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Magnetoelastic bilayer concept for skin curvature sensor

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Introduction

The present paper describes the magnetoelastic bilayer concept as a basic element for a novel skin curvature sensor. The dynamic changes of the curvature c of the human skin reflect a variety of physiological parameters, the composition of which depends on the skin region. Here is a large number of skin regions being ”informative”. For example, dynamic changes of c of the skin over the carotid artery show cardiac component due to volume pulsation of the carotid artery and respiratory component due to circumference changes of the neck. An other ”informative” skin region is located around the eyes [1]. Here c reflects movement of the eyes as being important for diagnosis of sleep disturbances.

In opposite to the above advantages in measuring of c, no direct sensors for assessment of c are currently available. In practice, the curvature of a cantilever, as an example shown in Fig. 1a, is measured by a strain gauge which is affixed to the cantilever. The strain gauge detects strain which is directly proportional to the internal stress σ. However, σ is strongly influenced by both geometry of the cantilever and its c. The cross-sensitivity to the geometry is given by arising tensile stress in the convex half of the cantilever. This stress is transmitted to the strain gauge through fixation and is strongly dependant on the geometry of the cantilever.

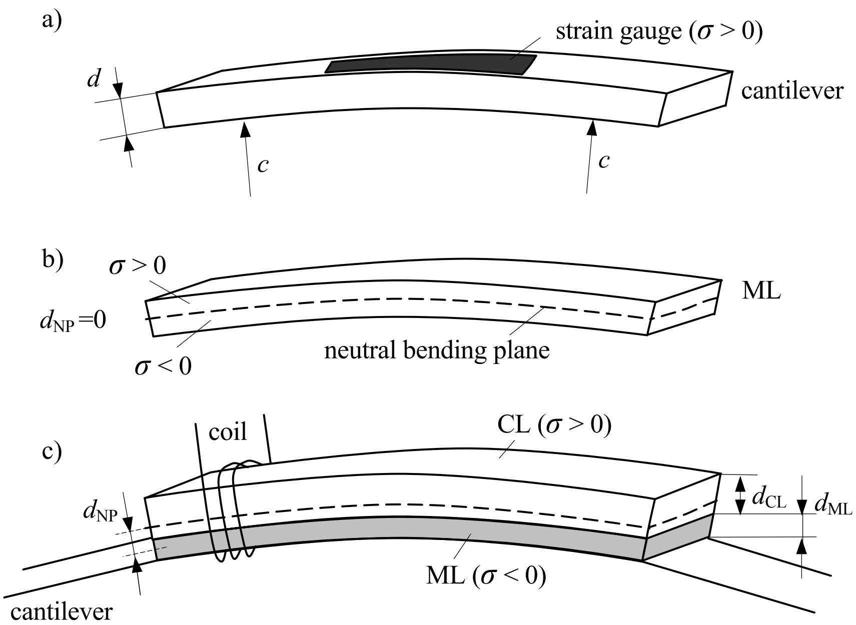

Fig. 1. **The measurement of curvature c applying (a) strain gauge on a cantilever with the thickness d and measuring the resulting stress σ (b) single magnetoelastic layer and (c) bilayer consisting of a magnetostrictive layer (ML) affixed to a counter layer (CL). The coil is used for sensor signal pick-up**

As a result, σ strongly depends on the thickness d of the cantilever, σ being proportional to d for given c (Fig. 1a). Thus, the strain gauges can be calibrated only for a given geometry of the cantilever in order to measure c. The resulting cross-sensitivity to the geometry of the object complicates the application of the strain gauges for measurements of c yielding a need of a direct curvature sensor.

Bilayer concept

The introduced magnetoelastic bilayer concept is based on the magnetoelastic effect which is well known to arise in the case of magnetostrictive soft magnetic materials. According to the positive or negative sign of the magnetostriction constant, a tensile stress (σ>0) yields an increase or a decrease of the permeability μ. This offers the most simple arrangement of a force or mass sensor using a single magnetoelastic layer [2]. For establishment of sensor signal a coil is used which surrounds the magnetostrictive material.

However, a single magnetoelastic layer (ML) is not sufficient for the curvature sensor. As shown in Fig. 1b, the bending of the single ML yields compression ($\sigma < 0$) in the concave half and tension ($\sigma > 0$) in the convex half. The neutral bending plane ($\sigma = 0$) remains in the centre of ML. Corresponding changes of μ of the latter regions partly compensate each other yielding a relatively small sensitivity to bending.

The magnetoelastic bilayer (Fig. 1c), as described in [3,4], consists of ML that is affixed to a nonmagnetic counter layer (CL). Bending the bilayer causes ML to experience in its whole volume either tensile or compressive stress, depending on the direction of bending. The resulting large changes of μ can be measured using a coil which surrounds the bilayer, as shown in Fig. 1c. The curvature sensor signal is given by the coil inductivity.

The curvature sensor is not affixed to the cantilever but tightly pressed, so that the axial motion of the bilayer is possible. The arising tensile stress in the convex half of the cantilever is not transmitted to the bilayer. As a result, the curvature sensor signal does not depend on the geometry of the cantilever, in opposite to the strain gauge (Fig. 1a).

In the case of the bilayer, the neutral bending plane is shifted from the centre of ML towards CL. As given in the case of Fig. 1c, mainly compressive stresses arise throughout ML. Assuming that ML and CL show the same width, identical Young's modulus (E_{ML}=E_{CL}) and the same thickness (d_{ML}=d_{CL}), the neutral bending plane will be shifted to the border area between ML and CL. The sensitivity to bending depends strongly on the distance d_{NP} of the neutral bending plane from the centre of ML (Fig. 1c). In the case of the single ML $d_{NP} = 0$. For small values of c and long bilayers the distance d_{NP} is given by

$$d_{NP} = \frac{(d_{ML} + d_{CL})}{2} \cdot \frac{1}{1 + \frac{E_{ML} \cdot d_{ML}}{E_{CL} \cdot d_{CL}}}. \tag{1}$$

As shown in the above equation, a high attractiveness of the bilayer approach results from the possibility to adjust the sensitivity in a simple way through the quantities E_{ML}, E_{CL}, d_{ML} and d_{CL}. In principle it can be arbitrary increased by increasing the ratio E_{CL}/E_{ML} and/or d_{CL}/d_{ML}.

The most crucial demand for the bilayer is to establish an ideal mechanical connection between ML and CL. A large variety of methods can be applied to fix the both layers, the most obvious is given by agglutination. As shown in Fig. 2, the agglutination introduces the third layer which influences d_{NP} and thus the sensitivity of the sensor.

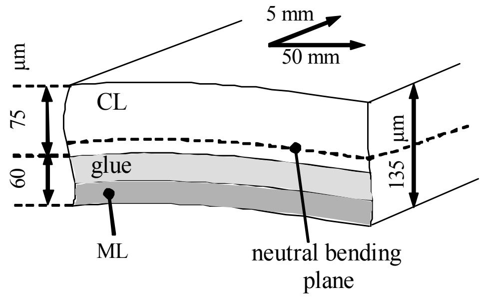

Fig. 2. **Composition and dimensions of the applied bilayer including the magnetostrictive layer (ML), the counter layer (CL) and the agglutination layer**

Furthermore, the stiffness of the bilayer is an important issue, especially if applied on the skin for its curvature registration. In this case a low stiffness and a low bending force are the prerequisites in order to avoid skin irritations. On the other side, the sensitivity to bending increases with increasing ratio E_{CL}/E_{ML} and/or d_{CL}/d_{ML} which means that the sensitivity increases with increasing stiffness. For applications on the skin a compromise should be made between low stiffness and high sensitivity.

Technical results

The curvature sensor prototypes were prepared using 40-50mm long and 5 mm wide bilayers exhibiting 25 µm ML thickness (Fe-based amorphous ribbon, VAC7505, E_{ML}=150GPa) in connection with either non-magnetic steel CL (Young's modulus 180GPa, thickness 45 µm) or aluminium CL (71GPa, 75 µm). The layers were connected by means of an agglutination layer (1GPa, 30 µm).

The sensitivity to bending is demonstrated in Fig. 3 for the case of the single ML and the bilayer. As shown in Fig. 1c, the bilayer was bent in order to induce compressive stresses ($\sigma < 0$) in ML. The single ML shows the lowest sensitivity of about 2 % of μ change per 1/m curvature. The sensitivity of about 9 % is given by the bilayer exhibiting aluminium as CL while non-magnetic steel as CL doubles the sensitivity to about 17 %. The latter increase of sensitivity is due to increased ratio E_{CL}/E_{ML} in the case of the non-magnetic steel. These results are in full agreement with Eq. 1.

As shown in Fig. 3, the single ML shows a finite value of the sensitivity in spite of the mutual compensation of σ in the stressed regions. The explanation is given in Fig. 4, showing a highly non-linear relationship between relative permeability μ_R of ML ($\mu = \mu_R \cdot \mu_0$ with permeability μ_0 of free space) and σ for $c = 0$. The tensile stress ($\sigma > 0$) induces lower $\Delta\mu$ compared to compression ($\sigma < 0$).

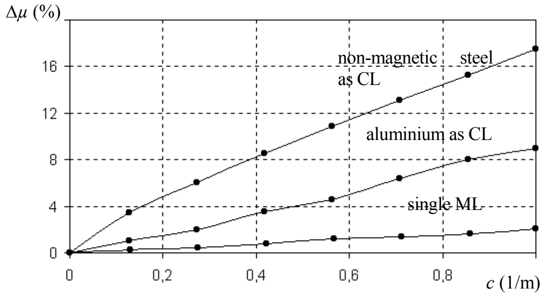

Fig. 3. **Sensitivity of the curvature sensor by measuring changes $\Delta\mu$ of the permeability μ over the curvature c. Data are given for a single magnetostrictive layer and the bilayer using non-magnetic steel or aluminium as counter layer (CL)**

If we apply this behaviour to the bend single ML, it follows that the compressed volume of ML contributes significantly more to $\Delta\mu$ than the stretched volume. The herewith resulting imbalance of the contributions yields a finite value of the sensitivity to bending.

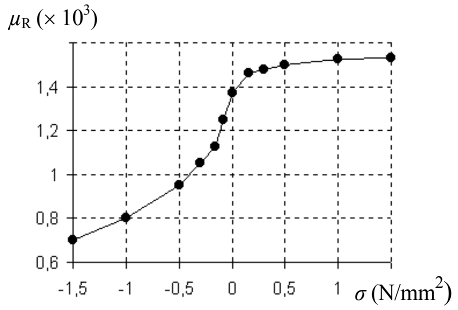

Fig. 4. **Values of relative permeability μ_R versus applied stress σ for the magnetostrictive layer**

Fig. 5 demonstrates the influence of the agglutination layer on the distribution of σ within ML using finite element simulation. The grey scale indicates the value of σ, white regions corresponding to the tensile stress, dark regions to the compressive stress. Fig. 5a shows the single ML yielding tension in the convex half and compression in the concave half. In opposite to the single layer establishment, ML within the bilayer - as shown in Fig. 5b - yields mainly compressive stress within its whole volume.

Fig. 5c demonstrates that the agglutination layer reduces the compressive stress within ML because of the relatively low Young's modulus of the agglutination layer and relatively large thickness compared to d_{CL} or d_{ML}. The reduction of the compressive stress corresponds to the decrease of the sensitivity to bending.

Application results

Fig. 6a shows the curvature sensor attached on the neck over the carotid artery. As shown in Fig. 6b, the bilayer with the coil was compactly embedded into a thin plastic envelope in order to guarantee a loose mechanical connection between BL and the skin (see section "Introduction"). The envelope ensures free motion of the bilayer in its axial direction but not in the normal. A

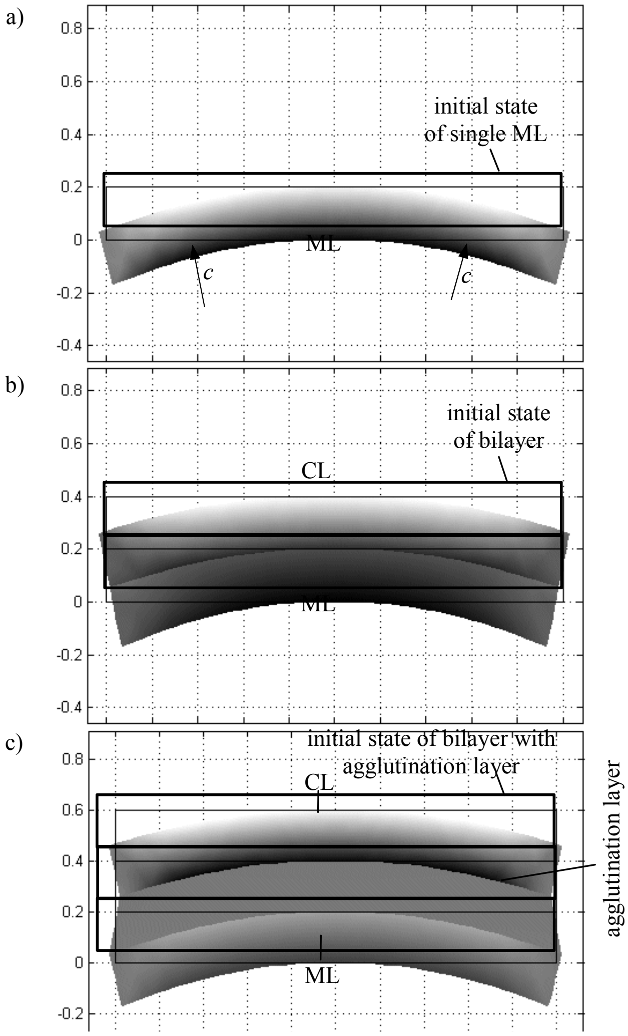

Fig. 5. **Stress distribution for curvature c = 1 (1/m) within (a) the single magnetoelastic layer (ML), (b) the bilayer without agglutination layer between ML and counter layer (CL) and (c) the bilayer with an agglutination layer. The white regions correspond to the tensile stress, dark regions to the compressive stress**

double-adhesive band was applied for fixation of the envelope on the skin.

According to Fig. 6b, we assume in approximation that the artery is located on a muscle of rigid constant geometry. This means that a radius r increase Δr yields a displacement of the sensor's central region by a distance $K \cdot 2 \cdot \Delta r$, the quantity K < 1 considering "damping" of displacement, especially due to a possible layer of fat tissue. In approximation, this displacement yields a proportional increase of the sensor curvature c and thus an increase of the sensor signal component s which reflects the cardiac activity.

In addition, s is influenced by changes of the neck circumference which arise during breathing. Thus, the respiratory component is introduced.

Fig. 7 shows a typical result. The resulting signal s of the curvature sensor has a character comprising the cardiac component oscillating whose frequency is f_C, superimposed by a slow respiratory component with

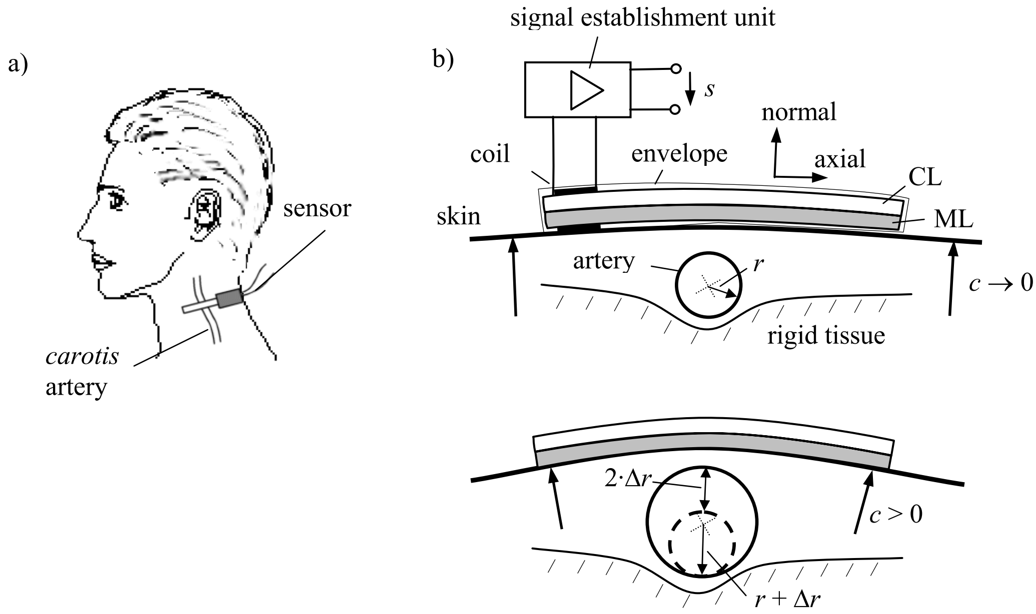

Fig. 6. **a) Application of the skin curvature sensor on the neck over the carotid artery. b) Setup for the skin curvature sensor. The sensor consists of coil for signal pick-up and of the bilayer including magnetoelastic layer (ML) and counter layer (CL). Propagation of blood pressure waves which arise due to cardiac activity induces periodic changes of artery radius *r*. As a result, the curvature *c* of the bilayer increases ($\Delta c \propto \Delta r$) and a cardiac component of the skin curvature signal *s* is induced**

frequency f_R. During the breath hold only the cardiac component is present.

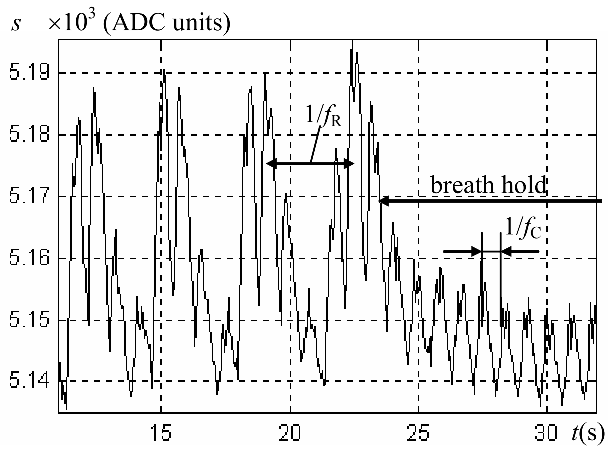

Fig. 7. **The signal *s* of the curvature sensor applied on the neck**

Conclusions

The magnetoelastic bilayer concept serves as a basis for curvature sensors. The concept makes use of mechanic stress distribution within the bent bilayer where pure tensile or compressive stress appears within the magnetoelastic layer. The sensor signal is obtained by means of a coil.

In opposite to the strain gauges which are normally used for registration of curvature the bilayer concept shows no cross-sensitivity to geometry of the object which curvature of which has to be measured.

The curvature sensor can be applied on the skin in order to measure its curvature changes of which reflect various physiologic parameters. If applied on the neck over the carotid artery, the cardiorespiratory activity is monitored. The curvature sensor is easy-to-handle, causes minimal inconvenience to the patient and is suitable for long-time monitoring.

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Magnetoelastinės dvisluoksnės medžiagos odos išlinkio registravimo davikliuose

Reziumė

Odos išlinkio registravimo davikliams tinka naudoti magnetoelastines dvisluoksnias medžiagas. Lenkiant vieną magnetoelastinį sluoksnį, jame atsiranda suspaustų bei ištemptų vietų. Atitinkami magnetinio laidumo pokyčiai kompensuoja vienas kitą. Dvisluoksnė medžiaga susideda iš tarpusavyje sujungtų magnetoelastinio ir nemagnetinio sluoksnių. Lenkiant tokią dvisluoksnę medžiagą, magnetoelastinis sluoksnis priklausomai nuo lenkimo krypties arba įtempiamas, arba suspaudžiamas. Atsirandantys dideli magnetinio laidumo pokyčiai gali būti registruojami rite. Nemagnetinio sluoksnio savybės bei sujungimo procedūra daro didelę įtaką lenkimo daviklio jautrumui. Daviklio efektyvumas yra demonstruojamas nuo kaklo srities registruojant širdies ir kraujagyslių sistemos bei kvėpavimo fiziologinius parametrus.

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