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Abstract. Early detection of cancer can significantly increase the survival chances of patients. Palpation is a traditional method in order to detect cancer; however, in minimally invasive surgery the surgeon is deprived of the sense of touch. We demonstrate how shearing elastography can recover elastic parameters and furthermore can be used to localize stiffness inhomogeneities even if hidden underneath the surface. Furthermore, the influence of size and depth of the stiffness inhomogeneities on the detection accuracy and localization is investigated. © 2018 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: [10.1117/1.JMI.5.4.046001](https://doi.org/10.1117/1.JMI.5.4.046001)]

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1 Introduction

In recent years, the minimally invasive surgery replaced more and more the classical open surgery. The advantages such as faster recovery of the patient and aesthetic aspects are responsible for the increased application of this technique.¹ In addition to the aforementioned benefits of minimally invasive surgery (especially for the patients), the limited access entails some difficulties for the surgeon. Especially, the loss of haptic feedback is very disadvantageous for tumor localization and removal. While in open surgery, the tactile sense is used for the discrimination between tumor and healthy tissues based on different stiffness,² the localization of a tumor in minimally invasive surgery is not possible, hence there is an intensifying investigation on different methods to give the surgeon back his sense of touch.^{3–7} Most methods such as CT or MRT that allow a detection of tumorous tissue can only be used in a preoperational scenario. Due to the environmental changes during the surgery, the comparison of the preoperational data and the real situation can lead to errors in the detection of the tumor. So, it is necessary to ensure an *in situ* measurement during the surgery. As a very promising technique, the elastography is investigated in different research groups.^{4–9} This method uses the different elastic properties of healthy and tumor tissues. Usually, it is a double exposure technique. So, two images of different loading states are compared. The deformation of the sample due to the loading depends strongly on its elastic parameters such as elastic modulus for linear elasticity or shear modulus and locking-stretch for hyperelasticity. Information about these parameters allows conclusions about the type of tissue. To measure the deformation, different methods were applied.^{4,5,10–12}

As we have shown in an earlier work, digital image correlation (DIC) combined with simulations can be used to measure elastic properties of soft materials.¹³ A major disadvantage of the conventional DIC is its limitation to in-plane deformation. To obtain access to the out-of-plane component, DIC can be extended using stereo vision.^{14,15} To obtain the out-of-plane information without the need of a second camera, the DIC has to be combined with another technique. Among others, holographic approaches such as digital holography and electronic speckle pattern interferometry were successfully tested

for an endoscopic use^{16,17} and are sensitive to the out-of-plane deformation. Holography was also applied for photo-acoustic tomography.¹⁸ But a disadvantage of conventional holographic techniques is the reduced stability. Because separate object beam and reference beam, traveling through different optical paths, are employed, environmental changes such as temperature drift, air drift, or vibrations can have significant influence on the measurement result.¹⁹ A technique that can overcome this problem is the so-called shearography.^{20–22} The shearography is an interferometric technique, which is based on the self-interference. The object-reflected light is split into two arms, whereas the mutual modification is applied before both beams become interferometrically recombined.

Shearography is a long established interferometric technique for defect detection on composite structures such as aircrafts and automobile components^{23,24} or in the field of conservation.^{25–27} Recently, the invention of the computational shearing interferometry allows the determination of the full complex wave field.^{28,29} Murukeshan and Sujatha³⁰ have also shown an endoscopic shearographic system for the detection of abnormal growth in the colon path. In their work, they use a biprism as shearing device, which does not allow the change of the shearing amount, which is responsible for the sensitivity of a shearographic system. Furthermore, it is not possible to use phase-shifting technique in this setup, which leads to a poor contrast of the shearograms and therefore to a reduced detection efficiency.

In our work, we investigate the application of a Michelson-based shearography system as a potential technique that can later be combined with the DIC for an elastographic approach. It is our goal to measure strain, a representative quantitative elasticity parameter, by which the discrimination among different types of tissue is enabled. This is accomplished via the comparison of two states (unloaded and loaded), whereas the phase gradient for each of the two states is obtained via shearing interferometry and the resulting strain via the subtraction of the two phase gradient maps. Therefore, lateral shearing was chosen since it delivers a constant (radial and rotational shearing poses a coordinate-dependent sensitivity), adjustable, and strain-related phase sensitivity. Due to the self-interference, an environmentally stable setup can be realized, which performs

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well under harsh surrounding conditions (operation theater and work shop area). Furthermore, the imaging of large areas in less than a second is enabled in a contactless manner. Despite these advantages, certain issues have to be considered in order to ensure a good functionality of the shearography within the minimally invasive environment. First, there are different possibilities to apply the loading. The most common excitation of the object is based on thermal source. For the use in minimal invasive surgery, these methods seem rather unsuitable due to dehydration of the tissue and limited space for thermal sources. Tactile methods, which are also conceivable, have the disadvantage of an only localized loading and a certain risk of injuring vessels, nerves, and other important functional tissues. A very convenient method is however a change of pressure, which represents a parameter that is already existent in minimally invasive surgery. To increase the space inside the body, a slight over pressure is applied, which can be controlled and hence be used for the loading. In addition to the loading mechanism, as mentioned, the sensitivity of the system also depends on the lateral shearing distance. A higher shearing distance leads to a higher sensitivity but increases the probability of speckle decorrelation.

In this paper, we investigate the influence of shearing distance and amount of loading on the measurement result with a special focus on the sensitivity and localization.

2 Methodology

A typical setup of a shearographic sensor is shown in Fig. 1. It consists of a laser with an expansion device (in this case, a combination of expansion lens and holographic diffuser is used to homogenize the initial Gaussian beam distribution), a Michelson-interferometer with a slightly tilted mirror, a camera, and an excitation mechanism. If the expanded laser beam illuminates a rough surface, the interference of the scattered waves leads to the occurrence of speckles.⁴

This speckle image is superimposed with a sheared version of itself (due to the tilted mirror M1) on the camera. The phase difference of the speckles, which overlap on the detector, can now be determined either by spatial³¹ or by temporal phase-shifting.^{32,22} As long as no dynamic effects are of any interest (as it is in our case), temporal phase-shifting leads in general to less noisy images. The implementation of temporal phase stepping can be accomplished changing the distance between the mirror M2 and the beam splitter. After the phase information is stored, the object is excited (for example, by the application of an external pressure). This loading results in a deformation of the object surface. If there are inhomogeneities in the structure

Fig. 1 Typical setup of a shearographic sensor.

of the object such as materials with different stiffness, the deformation in this part differs from the one in the homogeneous part. Due to this deformation, the phase difference between two interfering points in the sensor plane changes. Subtracting the phase maps before and after loading this change can be calculated, and information about strain and hence the elasticity distribution of the object can be obtained.

In contrast to holography, the resulting phase map is not directly related to the deformation, but the differential quotient along the shearing direction. In general, if one assumes a shearing direction in x and a shearing distance of δx , the resulting phase $\Delta\phi$ consists of three components:²²

$$\Delta\phi = \delta x \left(\frac{\partial u}{\partial x} \vec{k}_s \cdot \vec{e}_x + \frac{\partial v}{\partial x} \vec{k}_s \cdot \vec{e}_y + \frac{\partial w}{\partial x} \vec{k}_s \cdot \vec{e}_z \right), \quad (1)$$

where $\vec{k}_s$ is the sensitivity vector (vector along the bisector of the angle between illumination and detection direction), and $\frac{\partial u}{\partial x}$, $\frac{\partial v}{\partial x}$, $\frac{\partial w}{\partial x}$ are the differential quotients of the displacement u , v , w in x -direction. Because the displacement is expected mainly in out-of-plane direction, often illumination and detection-direction is chosen perpendicular to the object surface. Thus, the y - and x -component of $\vec{k}_s$ is zero and Eq. (1) can be simplified to

$$\Delta\phi = \delta x \left(\frac{\partial w}{\partial x} \vec{k}_s \cdot \vec{e}_z \right). \quad (2)$$

For a small shearing distance and with $\vec{k}_s \cdot \vec{e}_z = \frac{4\pi}{\lambda}$ (λ is the illumination wavelength), an estimation of the displacement gradient $\frac{\partial w}{\partial x}$ is possible:

$$\frac{\partial w}{\partial x} \approx \frac{\Delta\phi}{\delta x} \cdot \frac{\lambda}{4\pi}. \quad (3)$$

This displacement gradient can be interpreted as a part of the strain tensor:

$$\varepsilon = \begin{bmatrix} \varepsilon_{xx} & \varepsilon_{xy} & \varepsilon_{xz} \\ \varepsilon_{yx} & \varepsilon_{yy} & \varepsilon_{yz} \\ \varepsilon_{zx} & \varepsilon_{zy} & \varepsilon_{zz} \end{bmatrix} = \begin{bmatrix} \frac{\partial u}{\partial x} & \frac{1}{2} \left(\frac{\partial w}{\partial y} + \frac{\partial v}{\partial x} \right) & \frac{1}{2} \left(\frac{\partial u}{\partial z} + \frac{\partial w}{\partial x} \right) \\ \frac{1}{2} \left(\frac{\partial v}{\partial x} + \frac{\partial u}{\partial y} \right) & \frac{\partial v}{\partial y} & \frac{1}{2} \left(\frac{\partial v}{\partial z} + \frac{\partial w}{\partial y} \right) \\ \frac{1}{2} \left(\frac{\partial w}{\partial x} + \frac{\partial u}{\partial z} \right) & \frac{1}{2} \left(\frac{\partial w}{\partial y} + \frac{\partial v}{\partial z} \right) & \frac{\partial w}{\partial z} \end{bmatrix}. \quad (4)$$

As mentioned before, different illumination angles and shearing directions allow the detection of other parts of the tensor. But because shearing in z -direction is not possible, the gradients $\partial u/\partial z$, $\partial v/\partial z$, $\partial w/\partial z$ cannot be determined with shearography and has to be measured with another technique or obtained via solving an inverse problem using for instance a realistic finite element model from which other strain tensors and elastic parameters such as stress tensors and shear modulus can be obtained.¹³

3 Sample

To simulate an organ affected by a tumor, a silicone sample has been used. The important parameter for elastography is the shear modulus. In accordance to the surgical results, two different types of silicone, exhibiting different stiffness properties, had

Fig. 2 (a) Sketch of the silicone sample with dimensions in mm. (b) Photo of the sample with illumination from the back.

been chosen (tumorous tissue stiffer than soft tissue). According to Ref. 33, the elastic modulus of cancerous tissue, represented by hard-silicone spheres, can be up to 14 times larger than healthy tissue, represented by soft silicone. The hard-silicone spheres were placed at certain positions within the surrounding soft silicone. The whole sample consists of six spheres, three with different sizes but the same depth and three with different depths but the same sizes. A sketch of the sample can be seen in Fig. 2. Although the elastic properties of our sample are similar to those of a real tissue, the optical properties may differ. Due to different absorption or scattering, the speckle contrast can be reduced. For obtaining high speckle contrast, the wavelength should be adapted to the tissue.

4 Experiment

The experimental setup is shown in Fig 3. In detail, it consists of a Michelson-interferometer, a frequency-doubled Nd-YAG laser

(532 nm, 400 mW), an expansion optic to increase the diameter of the laser beam and a holographic diffuser, a PCO camera 1200-s CMOS with 1280×1024 pixels, and the loading mechanism.

As mentioned before, tactile and thermal methods seem rather unsuitable for the use in minimally invasive surgery, so we investigated only the loading by changing the pressure. Therefore, the sample is located inside a chamber ($45 \times 35 \times 15 \text{ cm}^3$) with a window at the front and the load is applied uniformly and contactless by changing the air pressure. To avoid reflections from the window into the camera, the chamber is slightly tilted, but the sample is perpendicular to the interferometer. The pressure inside the chamber is controlled via a Vaccubrand 510 NT pump. The pump has a pressure sensor combined with a control loop, what enables the generation of stable pressure conditions, with fluctuations of only 0.2 mbar during the measurement time of 0.5 s. The distance between

Fig. 3 Experimental setup.

the object (chamber) and the camera is roughly 80 cm, which leads to a 0.06-mm/pixel resolution. For sufficient sampling, the smallest speckle size results from the interference of light originating from the two most distant points at the exit pupil of diameter D . Hence, the speckle size a_s varies with the imaging distance z_2 , which can be expressed using the focal length f and the lens diameter to result in³⁴

$$a_s = 1.2 \cdot \lambda \frac{z_2}{D} = 1.2 \cdot \lambda \frac{f}{D} (1 + \beta') = 1.2 \cdot \lambda F (1 + \beta'), \quad (5)$$

the choice of the F -number has important influence on the image quality. In our case, we adjust the F -number so the speckle size matches approximately to the pixel size of 12 μm . This leads on one hand to a acceptable modulation depth and ensures on the other hand a high lateral resolution.³⁵

The data are acquired using the in-house developed software in ITOM,³⁶ which gives also the possibility to control a piezo (P3-150 piezo Controller). During a measurement, the piezo performs five equidistant phase steps and with a suitable algorithm the phase image can be obtained.³² Because speckles lead to very noisy phase maps, filter algorithms such as Gaussian filter are applied prior to the phase unwrapping algorithm. The unwrapping is performed with Goldstein algorithm³⁷ to gain the absolute phase.

5 Results

5.1 Sensitivity

The change of pressure, which can be achieved in minimally invasive surgery, is not unlimited, but the detection efficiency of the different tumors has to be guaranteed. However, the sensitivity in shearography can further be increased when increasing the shearing distance.

To ensure a good lateral resolution, we investigated the different foreign bodies in our phantom using individual

measurements of reduced field of view (not entire object is recorded in a single measurement), as shown in Fig. 4.

Here, we concentrate on the hard-silicone sphere on the upper-right side (13 mm diameter, 10 mm depth). This image was acquired with a shearing distance of 15 pixels (x -direction), which corresponds to 1 mm in the object plane. The pressure was decreased by 9 mbar between reference and measured shearograms. The hard-silicone sphere can be detected in (a) the modulated as well as in (b) the demodulated image.

With Eq. (3), the demodulated phase map can be converted into a strain map with the real strains $\partial w / \partial x$ [see Fig. 4(c)].

As previously mentioned, the detection efficiency depends on parameters that cannot be controlled in real life (depth and the size of tumorous tissue) and adjustable parameters (shearing distance and amount of pressure change). Therefore, the influence on the detection sensitivity of the latter two adjustable parameters will be investigated in the following. If the signal-to-noise ratio (SNR) is larger than three, we assume an unambiguous detection of the hard-silicone spheres. The signal is defined as $\frac{|\Delta_{\max}| + |\Delta_{\min}|}{2}$, with the maximum of the signal $\Delta_{\max}$ and the minimum of the signal $\Delta_{\min}$ (demodulated phase map). The noise is determined using the standard deviation in the intact part of the sample. The measurements showed that the standard deviation (in demodulated phase-images) is not affected by the amount of shearing or the pressure change applied. The value is around 0.4 rad. Note that due to the symmetry of the silicone spheres, the shearing direction has no significant influence on the signal, so we present here only the results for a shearing along the x -axis.

5.2 Influence of Sphere Size

In Fig. 5, the results for a shearing distance of 20 pixels (1.2 mm in the object plane) are shown. We investigated three different pressure changes: 2, 4, and 9 mbar. With smaller changes of pressure, no signal can be determined and with larger values speckle decorrelation occurs for the largest sphere, which results in a blurred signal. As expected, the signal correlates strongly with the amount of pressure change. Assuming a linear-elastic

Fig. 4 Shearogram of a hard-silicone sphere (a) phase image before demodulation, (b) phase image after demodulation, and (c) strain map.

Fig. 5 SNR for three different sphere sizes dependent on the applied change of pressure.

behavior, we can fit a linear function. In good approximation, the fitted graph drops down to zero for small pressure changes. It can be determined that for an unambiguous detection of all three spheres a change of only 4 mbar is sufficient (higher SNR than three). These conditions can easily be implemented during a surgery. Although the three curves show a linear dependence on the pressure change, a nonlinear dependence on the sphere size was observed. While the signals for 8 and 10 mm, sphere size exhibits only a small difference, the 13-mm one is about two times higher for 10 mm. The effect is caused by the different sphere sizes, which automatically result in a reduced distance to the surface changes for larger spheres (3.5 mm for the 13-mm sphere, 5 mm for the 10-mm sphere). As shown in Fig. 2, the center of all spheres are 10 mm under the surface.

Furthermore, the shearing distance influences the signal and therefore the SNR. We investigated the SNR for a fixed change of pressure of 4 mbar, which is shown in Fig. 6. Note that for fixed pressure changes of 2 or 9 mbar, the SNR is lower respectively higher, but the general behavior is identical to Fig. 6. Surprisingly, the data points can also be fitted with a linear function, although this would lead to an unphysically large signal for large shearing distances. Actually, the linear behavior is again just an approximation for small shearing distances. Because our maximal shearing distance is only around 1.2 mm in the object plane, which is relatively small compared to the sphere sizes employed (and therefore with the deformation), we can use the linear approximation here. Furthermore, the linear behavior ensures the validity of Eq. (3) and therefore the calculation of real strains out of the phase maps. Concerning the sensitivity of the system in Fig. 6, it can be seen that a shearing distance of 15 pixel (0.9 mm in object plane) allows the detection of all hard-silicone spheres. In summary, with the right combination of shearing distance and amount of pressure change, an unambiguous detection of most of the spheres is possible. Using our fitted

Fig. 6 SNR for three different sphere sizes dependent on the applied shearing distance.

curves, we estimated the necessary shearing distance to gain an SNR of three. The result varies between 0.1 mm in object plane for the 13-mm-diameter sphere with a pressure difference of 9 mbar and 1.8 mm (8 mm sphere/2 mbar). Although this is only a rough estimation, it shows that a relatively small shearing distance of under 2 mm in most cases is enough to detect the spheres.

5.3 Influence of Sample Location

That the depth has an important influence on the signal can be verified by evaluating the spheres in the other row of the sample. In Fig. 7, the results are shown for a depth of 6 and 10 mm (both spheres have a diameter of 10 mm). For a depth of 14 mm, the signal is in the range of the noise and cannot be evaluated. In principle, it is possible to detect it by using a higher pressure modulation than 9 mbar, but we concentrate here on smaller changes to ensure a suitable implementation in minimally invasive surgery. As expected, the SNR for 10 mm depth indicates also a nearly linear dependence on the pressure change. For the sphere positioned in 6 mm depth, the signal (respectively the SNR) is around four times higher than for the more deeply positioned sphere. Consequently, deeper positioned sample leads to a reduced sensitivity and hence requires a higher loading, which is not necessarily available in minimally invasive surgery. Furthermore, it is noticeable that the linear fit for the sphere near the surface does not drop to zero and also that the deviation of the data points is relatively large. So, the assumption of a linear-elastic behavior is no longer correct. In fact, soft tissue and silicone likewise are often described using a hyperelastic model (Arruda Boyce). For a small amount of pressure (in this case 2 and 4 mbar), it can still be approximated using Hooke's law.

5.4 Determination of Lateral Sample Localization and Its Size

Having verified that the discrimination of materials exhibiting different stiffness properties is possible using shearography, this section deals with the estimation of the physical extent of these spheres. For the use in minimally invasive surgery, a reliable localization is important to not damage healthy tissue or leave tumorous tissue in the body. While the correct center position can be calculated³⁸ if the shearing distance is known, the detected size depends not only on the shearing distance but also on material properties as well as on size and depth of the spheres. In Fig. 8, the results for the spheres (diameter 10 mm) with a depth of 10 and 6 mm are shown for a shearing distance of 20 pixel (1.2 mm) and a loading of 4 mbar. Besides the

Fig. 7 SNR for two different depth location of a sphere with 10-mm diameter dependent on the applied change of pressure.

Fig. 8 Strain map for a pressure change of 4 mbar and a shearing of 1.2 mm.

Fig. 9 (a) Size of the signal for a fixed depth of 10 mm and (b) size of the signal for a fixed sphere size of 10 mm.

difference in the maximum strain, which was evaluated in the previous chapter, it can be assumed that the signal for the deeper sphere is broader than the other one. For the evaluation of the signal size, we analyze a line plot through the maximum of the strain image. Due to the demodulation process, the signal is not symmetrical around the zero. We therefore defined a center line in the middle of the maximum and the minimum of the signal. The size is then the distance between the first intersection points (starting from the middle of the signal) of the signal and this center line. In Fig. 9(a), the signal size in dependence of the sphere size is shown for fixed shearing distance of 1.2 mm and pressure change of 9 mbar. The measured signal size is much larger than the real size of the spheres, even if taking into account the shearing distance. This means that the generated deformation on the surface is more than twice as large as the sphere diameter, which makes the estimation of the sphere size difficult. On the other hand, the absolute error seems to be identical for all of the three spheres. This behavior changes if the depth of the spheres varies. As can be seen in Fig. 9(b), the depth influences the detected signal size. If the sphere is closer to the surface, the estimation of the signal size improves. This ambiguity can be tackled via the application of model-based parameter identification using, for instance, finite element modeling (FEM) as in Ref. 13. The *a priori* knowledge of the pressure and adjustable parameters such as the location, geometry, as well as elastic properties of surrounding soft tissue and hard tissue are used as an input parameter for the FEM. Either a look-up table with various possible scenarios or an iterative adjustment of these parameters is conducted using the FEM with the goal to obtain a good match of simulated strain map and measured strain map. In that manner, the inverse problem can be solved and the important dimensional as well elastic parameters can be obtained. The proposed procedure will be an important part of future research content since the emphasis of this paper is to demonstrate that stiffness inhomogeneities can be measured using the shearography, directly resulting in strain-related values.

5.5 Reproducibility

For a reliable detection, the reproducibility is of great importance. We used the two spheres of 10 mm diameter of our silicone phantom, which are both positioned at a depth of 10 mm, see Fig. 2, for evaluating the reproducibility. In Fig. 10, the associated strain maps are shown. Although not all parameters could

Fig. 10 Comparison of the strain maps (1.2-mm shearing and 9-mbar pressure) of a spheres with a diameter and depth of 10 mm. (a) Sphere in the “diameter-row.” (b) Sphere in the “depth-row.” (c) Difference of both images.

be perfectly controlled during the manufacturing process of the sample, the signals do not show any significant difference in strain or size. The difference of both images shows a mean of $-28.6 \mu\text{m/m}$ and a standard deviation of $46.8 \mu\text{m/m}$.

The difference for two parts of the object without hard spheres leads to a similar standard deviation of $36.97 \mu\text{m/m}$. So, most of the signal difference in Fig. 10 is due to speckle noise.

Therefore, it can be assumed that the system shows a very good reproducibility, which is a requirement for use in minimally invasive surgery.

6 Conclusion

In this work, we presented a first step toward the application of a flexible Michelson-based shearography system for the application in a minimally invasive surgery environment. To evaluate the measurement system, a soft-silicone sample with hard-silicone spheres was manufactured, which exhibits similar elastic properties of healthy soft tissue in comparison to tumors tissue. The investigations showed that a detection of different sizes and depth of the spheres can be ensured using only a small amount of pressure change between 2 and 9 mbar, which can easily be provided in the minimally invasive surgery. Furthermore, an increase in the amount of shear results in an improvement of the SNR of the system.

The shearography system presented in this paper uses changing pressure, which represents the most prominent excitation source available in minimally invasive surgery. The pressure is needed to expand the body's cavity (e.g., thorax) to provide the surgeon with sufficient space for carrying out the operation. Due to the pressure, the shape and location of organs are deformed in comparison to preoperational data. Moreover, the patient may also have a different hydration state and body position in comparison to the time when the preoperational data were taken. Therefore, the preoperational data cannot directly be applied to the patient. It requires a correction and guiding system. Shearography system enables the intracorporeal discrimination between different tissues. It can also reveal information, which cannot easily be measured via preoperational imaging modalities such as fat and its thickness, due to its different elasticity parameters. Previously, in open surgery discrimination between different types of tissue based on their elastic properties was available, known as palpation. Therefore, the pressure excited shearography system represents a very important tool for guidance and discrimination during minimally invasive surgery.

However, the prediction of the size of the spheres cannot be performed based on shearography only, which is due to the strong dependence of the accuracy of deformation measurement on parameters such as depth and material properties. Furthermore, the most suitable shearing distance depends on the size of the foreign body of increased stiffness properties (tumor). Therefore, exact evaluation of the shearograms in minimally invasive surgery relies on further knowledge of the morphology of the sample, which can be obtained using preoperative data in combination with other techniques such as CT or MRT. For further investigations, the use of simulations and the combination with DIC can be helpful to solve the problem of the size estimation and the determination of other elasticity parameters such as stress and Young's modulus. The excellent reproducibility of the shearographic measurement system allows such a simulation approach using for instance finite-element-modeling.

Disclosures

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