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Noninvasive induction implant heating: An approach for contactless altering of mechanical properties of shape memory implants

Ronny Pfeifer[a,*], Michael Hustedt[a], Volker Wesling[a], Christoph Hurschler[b], Gavin Olender[b], Martin Mach[c], Thomas Gösling[d], Christian W. Müller[d]

[a] Laser Zentrum Hannover e.V., Hollerithallee 8, 30419 Hannover, Germany
[b] Laboratory for Biomechanics and Biomaterials, Department of Orthopaedics, Hannover Medical School (MHH), Anna-von-Borries-Str. 1-7, 30625 Hannover, Germany
[c] Institute of Electrotechnology, Leibniz University Hannover, Wilhelm-Busch-Str. 4, 30167 Hannover, Germany
[d] Trauma Department, Hannover Medical School (MHH), Carl-Neuberg-Str. 1, 30625 Hannover, Germany

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ABSTRACT

This article shows an approach to change the properties of an orthopaedic shape memory implant within biological tissue, using contactless induction heating. Due to inducing the one way-memory effect, triggered by the rise of temperature within the implant, the geometry and hence the mechanical properties of the implant itself, are altered. The power uptake of the implant, depending on the induction parameters as well as on its position within the induction coil, is shown. Thermographic measurements are carried out in order to determine the surface temperature distribution of the implant. In order to simulate biological tissue, the implant was embedded in agarose gel. Suitable heating parameters, in terms of a short heating process in combination with a reduced heat impact on the surrounding environment, were determined.

1. Introduction

Apart from biological aspects, e.g. local blood supply, hormones and growth factors, the healing of bone fractures is influenced by mechanical stimuli [1]. In particular, the stiffness on the stabilizing implant, e.g. an internal osteosynthetic plate, directly affecting the local strain allocation within the fracture gap, has a big impact on the bone healing process [2]. To this date, the "ideal" stiffness of an internal implant has not yet been entirely clarified. Furthermore, the alternation of mechanical properties, e.g. stiffness, over the course of healing is still a topic of research. However, with the exception of a fixateur externe, i.e. an external fixation of the bone or biodegradable implants, today, any necessary alteration of the mechanical properties of an implant results in further surgery.

This article shows an approach to modify the stiffness of a metallic shape memory implant within the human body, using contactless induction heating. In general, this principle could offer different possibilities in regard to compression or expanding, anchoring or altering further mechanical properties of implants.

1.1. One-way shape memory effect

The shape memory effect is based on a reversible martensitic phase transformation (Fig. 1). By cooling the parent phase austenite to a critical temperature M_s (martensite start temperature), the monocrystalline structure changes into twinned martensite. This transformation is finished by reaching the martensite finish temperature (M_f). In this state, the martensite can be deformed mechanically. The maximum reversible strain (ε), depending on the structure and its thermomechanical treatment, is about 6–8% (6.7% for polycrystalline NiTi [3], 8% for nanocrystalline NiTi [4] for nickel–titanium (NiTi) shape memory alloys (SMA) [5,6]. By raising the temperature above the austenite finish temperature (A_f), the martensite completely converts into austenite again and the SMA returns to its initial predetermined state, exhibiting the so-called one-way shape memory effect (SME) [5]. It should be noted that the Young's modulus of NiTi nearly doubles when converting from martensite into austenite (cf. Table 1).

1.2. Shape memory alloys for medical applications

Since their first description in 1932, SMAs have been of increasing interest in the industrial and scientific field [5,7]. To this date, several metallic alloys offering the SME are known, whereby Cu-based (CuAlNi, CuZnAl), Fe-based and NiTi-based alloy systems are

* Corresponding author. Tel.: +49 511 2788 161; fax: +49 511 2788 100.
E-mail address: r.pfeifer@lzh.de (R. Pfeifer).

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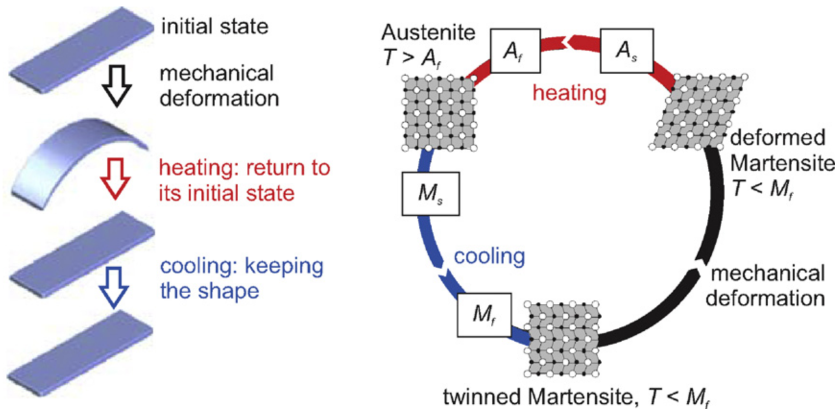

Fig. 1. Macroscopic (left) and microscopic (right) description of the one-way shape memory effect.

of commercial importance [8]. In addition, several polymers (SMP) which offer the SME, have been investigated as well throughout the last 20 years [8–11]. Furthermore, the combination of SMAs or SMPs with other materials, resulting in shape memory composites (SMC) or shape memory hybrids (SMH), offer further options to realize different phenomena and features [8].

Due to their outstanding properties, the most common SMAs today are the NiTi-SMAs. They contain a nearly equiatomic intermetallic compound of 54–60 wt% Ni, rest Ti. Compared to other SMAs, NiTi-SMAs offer the most pronounced one- and two-way SME and the pseudoplasticity, i.e. the ability to recover an apparently plastic strain on loading. Additionally, they show a high mechanical strength and a high efficiency in converting thermal into mechanical energy. Due to its resistant inert oxide layer, NiTi-SMAs are characterized by an excellent biocompatibility and corrosion resistance [12,13]. Table 1 shows typical physical properties of different SMAs and illustrates the benefits of NiTi compared to other SMAs.

Hence, NiTi-SMAs are widely used within the medical field. Today, they are applied for stents in vascular surgery and gastroenterology [14], guide wires for catheters [7], microgrippers and baskets [15], inferior vena cava filters, brackets, wires and files for orthodontic applications [16,17]. NiTi SMAs are further applied as staples for foot surgery [18] or as porous NiTi implants for intervertebral body fusion [12].

Whilst most clinical applications exploit the pseudoplasticity of NiTi-SMAs, only a few clinical applications are based on the one-way shape-memory effect, e.g. NiTi staples [5,18], such as surgical fixators in minimal access surgery (experimental) [19] or vena cava filters to treat pulmonary embolism [20]. However, several experimental studies, especially in the orthopaedic field, have been carried out in order to take account of a heat-induced shape modification. For example, NiTi SMAs have been used in order to exert a constant force on the bone after a fracture [21], for the correction of scoliosis by applying bending forces on the growing spine [22,23] and on long bones [24].

In most cases, the body temperature itself or the temperature increase due to electrical heating is used to heat the implant above A_f, hence triggering the SME. That means, the SME is initiated during or shortly after implantation. In contrast, using an “external heat source”, e.g. inductive heating, offers the possibility to induce the SME to an arbitrary date after implantation.

1.3. Inductive heating of shape memory devices

Contactless inductive heating generally can be used to increase the temperature of any conductive workpiece, triggering highly different effects when using NiTi-SMAs. So far, a few experimental applications are shown in literature. Giroux et al. used radio (f = 28 MHz) and low frequency heating (f = 8 kHz) to heat NiTi wires as an active part of a linear translator for leg lengthening [25]. External heating of stents by radio waves (f = 253 kHz) to inhibit hyperplasia, i.e. an exceeding proliferation of cells, was shown by Levitt et al. [28]. Floren et al. showed inductive stent heating (f = 200 kHz) to prevent instent restenosis [29]. The inductive heating of SMPs (f = 12.2 MHz) was shown by Buckley et al. by the experimental use of a shape-memory polymer for an endovascular thrombectomy device for stroke treatment and an expandable

Table 1
Physical and mechanical properties of NiTi-SMA [25–27]. Several parameters are depicted for both states of the SMA, the martensitic (M) and the austenitic state (A).

Properties	Unit	NiTi	CuZnAl	CuAlNi
Fusion point	°C	1250	1050	1020
Density	g/cm³	6.4–6.5	7.5–8.0	7.1–7.2
Thermal conductivity at 20 °C (A; M)	W/m K	8.6; 18	84; 120	30; 75
Specific heat	J/kg K	490	440	390
Dilatation coefficient (A; M)	μm/K	11; 6.7	17; –	18; –
Young's modulus (A; M)	GPa	70–83; 23–41	80–100	70–100
Yield strength	MPa	100–130	70	40
Tensile strength	MPa	875	800	700
Max. reversible strain (one-way m. effect) N = 1	%	6–8	4–6	4–6
Max. reversible strain (two-way m. effect) N = 1	%	3.2	1	0.8
Hysteresis	K	2–50	5–20	20–40
Electric resistivity at 20 °C (A; M)	μΩ m	0.5; 1.1	0.07; 0.12	0.1; 0.14
Magnetic susceptibility		3×10^6	2.9	2.9
Relative permeability		1.002	1.002	1.001
Corrosion resistance		++	+	o
Biocompatibility		++	–	–

The “o” means average, given the following ranking: very good: ++, good: +, average: o, poor: –, very poor: – –.