SMA AS MEANS FOR SENSITIVE BIOMEDICAL SENSORS

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Abstract

Shape memory alloys (SMA) are materials with specific electric, mechanic and thermal qualities originating in their microscopic structure and as such are used to design so-called Smart structures. In biomedicine, they are used mostly for their shape memory properties during intravasal surgery. The commonly used SMA material for biomedical purposes is Nitinol, a nickel-titanium alloy that exhibits, especially when annealed, among other properties also superelasticity and high resistivity with strain changes. This makes it suitable for usage in strain gauge sensors. In this work, NiTi annealed wires were used to create a strain gauge sensor, which exploits their unique properties. The properties of these sensors were then described using a climatic chamber, stretching device and a set of weights. Those properties were then compared to similar sensors, that are using different principles. The development, that followed, used these sensors in particular biomedical applications.

Keywords

shape memory alloy, NiTi, NiTi strain gauge, Belt sensor, Belt strain gauge

Introduction

In biomedicine, there is always a need for better sensors. Some of the signals, especially mechanical signals sensed on top of the human body, can be very weak. These signals include, but are not restricted to: arterial pulse, breathing, bowel movements, tremor, position in bed and others. Mechanical signal taken from body surface will likely be several signals superposed over each other. For proper filtration and isolation of desired signal, better resolution (both in time and value) is also desired. Strain gauges are devices for sensing value and changes in force (pressure), and many different technologies, setups and materials is used in them. Usually utilizing change of resistance with elongation.

We have decided to develop a force/pressure sensor using SMA (shape memory alloy) material, namely nickel-titanium alloy called Nitinol, which is widely used in biomedicine for intravascular applications due to its shape memory properties. The reason for this is the prior attempts to create sensors with these materials here at the university. These previous attempts resulted in a development of a Belt technology for sensor construction. This technology and unique material properties of NiTi wires allowed us to develop NiTi strain gauges. We then ran measurements to describe the electrical and mechanical properties of the sensor and the material itself.

In this paper, we present the general properties of NiTi wire, the Belt technology used to construct sensors, our own measurements and results and lastly some of our concurrent research and applications of such sensors.

Strain gauges are used in biomedicine mostly for coarse measurements, for example as limiting force sensor for mammography, robotic arms, weighing and lifting systems, assistive technologies and else. They are also used as part of mechanical machines like syringe pumps or dialysis machines. Their use for diagnosis and biomechanical signals sensing is very limited from obvious reasons [1]. For once the gauge usually has a particular position and range of values, for the results to be comparable. Secondly the gauges are usually constructed for larger strains and not very suitable for subtle mechanical signals. This results in biomechanical signals taken by other physical method (for example optically in case of arterial pulse or piezoelectric in case of baby monitor pads) or taken during physical examination by physician (auscultation).

We believe, that using an SMA, we can construct much more sensitive strain gauge, which would be able to sense certain biomechanical signal from the body surface. The measurements described in this paper was the first step in strain gauge for biomedical application development. We then used presented gauges in

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advanced patient bed monitoring application (patient position, continuous vital function monitoring and else) and will proceed in different applications.

Materials and methods

Nickel-Titanium wire

Shape memory alloys are metals that are able to change their inner crystal structure depending on the temperature or strain. The base for unique properties is the martensitic transformation, which allows the material to undergo diffusionless phase transformation between martensite and austenite phase. The austenite is generally a high temperature phase of metals characterized by cubic crystal structure. During cooling the crystalline structure slightly deviates and changes to tetragonal, orthorhombic, monoclinic or other structures associated with martensite. The displacement of particular atoms in the lattice is lesser than atomic distances and therefore, the phase transformation is isovolumetric (the change is negligible, below 0.1%) [2]. In shape memory alloys, this change is reversible.

In equiatomic Nickel-Titanium alloy, used also in the medicine [3], the austenite and martensite phases are both stable at low temperatures, as the atoms have not enough energy to move due to diffusion. When the material is influenced by temperature or strain stimulus, martensite or austenite (depending whether the material is cooled, heated or deformed) nuclei are formed in the material and spread in the volume. This phase change during stress or heat shows a hysteresis because of inner friction forces [4, 5]. Due to this change, several unique properties can be observed in these materials.

First is shape memory, that has given these materials their name, which is related to thermal-induced phase transformation. The wire is shaped in desired manner when in austenite form. The material is then cooled, which means that inner structure transforms into one of many types of martensite (the lattice can take up to 24 different shapes) [2], and due to isovolumetric transformation, and without any force acting on it, the shape stays the same (it is called self-accommodation). The wire can then be plastically deformed in the martensite phase, but when heated, it takes the original shape of the austenite phase. That is possible thanks to twinning, which is a change of orientation of lattice without slipping planes, therefore reversible. This temperature can be set to be close to 30 °C, which makes it ideal for application in the body, where it is used for intravascular stents, delivered through a catheter, which unfolds itself in the artery under the body temperature [3].

The second is superelasticity, which is related to the stress induced-phase transformation. The adjustment to the stress is provided by lattice changes, and therefore allows for up to 10% elastic deformation (could be higher in some alloys). This significant deformation is possible thanks to the plateau part of the stress/strain curve which appears in the hysteresis area [3].

Other properties, that shape memory alloys manifest, are for example pseudoplasticity, which ties to the shape memory, where the wire is deformed to other than original shape, to which it transforms back when heated; or ability to absorb mechanical impulse, or the ability to do work [2].

Given all the described properties, NiTi wire seems like a suitable material for sensor construction for its deformation range and changes in inner structure resulting in significant resistance changes. Just like other materials with higher resistance changes, it strongly depends on temperature, so means of measurement error suppression must be utilized, e.g. bridge circuit or constant temperature measurement.

The wire is mostly manufactured from ingots and then drawn. For medical purposes it is often produced using powder metallurgy, which results in more porous material, which has better biocompatibility properties. The wire can be then annealed with heat (in a furnace, by joule heating etc.) which results in even higher resistance change during deformations [2, 5].

Belt sensor technology

Belt is a technology of assemblement for strain-gauge, where the measurement wire changes its resistance because of the deformation of the entire sensor [6]. It consists of two elastic belts and a wire in between, so that the sensor is long and thin (Fig. 1).

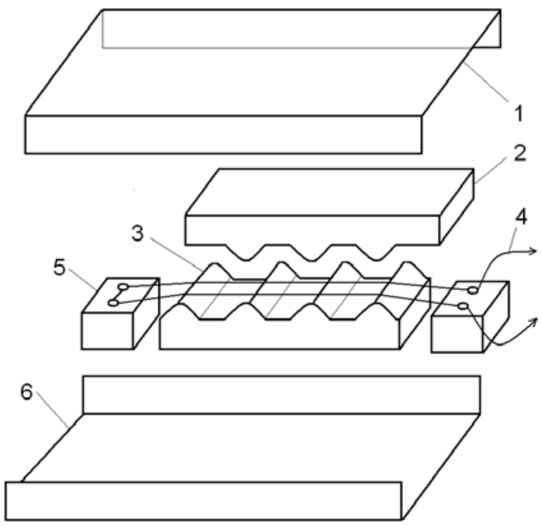

Fig. 1: Schematic of Belt sensor working principle; 1 – metal housing, 2 – elastic belt, 3 – protrusions, 4 – signal wires, 5 – NiTi wire loop. Adapted from [6] with a permission (common author).

The deformation is emphasized by special construction, where the measuring wire is between two longitudinal belts with alternating protrusions. Each protrusion faces a gap on the other belt and vice versa.

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The wire is then pinched with the protrusions against the other belt and this results in its further deformation (Fig. 2). Unlike ordinary strain-gauge, where the deformation is emphasized by using multiple loops of the wire in the axis of deformation, in this arrangement the deformation is significantly amplified even in a single loop. So much so, that the wire must be superelastic, to withstand the deformation without degrading.

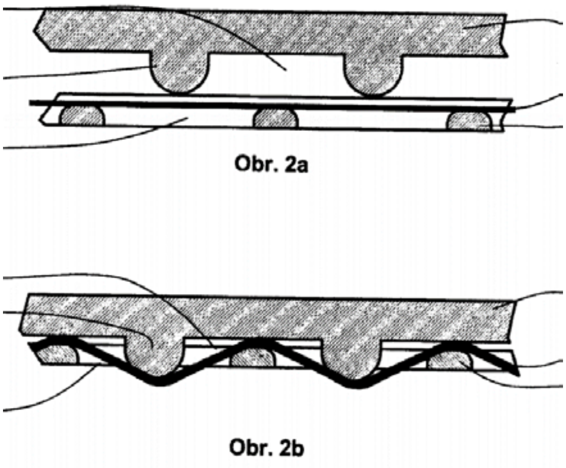

Fig. 2: Belt sensor during loading Adapted from [6] with a permission (common author).

The sensor is also usually equipped with a block, restricting the deformation from damaging the wire, when the forces acting on the sensors, for example weight of a human body, are large. These sensors can be made in all shapes and sizes and are suitable for usages, where the deformation would be too little for an ordinary strain gauge and signal had to be amplified significantly. The sensor can be for some applications further reinforced with a metal housing, which limits maximum deformation.

Measurements

Measurement layout

The goal of the measurement was to describe the physical properties of the wire, that we would then use in biomedical sensors. Much of the qualities of a NiTi wire is known, but as we developed a strain gauge sensor, we performed our own measurements, with focus on electrical properties, to develop resistance measurement setup as well as to verify annealed wire properties.

We performed several measurements on the wire itself, firstly measurement of the resistance depending on elongation, then resistance depending on the temperature, elongation and resistance depending on force and then we performed a measurement on NiTi based belt strain gauge, where we explored resistance dependency on the pressure.

The wire we used was a thin 0.05 mm diameter NiTi annealed wire, with expected resistivity for room temperature around $75–85 \cdot 10^{-8}\ \Omega \cdot \mathrm{m}$ [4]. For measurements we used a special mechanism (Fig. 3), which allowed us to stretch the wire to a defined elongation and simultaneously measure the force and resistance. The resistance was measured using wheatstone bridge, where the resistance of unknown resistor (sensor) in one of the branches is derived as:

$$R_x = \frac{\left(\frac{R}{2R} - \frac{U_x}{U}\right) R_2}{1 - \frac{R}{2R} + \frac{U_x}{U}}, \qquad (1)$$

where R is a resistance of two serial resistors in each branch (we used two 120 Ω ceramic resistors), R_2 is a resistance of a resistor (62 Ω) opposite to a measured one (a wire or a strain gauge), U is a supplied voltage (5 V) and U_x is a measured voltage between branches. The resistors were chosen so that the voltage is close to zero (when balanced), which allowed us to utilize the full resolution of the 16 bit A/D converter we used (on the range 0–1 V). The A/D conversion as well as voltage supply was done using NI USB 6211 module along with a program in NI LabView, where the data were also processed. Using our own A/D converter over premade ohmmeter device, allowed us to adjust the measurement for our needs, collect and process data, and prepare simultaneously the software for the chosen application (patient position in bed). The values of resistivity provided by our setup was verified with ohmmeter on different resistor sizes as well as the gauges themselves. Both this setup and the program was integral part of the following biomedical application (the measurements served also to test both setup and the software). Also, the A/D converter serves as the voltage source, and for its very limited current supply, the influence of self-heating is limited. The measurements were performed repeatedly over a longer time, which also limits this influence.

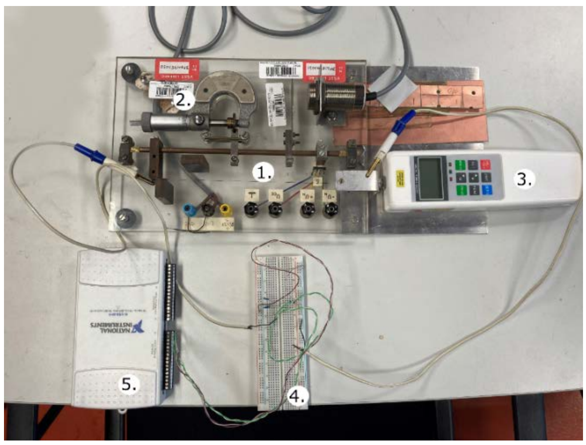

Fig. 3: Measurement setup; 1 – NiTi wire, 2 – defined elongation mechanism, 3 – force sensor, 4 – wheatstone bridge, 5 – A/D converter. Source: author.

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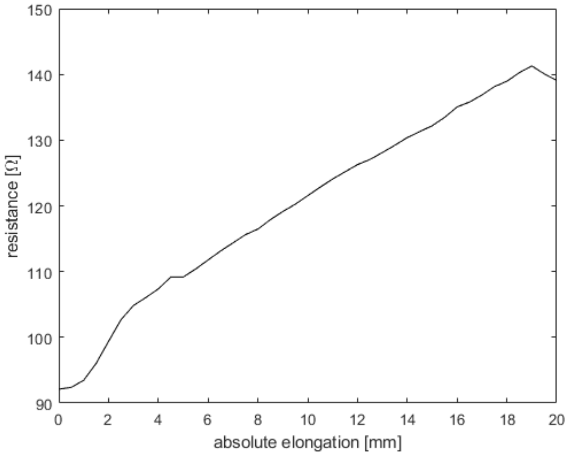

Fig. 4: Resistance on elongation dependence, the original wire length was 210 mm.

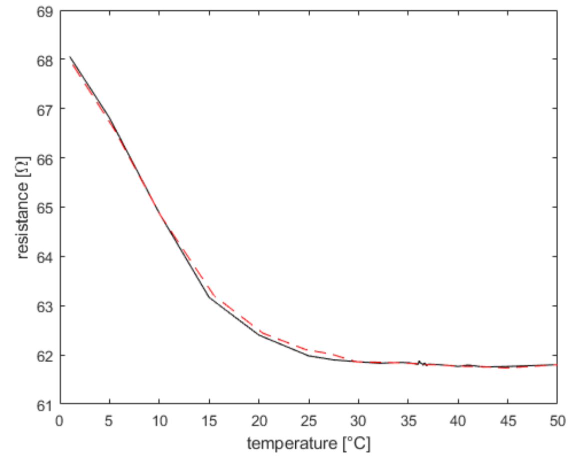

Fig. 5: Resistance on temperature dependency, measured on the gauge without loading.

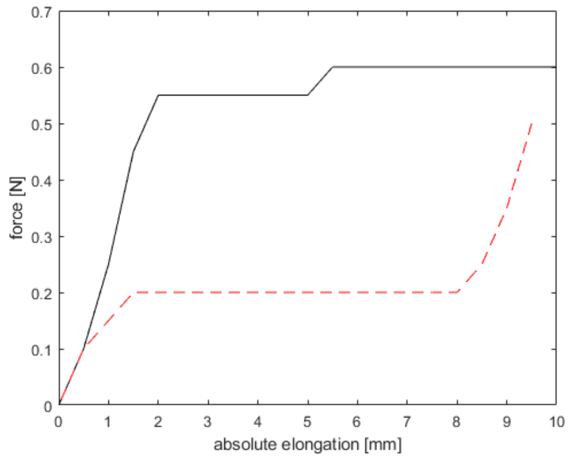

Fig. 6: Force dependency on elongation, large ability of a wire to deform is obvious, which causes the hysteresis part of the curve.

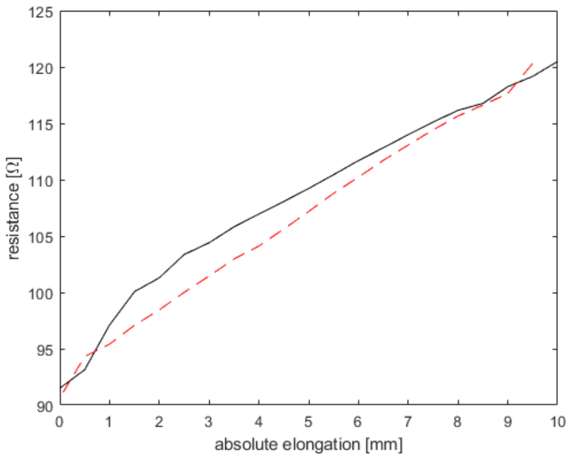

Fig. 7: Resistance on elongation dependency, with hysteresis.

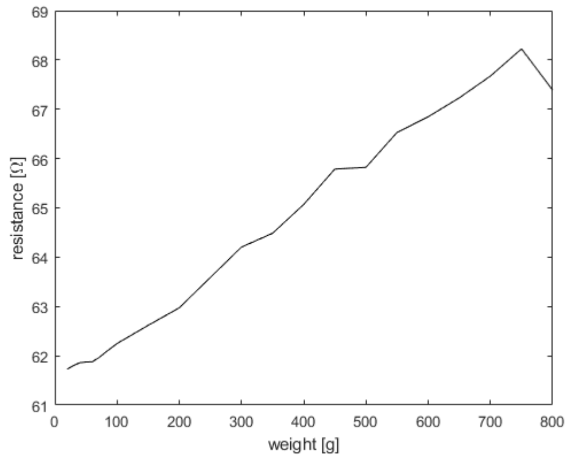

Fig. 8: Strain gauge measurement.

Figures 4 to 8 containing all the measured data. Measured values are the averages of five independent measurements during one course of measurement.

Plots description – black line is the loading curve, red dashed line is the release curve in case of hysteresis.

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Resistance depending on elongation

The first measurement was a resistance dependency on the elongation. We used 210 mm wire stretched up to 230 mm, which corresponds to NiTi tolerance to deformation up to 10% of length. The final values are an average of five independent measurements to eliminate human error and/or influence of possible hysteresis.

Usually as the metal wire deforms, its thickness decreases and length increases, which, along with minor lattice elastic changes in the material, increases the resistance, because that grows both with length and smaller area. In NiTi wire, the situation is a little different, because thin wire is used, the thickness of the wire has, considering a poisson ratio of 0.3 [4], negligible effect. The resistance change in NiTi is much larger than in metals, meaning there is a big change in the material itself. This resistivity change can be explained by phase transformation, where deformed wire acts as a serial-parallel combination of little resistors, nuclei of austenite and martensite.

For the reasons above, the resistivity itself grows with deformation. Due to complicated nature of the material changes, it is reasonable to describe it with linear coefficient, if possible. The resistivity derived from data (Fig. 4), with account to poisson ratio, in fact show some linear dependence. We can therefore describe it with a linear relation:

$$\rho = \beta\varepsilon + \rho_0, \tag{2}$$

with ρ as resistivity at certain deformation ε, with β being a coefficient factor from the slope. This effect does affect a sensor function as further increase of resistance during deformation, so it does not have to be considered other than a higher resistance increase with deformation, which only adds to sensor sensitivity. Base resistivity without deformation corresponds with that from the literature [7], and is $86.113\cdot10^{-8}$ Ω·m at room temperature.

Resistance depending on temperature

Like in other sensitive materials (for example semiconductors), the resistance changes significantly with temperature. For NiTi in particular the dependency is very nonlinear with hills for some temperature ranges and altogether quite complicated [7]. In biomedicine though, we are working either in contact with the body or at room temperature, where especially body temperature is a very stable environment with only several degrees differences. Even with this in mind, we performed a measurement in the desired temperature range to verify that the influence of temperature will be negligible. For this reason we measured a strain gauge, now acting as a wire piece of defined length, because there was no force acting on it, in temperatures from 0 to 50 °C with detailed steps around body temperature. The nominal resistance of the gauge at room temperature was 62.5 Ω.

The measurement has shown slight linear dependency (Fig. 5) around the body temperature.

Elongation and force

During the deformation, there are significant changes in inner structure. The lattice phase changes are related to minimal energy. This means that the change in lattice dissipates some of the energy. For that reason we suspected certain hysteresis between acting force and deformation. We measured the force needed for certain deformation along with current resistance (measurement similar to the first one), the force was measured with an additional force meter. First loading characteristics was measured and then, independently, release characteristics.

The data showed significant hysteresis in force to deformation relation and slight hysteresis even in resistance to deformation relation (Fig. 6 and 7). This was expected and given the small force required to return back to original shape (the force drops right at start of releasing) does not have a significant impact on the sensor measurement.

Strain gauge measurement

Last measurement we performed was a pressure on a strain gauge that we constructed using belt technology and a single NiTi wire loop. This was done by a set of weights that was placed on the sensor with a plate of defined area $S=176\cdot10^{-5}$ m². We then explored how the resistance will change with adding weight (Fig. 8).

This strain gauge is a primary outcome of this development.

Results

During the measurements we were able to describe physical and electrical properties of NiTi wire we used to construct a strain gauge using Belt technology. The used NiTi wire has a resistance of $86.113\cdot10^{-8}$ Ω·m at room temperature. Nominal resistance of a sensor is 62.5 Ω. For biomedical application the resistance dependence on temperature is almost negligible. The results of measurements were compared to literature and even consulted with the preceding papers authors [4, 5].

For strain gauge construction the belt technology is very beneficial, because it allows to make use of a superelasticity property of NiTi wire, which amplifies the deformation caused by sensor bending. The gauge factor K, derived as:

$$K = \frac{\frac{\Delta R}{R}}{\frac{\Delta L}{L}}, \tag{3}$$

(relative change in resistance divided by relative elongation) serves as a basic strain gauge parameter. For

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our sensor K = 5.2. In metal strain gauges, the milistrain changes cause resistance changes in similar order. The main advantage of NiTi strain gauge is a several order of magnitude higher resistance change, with a similar deformation of gauge itself, compared to metal strain gauges. ΔR for 20 mm deformation of the wire was 45 Ω. This corresponds with measurement using weight on the strain gauge, where even a slight change in weight with a step as small as 10 grams changes resistance of the sensor by 0.1 Ω. During the force-resistance-deformation measurement a 2 Ω change in resistance can be achieved with force as small as 0.1 N. All this applies to a single loop wire.

Discussion

We have developed a sensor using specific design, utilizing unique SMA superelastic and resistive properties. We are aware of the apparent simplicity of the measurements as our focus was more on the Belt strain gauge development and thus verifying the values from the literature, where the properties are described in detail, and would render our more detailed measurements obsolete.

From our internal measurements we know, that the sensor is sensitive enough to measure breathing frequency through the usual hospital bed mattress or to measure artery pulse wave on the body surface at the palpation site. Given that, we tried several novel applications, namely monitoring the position of a patient in bed, using matrix of sensors under mattress, with the possibility to also monitor breath; or monitoring the status of blood circuit using a gauge on the body surface.

This sensor is, unlike for example piezoelectric sensor, suitable for both static and dynamic loading measurements. For a particular application, the strain gauge will have to be structurally adjusted, especially the housing, which limits the measurable forces but protects the gauge from force over limit damage. Even during our measurements, we approached the limit for reversible (elastic) deformation. From this raises the first downside, which is artificially introduced upper limit, caused by housing, so that the wire doesn't reach irreversible deformation. Other disadvantage is change in resistance with temperature, but for given applications (sensor in contact with a body) the temperature is stable.

We want to emphasize, that all the measurements and sensitivity evaluation were done on a single loop of wire, so that the sensitivity can be always increased with more loops, as is usually done in strain gauges.

Strain gauges are used in vast applications in biomedical practice, the proposed Belt sensor can then replace those, where more sensitivity or other advantages can be utilized. The sensor can be easily constructed elastic (we had one made with 3D printing) and is therefore suitable for different pads, for example new-born breath monitoring, or different kinds of wearables.

Conclusion

The results lead us to conclude that NiTi wires can be used to construct more sensitive pressure and force sensors, and these new sensors are probably suitable for biomedical applications, which we proposed and/or tried. As mentioned in the beginning, these kinds of sensor (belt technology) were worked on before, meaning our development is more focused on adjusting these sensors for the biomedical applications. We did construct the strain gauge suitable for biomedical application, performed the described measurements, and continue working on particular applications.

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