pomenijo ogromne možnosti za nove komercialne aplikacije. To delo podaja pregled fizikalnega ozadja materialov z oblikovnim spominom, njihove fizikalne zakonitosti ter uporabnost v medicini. Predstavljeni so pojav oblikovnega spomina, psevdoelastičnost in druge osnovne lastnosti materialov z oblikovnim spominom. V nadaljevanju so predstavljeni medicinski pripomočki, ki izkoriščajo oblikovni spomin in sedanja komercialna uporaba. Ker je večina takšnih medicinskih pripomočkov izdelanih iz materiala nitinol so preučeni tudi vplivi, ki zadevajo nevarnosti uporabe tega materiala v medicinskih aplikacijah.

Key words: shape memory effect, shape memory alloys, SMA medical implants, Nitinol

Ključne besede: oblikovni spomin, materiali z oblikovnim spominom, SMA medicinski vsadki, nitinol

Review scientific paper

INTRODUCTION

Smart materials have been given a lot of attention mainly for their innovative use in practical applications. One example of such materials is also the family of shape memory alloys (SMA) which are arguably the first well known and used smart material. Shape memory alloys possess a unique property according to which, after being deformed at one temperature, they can recover to their original shape upon being heated to a higher temperature. The effect was first discussed in the 1930s by ÖLANDER[1] and GRENINGER and MOORADIAN[2]. The basic phenomenon of the shape memory effect was widely reported a decade later by Russian metallurgist G. Kurdjumov and also by CHANG and READ[3]. However, presentation of this property to the wider public came only after the development of the nickel-titanium alloy (nitinol) by BUEHLER and WANG[4]. Since then, research activity in this field has been intense, and a number of alloys have been investigated, including Ag-Cd, Au-Cd, Cu-Zn, Cu-Zn-Al, Cu-Al-Ni, Cu-Sn, Cu-Au-Zn, Ni-Al, Ti-Ni, Ti-Ni-Cu, Ni-Ti-Nb, Ti-Pd-Ni, In-Ti, In-Cd and others. Crystallography of shape memory alloys have been studied for the last four decades. Only a fraction of the available literature is listed here[5-14]. Because these materials are relatively new, some of the engineering aspects of the material are still not well understood. Many of the typical engineering descriptors, such as young's modulus and yield strength, do not apply to shape memory alloys since they are very strongly temperature dependent. On the other hand, a new set of descriptors must be introduced, such as stress rate and amnesia. That is why numerous constitutive models have been proposed over the last 20 years to predict thermomechanical behaviour[15-28].

THERMOMECHANICAL BEHAVIOR

These materials have been shown to exhibit extremely large, recoverable strains (on the order of 10 %), and it is these properties as functions of temperature and stress which allow SMAs to be utilized in many exciting and innovative applications. From a macroscopic point of view, the mechanical behavior of SMAs can be separated into two categories: the *shape memory effect* (SME), where large residual (apparently plastic) strain can be fully recovered upon raising the temperature after loading and unloading cycle; and the *pseudoelasticity* or *superelasticity*, where a very large (apparently plastic) strain is fully recovered after loading and unloading at constant temperature. Both effects are results of a martensite phase transformation. In a stress-free state, an SMA material at high temperatures exists in the parent phase (usually a body-centered cubic crystal structure, also refered as the austenite phase). Upon decreasing the material temperature, the crystal structure undergoes a self-accommodating crystal transformation into martensite phase (usually a face-centered cubic structure). The phase change in the unstressed formation of martensite from austenite is referred to as 'self-accommodating' due to the formation of multiple martensitic variants and twins which prohibits the incurrence of a transformation strain. The martensite variants, evenly distributed throughout material, are all crystallographically equivalent,

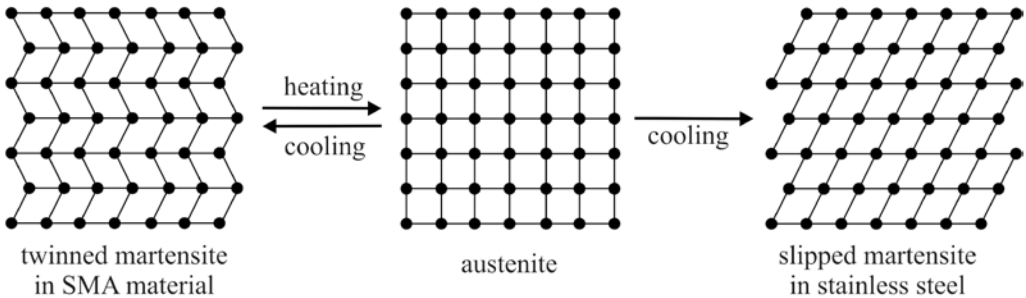

Figure 1. Martensite transformation in shape memory alloys and steels
Slika 1. Martenzitna premena v materialih z oblikovnim spominom in jeklih

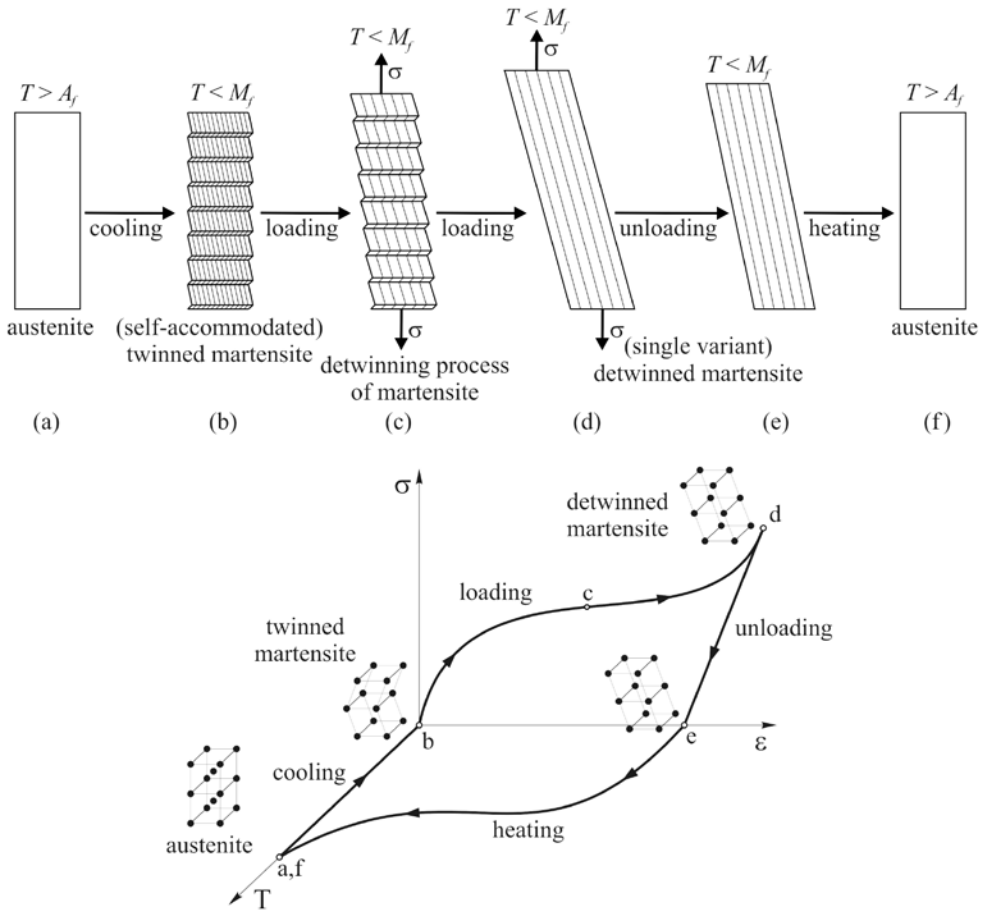

Figure 2. Shape memory effect
Slika 2. Pojav spomina oblike

differing only by habit plane. The process of self-accomodation by twinning allows an SMA material to exhibit large reversible strains with stress. However, the process of self-accommodation in ordinary materials like stainless steel does not take place by twinning but via a mechanism called slip. Since slip is a permanent or irreversible process, the shape memory effect cannot occur in these materials. The difference between the twinning and slip process is shown in Figure 1.

In the stress-free state, an SMA material has four transition temperatures, designated as M_f, M_s, A_s, A_f, i.e. Martensite Finish, Martensite Start, Austenite Start, and Austenite Finish, respectively. In the case of "Type I" materials, temperatures are arranged in the following manner: $M_f < M_s < A_s < A_f$. A change of temperature within the range $M_s < T < A_s$ induces no phase changes and both phases can coexist within $M_f < T < A_f$. With these four transformation temperatures and the concepts of self-accommodation, the shape memory effect can be adequately explained. As an example let us consider a martensite formed from the parent phase, Figure 2(a), cooled under stress-free conditions through M_s and M_f. This material has multiple variants and twins present, Figure 2(b), all crystallographically equivalent, but with different orientation (different habit plane indices). When a load applied to this material reaches a certain critical stress, the pairs of martensite twins begin "detwinning" to the stress-prefered twins, Figure 2(c). It means that the multiple martensite variants begin to convert to a single variant determined by alignment of the habit planes with the direction of loading, Figure 2(d). During this process of reorientation, the stress rises very slightly in comparison to the strain. As the single variant of martensite is thermodynamically stable at T < A_s, upon unloading there is no conversion to multiple variants and only a small elastic strain is recovered, leaving the material with a large residual strain, Figure 2(e). The detwinned martensite material can recover the entire residual strain by simply heating above A_f; the material then transforms to the parent phase, which has no variants, and recovers to its original size and shape, Figure 2(f), thus creating the shape memory effect.

The pseudoelastic effect can be explained, if an SMA material is considered to be entirely in the parent phase (with $T > A_f$), Figure 3(a). When stress is applied to this material, there is a critical stress at which the crystal phase transformation from austenite to martensite can be induced, Figure 3(b). Due to the presence of stress during the transformation, specific martensite variants will be formed preferentially and at the end of transformation, the stress-induced martensite will consist of a single variant of detwinned martensite, Figure 3(c). During unloading, a reverse transformation to austenite occurs because of the instability of martensite at $T > A_f$ in the absence of stress, Figure 3(e). This recovery of high strain values upon unloading yields a characteristic hysteresis loop, diagram in Figure 3, which is known as pseudoelasticity or superelasticity.

Many of the possible medical applications of SMA materials in the 1980's were attempting to use the thermally activated memory effect. However, temperature regions tolerated by the human body are

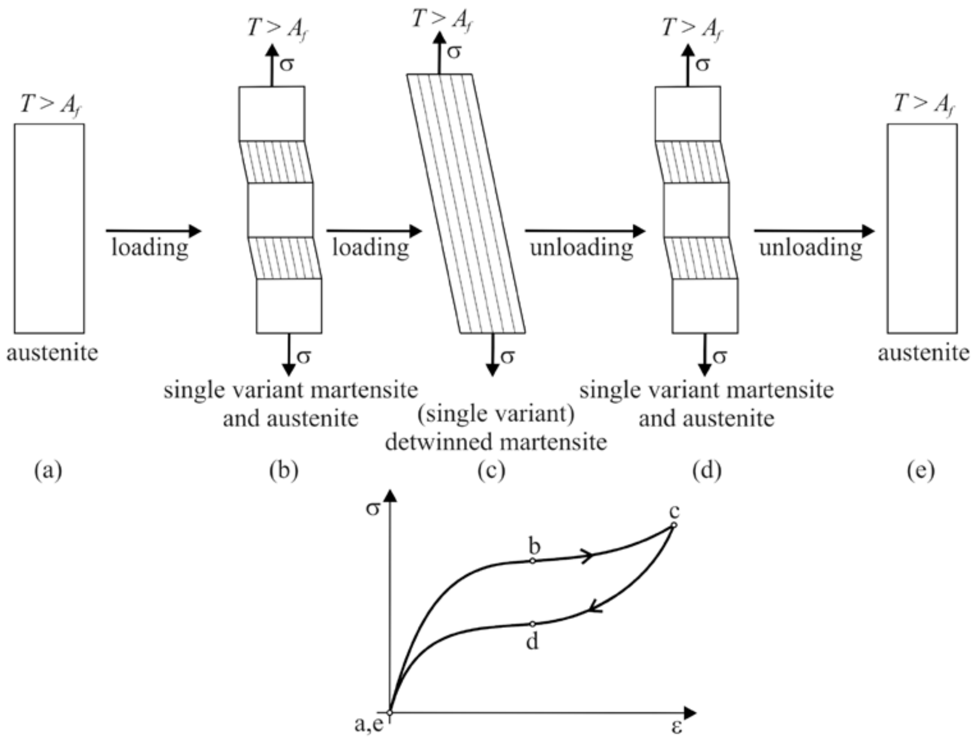

Figure 3. Pseudoelasticity or superelasticity
Slika 3. Psevdoelastičnost ali superelastičnost

very limited. Small compositional changes around the 50-50 % of Ti-Ni ratio can make dramatic changes in the operating characteristics of the alloy. Therefore very precise control of phase transition temperatures is required. On the other hand, pseudoelasticity is ideally suited to medical applications since the temperature region of optimum effect can easily be located to encompass ambient temperature through body temperature.

Biocompatibility

It is important to understand the direct effects of an individual component of the alloy since it can dissolve in the body due to corrosion and it may cause local and systemic toxicity, carcinogenic effects and immune response. The cytotoxicity of elementary nickel and titanium has been widely researched, especially in the case of nickel, which is a toxic agent and allergen[29-31]. Nickel is known to have toxic effects on soft tissue structures at high concentrations and also appears to be harmful to bone structures, but substantially less than cobalt or vanadium, which are also routinely used in implant alloys. Experiments with toxic metal salts in cell cultures have shown decreasing toxicity in the following order: Co > V > Ni > Cr > Ti > Fe [32]. The dietary exposure to nickel is 160-600 mg/

day. Fortunately most of it is eliminated in the feces, urine and sweat. Pure nickel implanted intramuscularly or inside bone has been found to cause severe local tissue irritation and necrosis and high carcinogenic and toxic potencies. Due to corrosion of medical implants, a small amount of these metal ions is also released into distant organs. Toxic poisoning is later caused by the accumulation, processing and subsequent reaction of the host to the corrosion of the Ni-containing implant. Nickel is also one of the structural components of the metalloproteins and can enter the cell via various mechanisms. Most common Ni^{2+} ions can enter the cell utilizing the divalent cation receptor or via the support with Mg^{2+}, which are both present in the plasma membrane. Nickel particles in cells can be phagocytosed, which is enhanced by their crystalline nature, negative surface energy, appropriate particle size (2-4 μm) and low solubility. Other nickel compounds formed in the body are most likely to be $NiCl_2$ and NiO, and fortunately there is only a small chance that the most toxic and carcinogenic compounds like Ni_3S_2, are to be formed. Nickel in soluble form, such as Ni^{2+} ions, enters through receptors or ion channels and binds to cytoplasmic proteins and does not accumulate in the cell nucleus at concentrations high enough to cause genetic consequences. These soluble Ni^{2+} ions and are rapidly cleaned from the body. However, the insoluble nickel particles containing phagocytotic vesicles fuse with lysosomes, followed by a decrease of phagocytic intravesicular pH, which releases Ni^{2+} ions from nickel containing carrier molecules. The formation of oxygen radicals, DNA damage and thereby inactivation of tumor suppressor genes is contributed by that.

On the other hand, titanium is recognized to be one of the most biocompatible materials due to the ability to form a stable titanium oxide layer on its surface. In an optimal situation, it is capable of excellent osteointegration with the bone and it is able to form a calcium phosphate-rich layer on its surface, Figure 4, very similar to hidroxyapatite which also prevents corrosion. Another advantageous property is that in case of damaging the protective layer the titanium oxides and calcium phosphate layer regenerate.

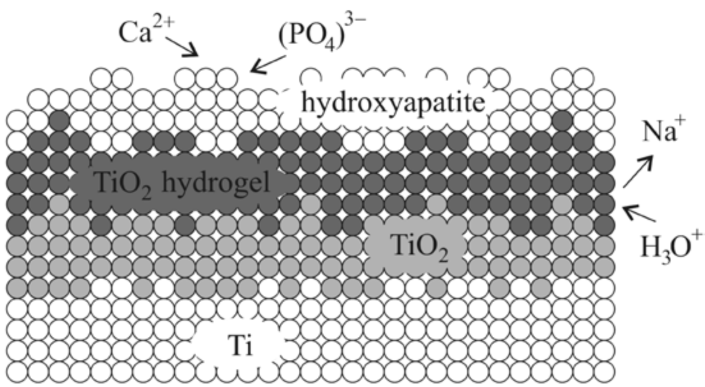

Figure 4. Formation of hydroxyapatite layer on titanium oxide film[33]
Slika 4. Nastanek hidroksiapatitne plasti na plasti titanovega oksida[33]

The properties and biocompatibility of nitinol have their own characteristics which are different from those of nickel or titanium alone. In vitro NiTi biocompatibility studies on the effects of cellular tolerance and its cytotoxicity have been performed on various cell culture models[34,35]. Human monocytes and microvascular endothelial cells were exposed to pure nickel, pure titanium, stainless steel and nitinol. Nitinol has been shown to release higher concentrations of Ni^{2+} ions in human fibroblast and osteoblast cultures, which did not affect cell growth[36-38]. Metal ion release study also revealed very low concentrations of nickel and titanium that were released from nitinol. Researchers therefore concluded that nitinol is not genotoxic.

For in vivo biocompatibility studies of nitinol effect, different experiments have been done on animals. Several in vivo nitinol biocompatibility studies which were done in the last decade disclosed no allergic reactions, no traces of alloy constituents in the surrounding tissue and no corrosion of implants. Studies of rat tibiae response to NiTi, compared with Ti-6Al-4V and AISI 316L stainless steel, showed that the number and area of bone contacts was low around NiTi implants, but the thickness of contact was equal to that of other implants. Normal new bone formation was seen in rats after 26 weeks after implantation. Good biocompatibility results of NiTi are attributed to the fact that implants are covered by a titanium oxide layer, where only small traces of nickel are being exposed.

Corrosion Behavior

The body is a complicated electrochemical system that constitutes an aggressive corrosion environment for implants which are surrounded by bodily fluids of an aerated solution containing 0.9 % NaCl, with minor amounts of other salts and organic compounds, serum ions, proteins and cells which all may modify the local corrosion effect. High acidity of certain bodily fluids is especially hostile for metallic implants. Acidity can increase locally in the area adjacent to an implant due to inflammatory response of surrounding tissues mediating hydrogen peroxide and reactive oxygen and nitrogen compounds. The local pH changes for infected tissues or near haematomas are relatively small, however these changes can alter biological processes and thereby the chemistry around the implant. It is known that small point corrosion or pitting prevails on surfaces of metallic implants. Another important feature is roughness of the surface which increases the reacting area of the implant and thereby add to total amount of corrosion. Therefore surface finishing is a major factor in improving corrosion resistance and consequently biocompatibility of medical devices[39, 40].

Corrosion resistance of SMA has also been studied in vivo on animals. Plates and stents have been implanted in dogs and sheeps for several months. Corrosion has been examined under microscope and pitting was established as predominant after the implants were removed. Thus surface treatments and coatings were introduced.

The improvement of corrosion resistance was considerable, since pitting decreased in some cases from 100 μm to only 10 μm in diameter.

Surface

The human response to implanted materials is a property closely related to the implant surface conditions. The major problems associated with the implants currently used are inadequate implant-tissue interface properties. Parameters that characterize surface property are chemical composition, crystallinity and heterogeneity, roughness and wettability or surface free energy which is a parameter important for cell adhesion. Each parameter is of great importance to biological response of the tissue. Another problem is implant sterilization which can remarkably modify desired parameters. Steam and dry sterilization are nowadays replaced by more advanced techniques like hydrogen peroxide plasma, ethylene oxide, and electron and γ-ray irradiation.

The surface of NiTi SMA has revealed a tendency towards preferential oxidation of titanium. This behavior is in agreement with the fact that the free enthalpy of formation of titanium oxides is negative and exceeds in absolute value the enthalpy of formation of nickel oxides by at least two to three times. The result of oxidation is an oxide layer of a thickness between 2-20 nm, which consists mainly of titanium oxides TiO_2, smaller amounts of elemental nickel Ni, and low concentrations of nickel oxides NiO. The surface chemistry and the amount of Ni may vary over a wide range, depending on the preparation method. The ratio of Ti/Ni on polished surface is around 5.5, while boiled or autoclaved items in water show decreased concentration of Ni on the surface and the Ti/Ni ratio increases up to 23 to 33 [41]. Different in vitro studies have shown how the physical, chemical and biocompatible properties of the implant surface can be improved[42-46].

Surface Improvements

Some of the most important techniques for improving the properties of Ni-Ti alloy surfaces are: (1) *Surface modification by using energy sources and chemical vapors* like hydroxyapatite, laser and plasma treatment, ion implantation, TiN and TiCN chemical vapor deposits. Hydroxyapatite coatings result in the best known biocompatibility and reveal a tendency to dissolution due to its relative miscibility with body fluids. Ion implantation and laser treatments usually result in surface amorphization that improves corrosion resistance, but the obtained amorphous surface layers are often not uniform. Laser surface melting leads to an increased oxide layer, decrease of Ni dissolution and improvement of the cytocompatibility up to classical Ti level. There is also a possibility that laser melted surfaces may be enriched in nickel, and become harder than bulk and swell. TiN and TiCN coatings are known to improve corrosion resistance but large deformations caused by the shape memory effect may cause cracking of the coating. Therefore, for plates and staples a plasma-polymerized tetrafluoroethylene has been introduced. (2) *Development of bioactive surfaces* is another approach to improve biocompatibility of the SMA. Human plasma fibronectin covalently immobilised to NiTi surface improved the attachment of cells while corrosion rates were reduced drastically. Studies showed NiTi surface improved with this method caused a development of calcium phosphate layers, which in fact eliminate the need for hydroxyapatite coatings[43,47]. (3) *Electrochemical processing for oxidation in air/oxygen* is a most common way of metal surface treatment. The technique combines electrochemical processes and oxidation in various media. Growth of native passive films that are highly adhesive and do not crack or break due to dynamic properties of SMA is promoted with this method. Oxide films obtained in air have different colors, thickness, and adhesive properties, with TiO_2 as a predominant oxide type. (4) *Oxidation of SMA medical devices in water and steam* is also one of the surface improvement techniques. Implants are preliminary chemically etched and boiled in water. The result is a surface with a very low Ni concentration, while etching removes surface material that was exposed to processing procedures and acquired various surface defects and heterogeneity. It also selectively removes nickel and oxidizes titanium. Surfaces obtained after oxidation in steam show better properties than those oxidized in water. (5) *Electrochemical techniques* are commonly used to passivate NiTi surfaces. Surface passivation using electropolishing is often considered as a treatment of first choice just because this technique is used for surface conditioning of stainless steels, Co-Cr alloys, etc. However, the universal techniques developed for surface passivation of various alloys used for medical purposes are not necessary efficient for NiTi.

It should also be noted that the implant surface coatings are not always beneficial. The major problem of titanium based alloys is that the formation of TiO_2, according to the chemical equation $Ti + 2H_2O \rightarrow TiO_2 + 4H^+ + 4e^-$, reduces the pH level at the titanium/coating interface. This means that if the coating is composed of hydroxyapatite, it can dissolve, which gradually leads to detachment of the coating.

Medical applications

The trends in modern medicine are to use less invasive surgery methods which are performed through small, leak tight portals into the body called trocars. Medical devices made from SMAs use a different physical approach and can pull together, dilate, constrict, push apart and have made difficult or problematic tasks in surgery quite feasible. Therefore unique properties of SMAs are utilized in a wide range of medical applications. Some of the devices used in various medical applications are listed below.

Stents are most rapidly growing cardiovascular SMA cylindrical mesh tubes which are inserted into blood vessels to maintain the inner diameter of a blood vessel. The product has been developed in response to limitations of balloon angioplasty, which resulted in repeated blockages of the vessel in the same area. Ni-Ti alloys have also become the material of choice for superelastic *self-expanding (SE) stents* which are used for a treatment of the superficial femoral artery disease, Figure 5(a). The SE nitinol stents are produced in the open state mainly with laser cut tubing and later compressed and inserted into the catheter. They can also be produced from wire and laser welded or coiled striped etched sheet. Before the compression stage, the surface of the stent is electrochemically polished and passivated to prescribed quality. Deployment of the SE stent is made with the catheter. During the operation procedure, when the catheter is in the correct position in the vessel, the SE stent is pushed out and then it expands against the inner diameter of the vessel due to a rise in temperature (thermally triggered device). This opens the iliac artery to aid in the normal flow of blood. The delivery catheter is then removed, leaving the stent within the patient's artery. Recent research has shown that implantation of a self-expanding stent provides better outcomes, for the time being, than balloon angioplasty[48-50]. *The Simon Inferior Vena Cava (IVC) filter* was the first SMA cardiovascular device. It is used for blood vessel interruption for preventing pulmonary embolism via placement in the vena cava. The Simon filter is filtering clots that travel inside bloodstream[51]. The device is made of SMA wire curved similary to an umbrella which traps the clots which are better dissolved in time by the bloodstream. For insertion, the device is exploiting the shape memory effect, i.e. the original form in the martensitic state is deformed and mounted into a catheter. When the device is released, the body's heat causes the filter to return to its predetermined shape. *The Septal Occlusion System* is indicated for use in patients with complex ventricular septal defects (VSD) of significant size to warrant closures that are considered to be at high risk for standard transatrial or transarterial surgical closure based on anatomical conditions and/or based on overall medical condition. The system consists of two primary components; a permanent implant, which is constructed of an SMA wire framework to which polyester fabric is attached, and a coaxial polyurethane catheter designed specifically to facilitate attachment, loading, delivery and deployment to the defect[52]. The implant is placed by advancing the delivery catheter through blood vessels to the site of the defect inside the heart. The implant remains in the heart and the delivery catheter is removed. Instruments for minimally invasive surgery used in endoscopic surgery could not be feasible without implementation of SMA materials. The most representative instruments such as *guidewires*, *dilatators* and *retrieval baskets* exploit good kink resistance of SMAs[53]. *Open heart stabilizers* are instruments similar to a steerable joint endoscopic camera. In order to perform baypass operations on the open heart stabilizers are used to prevent regional heart movements while performing surgery. Another employment of the unique properties of SMAs such as constant force and superelasticity in heart surgery is a *tissue spreader* used to spread fatty tissue of the heart, Figure 5(b).

In general, conventional orthopedic implants by far exceed any other SMA implant in weight or volume. They are used as fracture fixation devices, which may or may not be removed and as joint replacement devices. Bone and nitinol have similar stress-strain characteristics, which makes nitinol a perfect material for production of bone fixation plates, nails and other trauma implants[54]. In traditional trauma surgery bone plates and nails fixated with screws are used for fixation of broken bones. Shape memory fixators are one step forward applying a necessary constant force to faster fracture healing. *The SMA embracing fixator* consists of a body and sawtooth arms[55]. It embraces the bone about 2/3 of the circumference, Figure 5(c). The free ends of the arms which exceed the semi-circle are bent more medially to match the requirement fixation of a long tubular body whose cross section is not a regular circle. The applied axial compression stress is beneficial for enhancing healing and reducing segmental osteoporosis caused by a stress shielding effect. Its martensitic transformation temperature is 4-7 °C and shape recovery temperature is around the body's normal temperature, 37 °C. Similar to the embracing fixator is the so called *Swan-Like Memory-Compressive Connector (SMC)* for treatment of fracture and nonunion of upper limb diaphysis. The working principle of the device is similar with one important improvement. The SMC trauma implant is able to put constant axial stress to a fractured bone[56]. For fixation of tibial and femoral fractures nails fixated with screws are normally used. New *SMA inter-locking intramedullary nails* have many advantages compared to traditional ones. For example, when cooled SMA inter-locking nails are inserted into a cavity, guiding nails are extracted and body heat causes bending of nails into a preset shape applying constant pressure in the axial direction of the fractured bone[57]. The SMA effect is also used in surgical fixators made from wire. Certain device which have been developed to fix vertebra in spine fractures are similar to an ordinary staple. *Staple shaped compression medical devices* are also used for internal bone fixation[58]. The compression staple is one of most simple and broadly used SMA devices in medicine, Figure 5(d). Since its introduction in 1981, over a thousand patients have been all successfully treated using this device. *The SMA Patellar Concentrator* was designed to treat patellar fractures, Figure 5(e). The device exerts continuous compression for the fixation of patella fracture. The shape of patellar concentrator consists of two basic patellae claws, conjunctive waist and

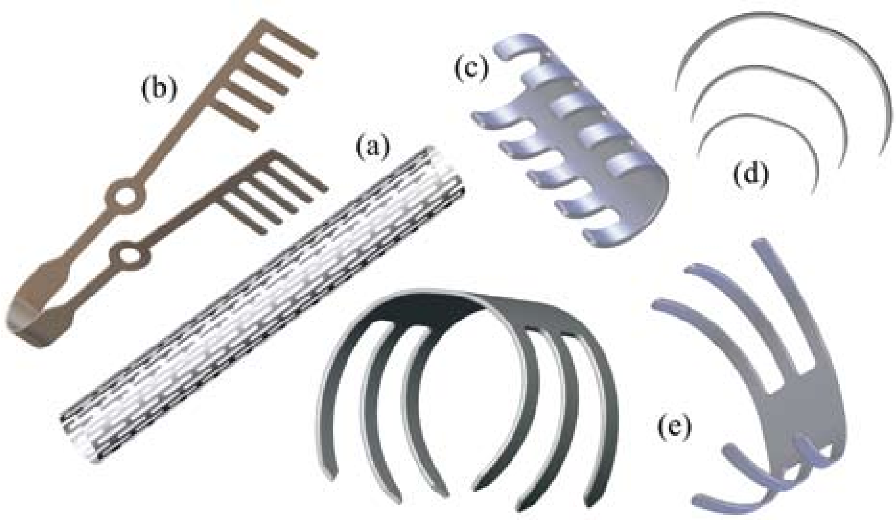

Figure 5. Examples of nitinol medical devices
Slika 5. Primeri medicinskih pripomočkov iz nitinola

three apex patellae claws. The thickness of the device may vary between 1.8 and 2.2 mm depending on different sizes of the concentrator. In clinical surgery, the claws are unfolded and put over fractured patella. Exposed to body temperature, the device tends to recover to its original state resulting in a recovery compressive force[59].

Dentists are using devices made from SMA for different purposes. NiTi based SMA material performs exceptionally at high strains in strain-controlled environments, such as exemplified with *dental drills* for root canal procedures. The advantage of these drills is that they can be bent to rather large strains and still accommodate the high cyclic rotations[52]. Superelastic SMA wires have found wide use as orthodontic wires as well, Figure 6(b). *NiTi orthodontic archwire* was first produced in batches and clinically used in China in the beginning of 1980's [60]. Due to its unique property-superelasticity, the wire exerts gentle and retentive force to teeth, which is superior to stainless steel wire. Shape memory bracelet do not require as frequent visits to the dentist as the classical ones because of their ability to self adjust. The therapeutic period is therefore cut down by 50 %.

Lately a special *fixator for mounting bridgework* has been developed, Figure 6(a). A small piece of SMA metal is notch on both sides and placed between teeth and bridgework. As the temperature rises the notched area of metal is expanded on both sides causing a permanent hold of bridgework. The tooth fixator can also be used to prevent a loose tooth from falling out.

CONCLUSIONS

A SMA implants and medical devices have been successful because they offer a possibility of performing less invasive surgeries. Nitinol wires in medical instruments are more kink resistant and have smaller diameters compared to stainless steel 316L or polymer devices. Research to develop composite materials, containing SMA which will prove cost efficient and porous SMAs which will enable the transport of body fluids through its bulk is currently underway.

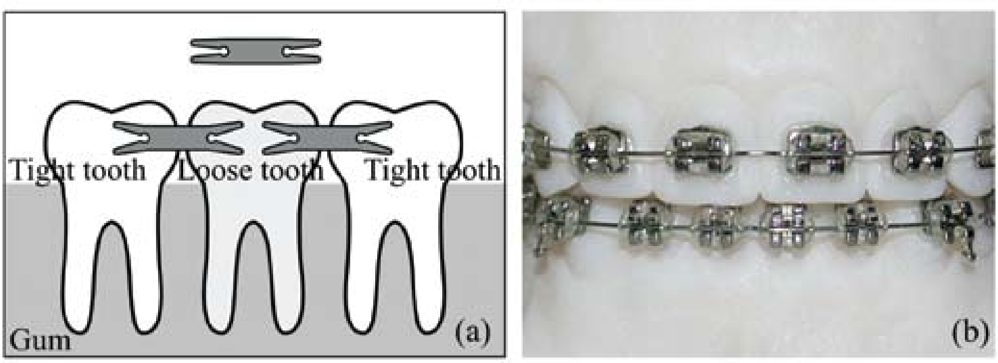

Figure 6. Dental applications of nitinol
Slika 6. Zobozdravstvene aplikacije nitinola

Povzetek

Materiali z oblikovnim spominom v medicini

Materiali z oblikovnim spominom so zgodba o uspehu uporabe v medicini in še vedno pridobivajo na svoji uporabnosti. Velike prednosti s kirurškega stališča pomenijo ogromne možnosti za nove komercialne aplikacije. To delo podaja pregled razvoja materialov z oblikovnim spominom, njihove fizikalne zakonitosti ter uporabnost v medicini. Predstavljeni so pojav oblikovnega spomina, psevdoelastičnost in druge osnovne lastnosti s spominom oblike. V nadaljevanju so predstavljeni medicinski pripomočki, ki izkoriščajo oblikovni spomin in sedanja komercialna uporaba. Ker je večina takšnih medicinskih pripočkov izdelanih iz materiala nitinol so preučeni tudi vplivi, ki zadevajo nevarnosti uporabe tega materiala v medicinskih aplikacijah.

Medicinski implantati in ostali pripomočki iz materialov z oblikovnim spominom so uspešni predvsem zaradi zmožnosti izvajanja manj invanzivnih kirurških posegov. Žice in nitinola in medicinski instrumenti so bolj odporni proti prepogibom in imajo v primerjavi z nerjavečim jeklom ali polimeri manjše prečne premere. V teku so številne raziskave, ki se ukvarjajo z razvojem in uporabnostjo kompozitnih materialov, ki v svoji sestavi vsebujejo tudi materiale s spominom oblike. Porozni materiali s spominom oblike, ki bi omogočali prehod telesnih sokov skozi prerez, pa so trenutno tik pred tem, da preidejo v uporabo za komercialne namene.

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