Abstract— Despite better performance over primary repairs, tension-free ventral hernia repairs with mesh still suffer from a high recurrence rate. High stress gradients in the mesh are thought to contribute to hernia recurrence. We propose a postoperative monitoring system based on a coupled pair of magnetoelastic strain sensors to enable patients and physicians to non-invasively measure and track the strain distribution across the hernia mesh. Our design combines an encased resonator with a spring-loaded transducer to achieve high signal amplitude with a wide dynamic range. We also demonstrate a fabrication protocol to integrate the resonant strain sensors with a commercial polypropylene mesh. The packaged sensor is capable of detecting up to 37.5 millistrain, an order of magnitude greater than previously demonstrated.

I. INTRODUCTION

Ventral hernia repairs are among the most common procedures performed by general surgeons in the United States, with an estimated 348,000 cases in 2006, resulting in $3.2 billion in associated expenses [1]. One of the primary drivers of cost and poor patient outcomes is the high recurrence rate both in primary repairs (30-50%) and tension-free mesh repairs (10-20%) [2], [3].

Ventral hernias occur when an internal organ protrudes through a tear in the abdominal muscle wall and are typically repaired either through primary closure (small defects <2-3 cm can be closed by directly suturing the edges of the wound together) or, for larger defects, tension-free repair (a synthetic or biological mesh is used to bridge or reinforce the defect). In addition to complications such as infection and adhesion formation, one of the major causes of hernia recurrence is high stress concentrations across the surgical mesh (in particular at the tissue-mesh interface) with mechanical failure accounting for 18% of all recurrent hernias [2]. A postoperative monitoring system to non-invasively measure the strain distribution across the mesh can assist patients in tracking the impacts of their post-surgical activities during the healing period as well as enable physicians to conduct postoperative checkups in the event of a potential failure.

Strain is traditionally measured using metallic or piezoresistive strain gauges, combined with measurement electronics to detect the strain-dependent shifts in voltage or resistance. However, when scaling up the number of strain gauges necessary to produce high resolution strain maps, the burden from extra interconnects, power source, and other electronics increases drastically. The embedded electronics also precludes customization of the mesh size to the patient’s wound geometry, which is critical for successful repairs. Alternatively, numerous optical-based strain measurement strategies, including digital image correlation and moire interferometry, have been developed as non-contact methods to measure stress gradients in safety critical systems [4], [5]. These methods typically involve tracking the position of markers on the substrate surface to calculate strain. An optical approach, while potentially useful for guiding surgeons during the surgical repair, is incapable of visualizing the mesh inside tissue during the postoperative period.

Magnetoelastic materials have been used in sensors to measure a wide range of physical parameters, including pressure, pH, mass loading, and fluid viscosity [6]. These sensors are governed by the Villari effect, which describes how a material’s magnetic susceptibility is coupled to its mechanical deformation due to the realignment of the magnetic moments within the material. Thus, physical parameters that impact the mechanical state of the sensor can be detected remotely via magnetic fields, making them favorable for use as implantable sensors. Previously, magnetoelastic sensors based on Metglas 2826MB, a commercially available, ribbon-like amorphous ferromagnet, have been demonstrated in various biomedical applications, including detecting stresses in bone fractures, actuating fluid flow to relieve glaucoma, and monitoring mass loading on bile duct stents [7]–[9]. In this paper, we propose a coupled magnetoelastic resonant sensor for tracking large strains in hernia mesh prosthetics.

II. THEORY OF MAGNETOElastic RESONANT SENSORS

A. Resonance Frequency and the ΔE Effect

Magnetoelastic sensors can be measured either through non-resonant or resonant methods. Non-resonant detection strategies involve directly measuring the magnetic permeability of the sensor. However, these methods tend to be more prone to picking up stray magnetic fields as well as orientation discrepancies.

Alternatively, magnetoelastic materials will mechanically vibrate at a particular resonant frequency when exposed to an external AC magnetic field. The resonant frequency of these ribbon sensors can be calculated from Equation 1 [6].

$$f_0 = \frac{1}{2\pi} \sqrt{\frac{E}{\rho(1-\nu^2)}}$$

where L is the sensor length, E is the Young’s modulus, ρ is the density, and ν is the Poisson’s Ratio. The magnetic flux generated by the mechanical oscillations can be detected with a pick-up coil.

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When a magnetoelastic sensor is strained, the magnetic dipoles in the material realign, resulting in a change in the magnetic field. The stress-induced magnetization causes a shift in the Young’s modulus (also known as the ΔE effect) (Equation 2), which in turn affects the fundamental resonant frequency of the sensor (Equation 1).

\[ E|_{\mathcal{B}_0} = \frac{\sigma}{\varepsilon_{\text{elastic}} + \varepsilon_{\text{magnetic}}} \]  

(2)

B. Determinants of Signal Amplitude

The signal amplitude of the resonant peak(s) is dependent on the degree of mechanical vibration. Thus, a critical consideration in designing magnetoelastic sensors is the tradeoff between sensor size and signal amplitude. An external DC biasing field is required to shift the operating point of the sensor to a regime with the highest magnetostriction (strongest coupling between the material’s mechanical and magnetic properties), so that the mechanical vibrations are large enough to be detected. The DC biasing field can be provided via an additional biasing coil or a permanent magnets placed adjacent to the resonator. Signal amplitude is also strongly dependent on the degree of damping due to the viscosity of the surrounding environment or mass loading such as from surface coatings. In water, the signal amplitude measured with our coil setup drops by approximately half of the measured signal in air. When a magnetoelastic sensor is clamped on both sides to measure strain, the mechanical oscillations are further dampened. Finally, the detected signal amplitude is a function of the distance between the interrogation coil and the sensors. Signal amplitude drops off exponentially with increasing distance due to the fall-off of electromagnetic energy transmitted and received from the sensor. In order to achieve postoperative monitoring, a measurement distance of at least 5 cm is required to accommodate >99% of people [10].

C. Coupled Strain Sensor Design

Huber, et al. reported a coupled strain sensor design that produces high sensitivity strain measurements using two strips of magnetoelastic materials, a so-called “transducer” and a “resonator” [11]. The transducer is bound to the substrate of interest to detect strain. The resonator is placed immediately adjacent to the transducer in a rigid casing where it can freely resonate. The resonator’s resonant frequency is biased by the stress-induced magnetization of the transducer to provide strain information. Unfortunately, most magnetostrictive materials magnetically saturate at low strains on the order of tens to hundreds of μstrain. For our application, due to the highly flexible nature of the hernia meshes, high strains on the order of tens to hundreds of millistrain is observed.

Pepakayala, et al, proposed a magnetoelastic strain sensor with an integrated spring component to increase the dynamic range of the sensor to a peak of 2.8 millistrain [12]. However, due to constrained boundary conditions of the sensor, the signal amplitude is very low (on the order of tens of μV) even at close proximity with the interrogation coil.

The proposed design in this paper takes advantage of the high signal amplitude generated from an unconstrained resonator, combined with the increase in dynamic range achieved by incorporating spring elements into the transducer design to achieve strain measurements up to 37.5 millistrain, an order of magnitude greater than previously demonstrated.

III. SENSOR DESIGN AND EXPERIMENTAL METHODS

A. Sensor Integration with Polypropylene Hernia Mesh

Currently, woven or knitted polypropylene meshes are the most commonly used meshes for hernia repairs, with over one million meshes implanted each year [13], [14]. Polypropylene meshes are low cost, have high tensile strength, and suffer from low infection risk. The highly porous structure formed by weaving or knitting improves flexibility of the mesh and encourages integration with tissue for optimal healing [13], [14]. In order to reduce the risk of adverse events and facilitate passage through FDA approval, our sensors are designed to be integrated into currently used polypropylene meshes and encapsulated in biocompatible materials.

B. Sensor Materials and Design

Metglas 2826MB (Fe_{80}Ni_{13}Mo_{3}B_{18}) (Metglas Inc., Conway, SC) was chosen due to its high magnetostriction coefficient, high magnetic permeability, and low biasing field. Metglas 2826MB is commercially available at low cost in ribbon-form and is easily batch patterned for high-throughput processing. Arnokrome 5 (coercivity = 20-50 Oe, remanence = 2-16 kGauss) (Arnold Magnetics, Rochester, NY) ribbon was used to provide a biasing field.

The Metglas 2826MB transducer was patterned using photochemical machining (PCM) (Northwest Etch Technology, Tacoma, WA). Briefly, this process involves spinning a coat of photoresist over the metal, selectively UV exposing the photoresist with a mask, chemical etching of the metal through non-exposed regions in the photoresist, and finally stripping the photoresists to leave only the desired geometries. PCM can achieve high resolutions down to 127 μm features. The transducer design consists of a 10x2 mm bar component with spring components and anchoring points on either side. Dimensions are shown in figure 2F.

The Metglas 2826MB resonator and Arnokrome 5 were cut into 7x1 mm and 9x3 mm ribbons respectively using a DAD3240 Disco automatic dicing saw (Disco Corporation, Tokyo, Japan). To form a semi-rigid casing around the resonator to permit undampened oscillations, a polypropylene casing was fabricated by hot embossing a 1/32” sheet of polypropylene with a laser-etched acrylic mold at 50°C and 13 MPa for 10 minutes. For commercial fabrication
metallic mold can be machined to increase the embossing temperature for faster, more robust processing. The patterned Metglas 2826MB transducer and permanent magnet is positioned inside the polypropylene casing and sealed with 50 µm heat-sealable, bi-axial magnetic (BOPP) film (Impex Global, LLC, Houston, TX) by applying 10 kPa pressure at 150°C for 10 seconds across the casing surface. Low pressure and short seal times are necessary to prevent deformation of the polypropylene casing. Once sealed, the encased resonator is aligned with the transducer and attached at the center point. Due to the semi-rigid nature of the resonator casing, it is critical that the resonator is attached only at the center point to enable the underlying mesh to flex freely. Full attachment to the transducer would not only restrict movement of the underlying mesh and transducer, thus preventing strain measurements, but also create additional stress concentrators within the mesh, thus preventing strain measurements, but also create additional stress concentrators within the mesh.

C. Coil Construction and Setup

To detect the resonant frequencies in the strain sensors, a set of three coils were constructed following the parameters listed in Table 1. A Keysight 33521B waveform generator (Keysight Technologies, Santa Rosa, CA) and LT 1210 current amplifier (Linear Technology, Milpitas, CA) were used to sweep a time-varying magnetic field in the regime of interest. The DC bias and receive coils are positioned orthogonally to the transmit coil at a null point between the transmit coils to minimize feedthrough from the excitation magnetic field. The DC bias coil is not used in experiments where the Arnokrome 5 permanent magnets are in place. All three coils are oriented so that the emitted/received magnetic fields are aligned with the longitudinal axis of the magnetoelastic resonator. A R&S FSP 13 spectrum analyzer (Rohde & Schwarz, Columbia, MD) and custom analysis scripts written in Matlab is used to detect the magnetic resonance picked up by the receive coil. Resonance peak data was collected using a frequency sweep with no biasing field as a calibration reference to account for environmental effects on the magnetic field. Data was passed through a bandpass filter to filter out drift over time and noise in the frequency sweeps.

D. Strain Measurement Setup

A Mark-10 ES20 linear tensiometer (Mark-10 Corporation, Copiague, NY) was modified to induce known strains on the instrumented mesh. A custom 1.5 cm wide paper grip was machined out of aluminum by the Marvel Nanofabrication Laboratory Machine Shop (UC Berkeley, CA). Aluminum has a magnetic permeability similar to air, which is beneficial for not redirecting magnetic field energy away from the sensor.

IV. RESULTS AND CONCLUSION

A. Signal Amplitude is Halved When Sensor is Clamped

Figure 3a shows the effect of anchoring the transducer sensor. As expected, the resonance frequency stays fairly constant (311.8 kHz and 313.8 kHz respectively for the free and bound sensors) – the small shift is likely due to small strains induced in the clamping process. However, the signal amplitude drops from 467 µV to 214 µV upon clamping.

B. Tradeoff Between Signal Amplitude and Sensor Size

Because the resonator needs to be encapsulated in a semi-rigid structure, it is ideal to minimize the size as much as possible to reduce the impacts on flexibility of the mesh. However, Figure 3b shows that as sensor size is decreased, the signal amplitude drops proportionally; a 3 mm sensor is undetectable at a distance of ~1 cm from the interrogation coil. Furthermore, as sensor size is reduced a higher biasing field
necessary for use with these highly elastic meshes. The target would be to measure strains up to 100 millistrain at a distance of 5 cm. In addition, further work is required to assess the bending strains induced when mesh is wrinkled. One downside of this approach is that the sensors are not MRI compatible. However, since CT is used over MRI in the vast majority of hernia cases, this should have minimal impact on patient recovery and postoperative management.

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Figure 3. A. A spring-loaded sensor with 7mm bar was measured with ends free and clamped. B. As sensor size decreases, the signal amplitude drops and the required biasing field increases. The measured resonance frequency tracks very closely with the theoretical resonance frequency. C. For a 7x1 mm resonator 16 Oe biasing field is sufficient for good signal detection. D. Strain measurements are conducted with the Mark-10 tensiometer. E and F. Linear strain is measured up to 37.5 millistrain with the packaged coupled sensor setup.

C. Dynamic Range of Coupled Magnetoelastic Sensors

Resonance frequency shifts induced by known strains were measured using the Mark10 tensiometer to establish a calibration curve. With the assembled coupled magnetoelastic design combined with a spring-loaded transducer, we were able to achieve linear strain measurements up to 37.5 millistrain, corresponding to a resonance frequency of 309.9 kHz at a distance of 1.5 cm from the interrogation coil before saturation of the transducer. The SNR remained fairly constant with strain. The dynamic range of the sensor can be increased further by optimizing the relative lengths of the bar vs the spring components in the resonator. Some hysteresis was observed when the strain was decreased, but can likely be calibrated out by tracking whether the strain shifts are on the ascending vs descending portions of the hysteresis curve.

V. CONCLUSION

In this paper, we demonstrate a coupled magnetoelastic strain measurement system based on an encased resonator for high signal detection coupled with a patterned spring-loaded transducer for increased dynamic range measurements up to 37.5 millistrain. We’ve also shown an assembly methodology to integrate the coupled sensors with a flexible mesh structure. Future work will include refinement of transducer geometry and additional integration of a permanent magnet to obtain higher sensitivity data at an expanded dynamic range.