Sensorizing a Compression Sleeve for Continuous Pressure Monitoring and Lymphedema Treatment Using Pneumatic or Resistive Sensors

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***Abstract*— Smart soft wearable devices have great potential to change how technology is integrated into daily life. A particularly impactful and growing application is continuous medical monitoring; being able to stream physiological and behavioral information creates personalized datasets that can lead to more tailored treatments, diagnoses, and research. An area that can greatly benefit from these developments is lymphedema management, which aims to prevent a potentially irreversible swelling of limbs due to causes such as breast cancer surgeries. Compression sleeves are the state of the art for treatment, but many open questions remain regarding effective pressure and usage prescriptions. To help address these, this work presents a soft pressure sensor, a way to integrate it into wearable devices, and sensorized compression sleeves that continuously monitor pressure and usage. There are significant challenges to developing sensors for high-pressure applications on the human body, including operating between soft compliant interfaces, being safe and unobtrusive, and reducing calibration for new users. This work compares two sensing approaches for wearable applications: a custom pouch-based pneumatic sensor, and a commercially available resistive sensor. Experiments systematically explore design considerations including sensitivity to ambient temperature and pressure, characterize sensor response curves, and evaluate expected accuracies and required calibrations. Sensors are then integrated into compression sleeves and worn for over 115 hours spanning 10 days.**

I. Introduction

As the world of small wearable devices enters the world of big data, new possibilities arise for interdisciplinary advances and improving quality of life. One such path is to leverage unobtrusive continuous monitoring to improve our understanding of human behavior and assistive devices. Medical applications of smart textiles and their personalized data streams can lead to more accurate diagnoses and more effective, customizable treatments.

Tactile information in particular can provide valuable insights about how humans interact with their environment or about the performance of assistive devices. However, integrating such sensors into a wearable garment poses significant challenges. They must accurately detect pressure on soft deformable human skin, even over long periods including various activities, postures, and ambient conditions. For safety and comfort, sensors must be small and conformal so that a sensorized garment is no more obtrusive than the original. Widespread adoption also necessitates minimizing calibration or maintenance by the user, and scalable fabrication.

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[1] Videos and details: https://people.csail.mit.edu/delpreto/smart-sleeve/

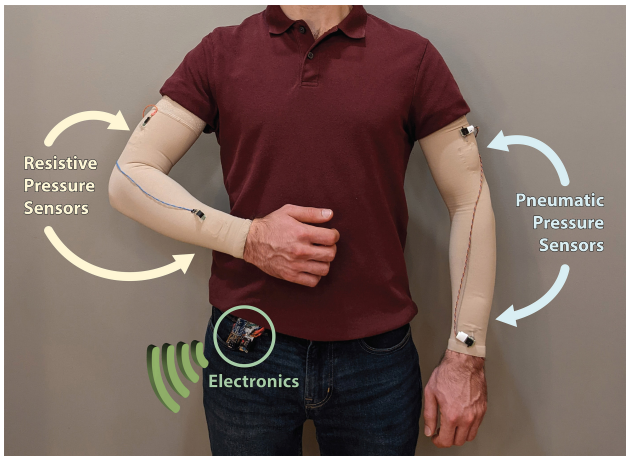


Fig. 1: The self-contained wearable system features soft pneumatic or resistive sensors unobtrusively integrated into a compression sleeve to monitor pressure and usage.[1] Small electronics perform data acquisition and communication.

This work aims to address these challenges by designing, evaluating, and embedding pressure sensors in textiles such as the compression sleeves shown in Figure 1. It presents a soft pouch-based pneumatic sensor and its application to a wearable monitoring system, and compares it to a commercially available resistive sensor. Experiments investigate accuracy, characterization, calibration, and robustness to ambient conditions. Results can benefit deployments of soft pressure sensing for wearable devices or robotic applications.

The current work uses the resulting sensors and insights to take a step towards improving treatment of lymphedema following breast cancer surgery. This is a chronic and progressive disease characterized by swelling in the arm, hand, or trunk on the side of breast cancer treatment, which may progress to an irreversible stage [1]. The possibility of developing lymphedema is a major fear for breast cancer patients, particularly due to its impact on quality of life and social dynamics. Main risk factors for lymphedema include lymph node surgery, elevated body mass index, and regional lymph node radiation [2]. Lymphedema affects 21% of women treated for breast cancer [3], and these patients experience a variety of symptoms including heaviness, fullness, and discomfort in the affected limb [4], [5]. They also suffer from lower quality of life than those without lymphedema [6].

Compression therapy is the mainstay of modern lymphedema management, as preliminary studies suggest short-term compression can be beneficial [7], [8], [9], [10]. Compression sleeves reduce swelling by providing a gradient

pressure that decreases from the wrist to the upper arm. Physiologically, they reduce edema by increasing interstitial pressure and tissue fluid drainage, stimulating lymphatic contractions and breaking down fibrosclerotic tissue caused by chronic edema [11]. They are prescribed as either USA Class I or Class II, with target pressure gradients of 20–30 mmHg or 30–40 mmHg, respectively. These pressures have been shown to counteract hydrostatic venous pressure in the arm in the upright position without impeding the lymphatic pump [12]. The garments are available in multiple sizes and lengths, which aim to impart the prescribed pressure on varying limb sizes and shapes. Patients are fitted by a trained healthcare professional, and instructed to wear the garment for at least 12 hours per day. However, patients may not adhere to this prescription since sleeves may be perceived as aesthetically displeasing or uncomfortable. Unfortunately, studies analyzing the effect of compression do not typically report adherence data regarding garment-wearing [7], or may rely on patients' self-reports [10]. In some studies, reported adherence has been poor; for example, one study found that only 39–41% of patients reported wearing their compression garments at least 75% of the prescribed time [13]. In addition, the pressure achieved by a sleeve may not match the expected range.

It is imperative that wear time and pressure are objectively monitored within research studies, to determine the true compression dose received and to determine effective treatment parameters. For example, if the standard prescription of wearing sleeves for 12 hours per day could be reduced, this would reduce treatment burden and improve quality of life. This work moves towards this goal by using soft sensors to continuously monitor exerted pressure and usage duration.

In particular, this work includes the following contributions:

- A soft pouch-based pneumatic pressure sensor suitable for long-term wearable monitoring and rapid fabrication;
- Comparisons of the pneumatic sensor with a commercially available resistive sensor, including design considerations, fabrication, and performance;
- Systematic evaluations of the sensors including sensitivity to ambient temperature or pressure;
- Characterization of the sensors and their expected accuracy, with an emphasis on reducing required calibration;
- Integration of the sensors into compression sleeves to continuously monitor pressure and wear duration;
- Preliminary deployments of the wearable monitoring system, spanning 10 days with over 115 hours of usage.

The remainder of this paper discusses related sensing approaches in Section II, sensor fabrication and sensitivity experiments in Section III, sensor characterization and accuracy studies in Section IV, sleeve integration and deployments in Section V, and conclusions in Sections VI and VII.

II. Related Work

Due to their great potential to enhance capabilities of soft robots and wearable devices, there has been substantial efforts to develop flexible pressure sensors. These explore a variety of operating principles including piezoresistive, capacitive, or ionic approaches [14], [15]. Studies moving towards wearable pressure sensing have often focused on material and fabrication advancements, such as using nanowires [16], conductive fibers for capacitive sensing [17], piezocapacitance [18], hydrogels for resistive sensing [19], disposable paper [20], or customizable knitting [21]. Such investigations tend to focus on providing sensitivity to small forces, fast response times, and health applications such as pulse detection. However, there remain significant challenges for increasing flexibility, repeatability, and ease of fabrication [22]. Common piezoresistive sensors also have open questions such as understanding how design parameters impact performance and how to model or mitigate hysteresis [23].

Moving towards addressing such challenges, alternative approaches have focused on pneumatic principles. These include small tubing within knitted wearables [24] or robotic sensors [25], and larger air chambers [26]. A variety of pneumatic and electrical approaches have also been used to create active wearable devices with both sensing and actuation for assistance or therapy [27], but remaining sensor challenges include reliability, accuracy, fabrication, and portability. As demonstrated in [28], commodity materials can be used to create inexpensive pneumatic sensors for robot grippers that simplify fabrication and exhibit a linear response with low drift. The current work builds on such approaches to develop a pneumatic pouch-based sensor for integration into a compression sleeve, and evaluates its accuracy, reliability, robustness, and sensitivity to ambient conditions as compared to resistive sensing in the context of continuous wearable monitoring for lymphedema treatment.

Although compression therapy is the state of the art for lymphedema management, there have been limited quantitative studies to evaluate the required dosage and improve effectiveness. As explored in [29], this is due in part to the difficulty of sensorization; the pressure achieved by the garment is often unknown, and an initial investigation revealed that it was typically lower than the target. Additionally, [29] tested a selection of sensing modalities and observed highly variable performance, although pneumatic and resistive approaches had the highest precision; conclusions included the importance of calibration and of using in-vivo tests rather than rigid mock limbs. These sensors are typically not designed for continuous monitoring, and thus cannot address questions of changing pressure or adherence to prescribed durations.

This work aims to leverage advances in soft sensing to design and deploy an integrated wearable system for continuous pressure monitoring that can provide vital information about lymphedema treatment. It focuses on enabling long-term, unobtrusive, nearly plug-and-play sensing of whether the sleeve is being worn and what pressure gradient is being exerting on the arm. This can lead to deployments that improve patient care by yielding insights and verifications about effective compression sleeve prescriptions. The experiment-based sensor design and characterization results can also benefit future studies that aim to use soft pneumatic or resistive pressure sensing for robotics or wearables.

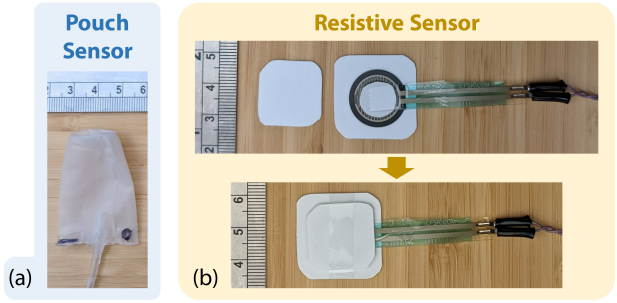


Fig. 2: Soft vinyl bladders form pneumatic sensors (a), and a plastic sheath guides pressure on resistive sensors (b).

III. Sensor Design and Fabrication

There are a few key metrics to consider when designing and fabricating sensors for the compression sleeve. Their *accuracy and repeatability* over time must be sufficient to measure exerted pressure and usage duration for lymphedema treatment evaluation. Based on consultations with expert clinicians, pressure should be measured to within approximately ± 2 mmHg and usage duration should be estimated to within approximately 15 minutes over a 12-hour period. Performance should also be reliable despite *varying environmental conditions* such as ambient temperature or pressure, and despite *operating at a soft and dynamic interface* between the human arm and the stretchable sleeve. Sensors should also be *robust* to daily wear including common activities, and be *unobtrusive* such that they do not cause the user any discomfort or adverse effects. Finally, a *scalable and customizable fabrication approach* should streamline sleeve integration and future widespread deployments.

Two sensing modalities were investigated based on these criteria. Firstly, a custom pneumatic sensor uses a small soft bladder connected in a closed system to a pressure transducer. Secondly, a commercially available force sensitive resistor (FSR) is fastened between plastic layers that guide pressure distribution. The fabrication methodologies, design considerations, and sensitivity experiments are discussed in the following sections. Additional details including a list of materials are also available online.[1]

A. Pouch Sensors

The pouch sensor, shown in Figure 2a, consists of an air bladder connected to a pressure transducer. It is fabricated using an index or middle finger of a large vinyl powder-free medical examination glove from MedPride. Flexible silicone tubing with a 1.5 mm inner diameter, a 3 mm outer diameter, and a durometer of 50A is inserted approximately 1 cm into the pouch. The interface is sealed using Smooth-On Sil-Poxy.

The current implementation uses a D2-P4V-Mini pressure transducer from All Sensors. Its operating pressure range is ± 30 inches of water (± 56 mmHg), and it outputs an analog voltage proportional to the difference in pressure between two input ports. The pouch tubing is connected to the positive port, and the negative port can be left open or sealed.

This approach uses a simple fabrication method and commodity materials to yield a soft robust pressure sensor with linear response. Its softness is important for user comfort and safety, especially in high-pressure and long-duration applications. It can also be rapidly prototyped and customized. Key design considerations include the pouch size, how much it is inflated, and whether to use an open or closed reference port. A smaller and less inflated pouch is desirable since it minimizes obtrusiveness for the person wearing the sleeve. However, since the user's arm is deformable, a pouch that is too small may sink into an indentation of the skin and yield inaccurate pressure estimations of the sleeve passing on top of it. Under-inflating the pouch may also allow it to be flattened by the sleeve, blocking the inlet of the tube inside the pouch. Section III-C will further explore such parameters.

B. Resistive Sensors

A commercially available FSR was used to explore resistive pressure sensing, namely the circular FSR 402 from Interlink Electronics. It is 0.6 mm thick, has a 1.8 cm diameter, and its active area has a 1.3 cm diameter. Its nominal sensitivity range is 0.1 – 10.0 N (5.7 – 565 mmHg if force is distributed over the entire active area). Connecting the sensor in a voltage divider yields a voltage that depends on applied pressure.

The sensor's thin profile and small surface area allow it to be unobtrusive, but also allow it to sink into indentations of the skin and yield inaccurate results. Furthermore, uniformly stretching the sleeve over the sensor may causes its inactive edge to support much of the pressure and preclude successful measurements. To address these challenges, the sensor is fastened within a custom plastic sheath as shown in Figure 2b. White flexible plastic polystyrene sheets with a thickness of 0.5 mm were used. The FSR is glued to a 2.5 cm square to avoid sinking into the arm. A circle with a diameter of 1.1 cm is then centered over the active area, lightly taped on one side to prevent movement while not applying pressure. Finally, a 2.0 cm square is lightly taped on top. This sheath provides a larger surface area for the sleeve interface, and focuses exerted pressure onto the sensor's active area.

This approach uses commercially available sensors and materials to create a thin, unobtrusive sensing layer. However, the need for a semi-rigid plastic sheath introduces edges on the skin that may be undesirable for long-term usage. FSRs also exhibit nonlinear responses, hysteretic effects, and variations between sensors that may limit accuracy and repeatability or require calibration. These will be explored further below.

C. Sensitivity to Ambient Conditions

An important goal for the deployable wearable system is to maintain medically relevant accuracy while reducing how much calibration is required. This is coupled with reducing sensitivity to ambient conditions; for example, if the sleeve is fitted in an air-conditioned doctor's office at sea level, and then the patient travels to an elevated city on a hot day, the sensor would ideally continue to generate actionable data without requiring the user to calibrate. This goal can also help inform sensor design parameters.

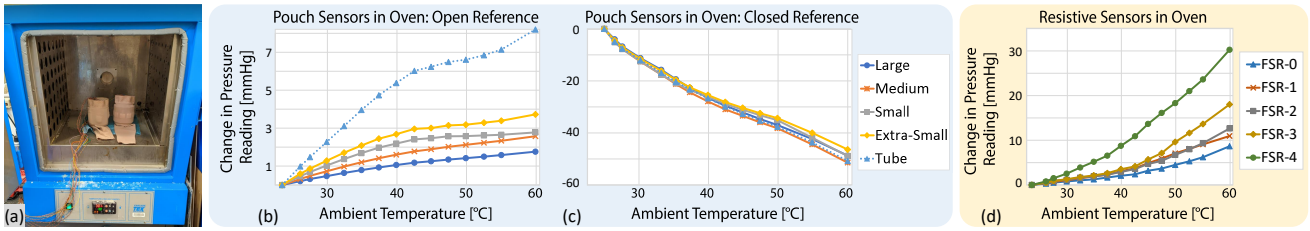


Fig. 3: Sensors placed on foam-wrapped cylinders under a compression sleeve are placed in an oven (a) to study ambient temperature sensitivity of pouch sensors with open (b) or closed (c) reference ports and of resistive sensors (d).

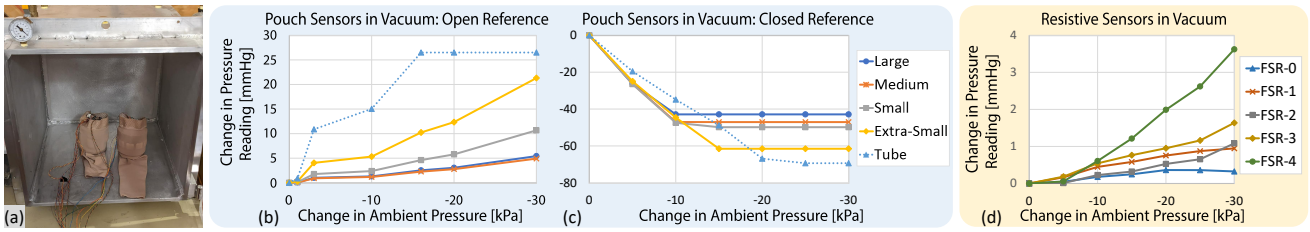


Fig. 4: Sensors placed on foam-wrapped cylinders under a compression sleeve are placed in a vacuum (a) to study ambient pressure sensitivity of pouch sensors with open (b) or closed (c) reference ports and of resistive sensors (d).

1) Experimental Methodology: Sensors were placed in an oven and a vacuum chamber to simulate varying ambient temperatures and pressures, as shown in Figures 3a and 4a. Mock arms were created by wrapping medium-density foam with a thickness of 0.5 cm around metal cylinders with a diameter of 9 cm. Five pneumatic or resistive sensors were placed around each cylinder, and Class I Juzo sleeves were stretched on top. A size VI sleeve exerted approximately 9 mmHg on the pneumatic sensors, and a size V sleeve exerted approximately 12 mmHg on the resistive sensors.

During the oven experiments, temperature was steadily increased from 23.5° C to 60° C at approximately 2.5° C increments. At each point, temperature and sensor readings were allowed to settle and then a 1-minute average of sensor readings was recorded. During the vacuum experiments, pressure was gradually reduced from sea level to -30 kPa (-225 mmHg) and readings were allowed to settle at each point before recording a 1-minute average.

These were performed twice for the pouch sensors: once with the transducers' reference ports open to the atmosphere, and once with them closed. In the closed case, each reference was connected to a 1.5 cm tube that was sealed with Sil-Poxy.

The five pouch sensors were fabricated with varying sizes to study how size impacts sensitivity. Four were created using vinyl gloves as described above. Their base width was 3 cm, and their active lengths were 6 cm, 4 cm, 2 cm, and 1 cm. A fifth pouch was created using a tubular latex balloon; its diameter was 0.75 cm and its active length was 6 cm. These sensors are referenced below as large, medium, small, extra-small, and tube sensors, respectively. Each pouch was sealed to a tube that extended 4 cm from its base. Each pouch was filled with air such that it would create a bulge under the sleeve of approximately 4 mm; this helps unify how the sleeve exerts pressure on each sensor, and standardizes the eventual user experience. This resulted in volumes of approximately 7 mL, 3.5 mL, 2 mL, 1 mL, and 1 mL, respectively.

2) Results and Impact on Design: Figures 3 and 4 present the results. Raw sensor readings were converted to pressure values using transfer functions fit to all pneumatic sensors or all resistive sensors as described below in Section IV.

Results suggest that larger pouches are less sensitive to changes in ambient temperature and pressure. The relationship between size and temperature sensitivity can be interpreted by first considering the ideal gas law: $PV \propto T$ where P, V and T are pressure, volume, and temperature, respectively. The amount of pressure change required to compensate for a change in temperature thus depends on how much the volume will change in response to that pressure change. This in turn can be considered within the context of an elastic material. Modeling the pouch as a thin-walled cylinder, the strain ϵ of the material is given by $\epsilon = Pr_0(\frac{1}{Et})$ where E, t, and r_0 are the Young's modulus, thickness, and initial radius, respectively [30]. A given change in pressure thus causes a larger change in strain if the radius is larger; this implies that changing the pouch volume requires less pressure change for larger pouches. Thus, when the temperature changes, larger pouches can change their volume more readily and require smaller changes in pressure. Similar reasoning can be applied to the vacuum tests, where a readily changeable volume allows the pouch pressure to decrease and more closely match the decreasing ambient pressure. Note that the nonlinearities at higher temperatures and lower ambient pressures may be related to nonlinear material properties in those regimes, and the saturation during vacuum tests may be due to leakages.

Results also suggest that using an open reference is less sensitive than a closed reference. In the closed case, the sealed semi-rigid tube cannot readily change volume. During oven tests, its pressure thus increases faster than the pressure in the pouches as temperature increases; this yields increasingly negative differential pressures. During vacuum tests, its internal pressure remains essentially constant, so the differential pressures decrease as pouch pressures decrease.

Based on these experiments, larger pouches with an open reference port are desirable to minimize ambient sensitivity. Balancing this with the desire to reduce obtrusiveness by minimizing size, a medium pouch with an open reference port was selected. This configuration exhibits an average change in detected pressure of approximately 1.4 mmHg at 37.4° C (near body temperature or a hot day) and 2.3 mmHg at -16 kPa (traveling from sea level to an elevation of approximately 1,500 m). These results are promising for not requiring users to recalibrate as ambient conditions vary.

Most of the FSRs exhibit comparable temperature sensitivity as the pouch sensors within typical weather conditions, but are significantly more sensitive to higher temperatures. This may be due in part to the properties of the resistive material, and to the nonlinear relationship between resistance and pressure discussed below in Section IV. Regarding the vacuum tests, the FSRs generally exhibited low sensitivity to ambient pressure as expected. The observed increase in measured pressure may be partly due to the foam around the metal cylinder expanding as ambient pressure drops.

There were also notable variations among the five FSRs, especially at higher changes in temperature or pressure; this may be due to inherent variability between the sensors, or to factors such as sheath fabrication and mounting on the cylinders. Sensor variability will be explored further for resistive and pneumatic sensors in Section IV.

IV. Characterization and Calibration: Experiments and Results

Important considerations are how accurately each sensor type can estimate pressure values under a compression sleeve, and how much calibration will be needed for each instance of a smart sleeve. Ideally, the sleeve will be plug-and-play: a new user could don a new sleeve and immediately begin receiving personalized treatment feedback. To address these aspects, the response curve of each sensor is characterized to convert voltage readings to pressure values and infer the expected accuracy during real-world deployments.

A. Methodology

Experiments were performed to determine an appropriate parameterization for the response curve of each modality, and to investigate generalizability of the parameters between sensor instances. To do this, 5 instances of each sensor type were fabricated. For pouch sensors, a medium size with 4 mL of air was selected based on the ambient sensitivity study.

A testing rig was constructed as shown in Figure 5a to systematically collect data that maps voltages to ground-truth pressures. Key considerations are to be as close to real-world conditions as possible, including how pressure is exerted and the deformability of materials, and to enable systematic pressure variation. The presented setup aims to mimic a variable-diameter arm wearing a compression sleeve. A rigid cardboard poster tube with diameter 5.5 cm was wrapped in medium-density foam and cut in half. The bottom of each half is hinged to a base, and the tops can be spread apart using a clamp. A sensor being evaluated is lightly taped to one half, and a Juzo Pressure Monitor is inserted into the other half; the latter is used to measure sleeves during clinical visits, and can serve as the ground-truth measurement. A Class I size II Juzo compression sleeve was stretched around the device and sensors. As the rig is spread, the sleeve stretches and exerts increasing pressure on both the sensor and the Juzo monitor.

Data is streamed and synchronized from sensors and a webcam using the `ActionSense` multimodal recording framework [31]. Sensors are powered by $V_{regulated} = 3.3$ V, which is measured along with the data. Sensor readings are acquired using the 16-bit ADS1115 analog-to-digital converter (ADC). Voltage readings are normalized by $V_{regulated}$ in below analyses, to provide robustness to variable supplies.

Given the target pressure ranges of compression therapy, the rig was spread to exert pressures between 10 and 50 mmHg at intervals of 2 mmHg. At each one, the sensor readings were allowed to settle and then a 15-second average was recorded.

B. Pouch Sensors

Results for the pouch sensors are shown in Figure 5b. As expected, the voltage response is linear with respect to pressure. A linear curve is thus fit to the data of the form

$$Pressure = \left(\frac{V_{sensor}}{V_{regulated}}\right) m + b \tag{1}$$

where m and b are parameters optimized via the `curve_fit` function of the SciPy Python package. This is first done for each sensor individually. The average error of these five fits was 0.8 ± 0.2 mmHg ($2.9\% \pm 0.9\%$) across all tested pressures. This is acceptably low for the target application.

However, the goal is to avoid per-sensor calibration, so two analyses were performed to investigate the expected accuracy of a new sleeve. Firstly, the voltages of one sensor are converted to pressure using parameters fit to another sensor. Each possible sensor pair was tested, and Figure 6a shows the resulting error matrix. Across all pairs, the expected error would be 1.8 ± 0.7 mmHg ($6.8\% \pm 2.8\%$). Secondly, voltages of one sensor are converted using parameters fit to aggregated data from all other sensors – a more robust curve fit to multiple sensors is used for the new sensor. In this case, expected error improves to 1.5 ± 0.4 mmHg ($6.0\% \pm 1.4\%$). These results suggest that pouch sensors exhibit acceptable accuracy and repeatability across sensors, and may be deployable as plug-and-play devices without per-sleeve calibration.

To increase accuracy further, since characterization errors will compound with errors induced by ambient conditions, a minimal calibration procedure is investigated that assumes a patient has their sleeve measured once at a clinical visit. The current analysis implements this paradigm by fitting a curve to all sensors except one, then adjusting the curve for the left-out sensor based on a single calibration point. It simulates cases where the calibration measurement is 15 different pressures evenly spaced between 16 and 44 mmHg, to account for differing sleeve fits across patients. For each case, it computes a new b using Equation 1 and an m optimized to all other sensors. It then evaluates the adjusted curve at pressures within ± 6 mmHg of the calibrated pressure, since the pressure of a

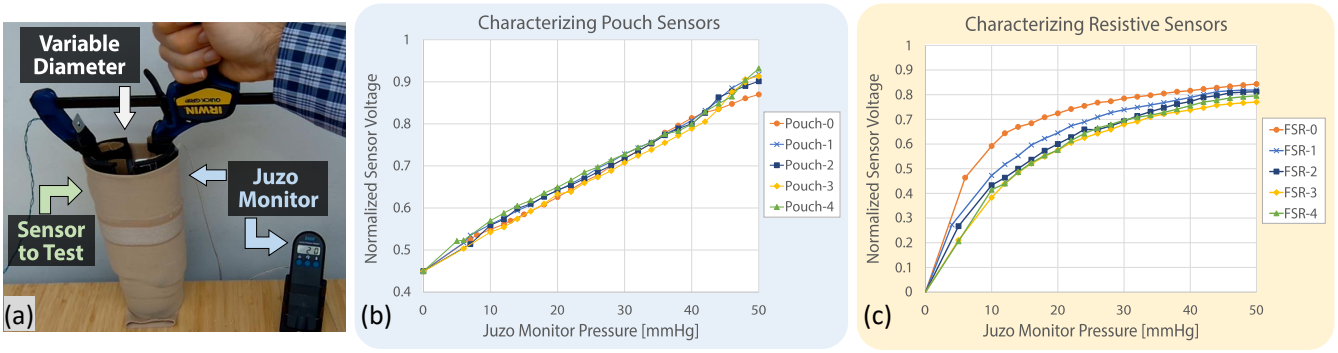
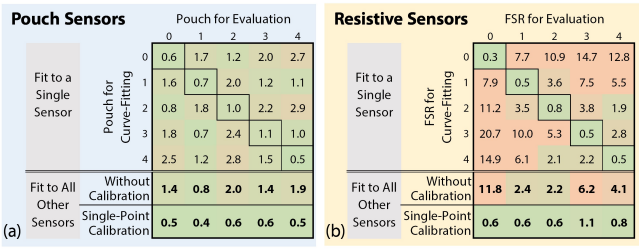


Fig. 5: A variable-diameter mock arm (a) was used to vary the exerted pressure on both a test sensor and a ground-truth Juzo monitor. Pouch sensors exhibited a linear response (b) while FSRs exhibited a nonlinear response (c).



Pouch Sensors

		Pouch for Evaluation 0	1	2	3	4
Fit to a Single Sensor	Pouch for Curve-Fitting 0	0.6	1.7	1.2	2.0	2.7
	1	1.6	0.7	2.0	1.2	1.1
	2	0.8	1.8	1.0	2.2	2.9
	3	1.8	0.7	2.4	1.1	1.0
	4	2.5	1.2	2.8	1.5	0.5
Fit to All Other Sensors	Without Calibration	**1.4**	**0.8**	**2.0**	**1.4**	**1.9**
	Single-Point Calibration	**0.5**	**0.4**	**0.6**	**0.6**	**0.5**

(a)

Resistive Sensors

		FSR for Evaluation 0	1	2	3	4
Fit to a Single Sensor	FSR for Curve-Fitting 0	0.3	7.7	10.9	14.7	12.8
	1	7.9	0.5	3.6	7.5	5.5
	2	11.2	3.5	0.8	3.8	1.9
	3	20.7	10.0	5.3	0.5	2.8
	4	14.9	6.1	2.1	2.2	0.5
Fit to All Other Sensors	Without Calibration	**11.8**	**2.4**	**2.2**	**6.2**	**4.1**
	Single-Point Calibration	**0.6**	**0.6**	**0.6**	**1.1**	**0.8**

(b)

Fig. 6: Errors of fitted curves in mmHg for pouches (a) and FSRs (b) are summarized for using parameters from another sensor or multiple sensors with or without brief calibration.

deployed sleeve is expected to remain relatively consistent for a given subject between office visits. Using this single-point calibration paradigm, the expected error was 0.5 ± 0.1 mmHg ($1.8\% \pm 0.3\%$). This is well within desired specifications, and is promising for practical deployments.

C. Resistive Sensors

The same data collection methodology was also used for the resistive sensors; results are shown in Figure 5c. A curve function was then derived that aims to fit the nonlinear response while keeping the number of parameters low.

An FSR is expected to exhibit an exponential mapping between applied pressure and resistance of the form

$$R_{sensor} = (a)\,(Pressure)^{(-b)} \tag{2}$$

as described in [32], where pressure has been substituted for force since the constant factor of area can be encapsulated within the optimized a and b. This resistance is then converted to voltage via a voltage divider with a fixed resistor. The sensed voltage at their junction is given by

$$V_{sensor} = (V_{regulated}) \left(\frac{R_{divider}}{R_{divider} + R_{sensor}} \right) \tag{3}$$

where $R_{divider}$ is selected to maximize the range of measured voltages around the expected operating point. Currently, a 5.6 kΩ resistor is used. Combining Equations 2 and 3 yields

$$Pressure = \left[\frac{(a)\left(\frac{V_{sensor}}{V_{regulated}}\right)}{R_{divider}\left(1 - \frac{V_{sensor}}{V_{regulated}}\right)} \right]^{\left(\frac{1}{b}\right)} \tag{4}$$

which provides a well-founded curve to approximate the sensor response while only using two tunable parameters.

The same curve-fitting and evaluation methodology as described for the pouch sensors was then applied, and results are shown in Figure 6b. Fitting to each individual sensor yields an average error of 0.5 ± 0.2 mmHg ($2.0\% \pm 0.7\%$), fitting to a single previous sensor without calibration yields an average error of 7.8 ± 5.0 mmHg ($28.2\% \pm 19.5\%$), and fitting to all previous sensors yields an average error of 5.4 ± 3.5 mmHg ($19.8\% \pm 15.2\%$). To adjust a curve using the single-point calibration paradigm, the analysis explored computing a new a or b as well as adding an offset to shift the curve. Adding an offset was found to be most accurate, and yielded an average error of 0.7 ± 0.2 mmHg ($2.6\% \pm 0.7\%$).

Results suggest that using FSRs without calibration may not be sufficiently accurate for the current application. Variability may be due to the sensors, or to factors such as sensitivity to placement on the cylinders. Using a single-point calibration yields accuracies that are comparable to the pouch sensors.

V. System Integration and Deployment

Using insights from the sensor evaluations, deployable smart sleeves were fabricated and worn for extended periods of time to demonstrate the feasibility of using the sensors for continuously monitoring pressure and usage duration.

A. Sensor Integration

Patches of thin fabric were affixed to the inside of compression sleeves at the forearm and upper arm, to create sensor pockets as shown in Figure 7a-b. These hold sensors in consistent positions, prevent sensors from contacting the user's arm directly, and allow sensors to be removable. Sensor wires were left outside the sleeve, guided by the user's shirt; future iterations may sew conductive threads into the sleeve.

B. Electronics and Software

A small wearable control board was created based around the Adafruit QT Py ESP32-C3 microcontroller, which features WiFi and Bluetooth. For improved accuracy, an ADS1115 ADC measures the sensor outputs and regulated voltage. A 500 mAh single-cell lithium-ion battery powers the circuits. These components, along with a charging circuit and FSR voltage dividers, are mounted on a 3 cm x 4 cm breadboard. A

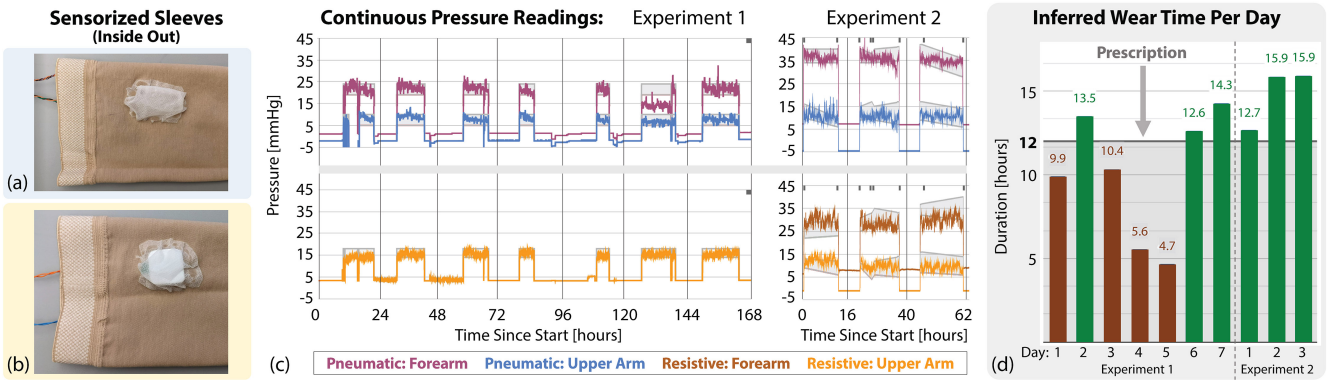


Fig. 7: To deploy an integrated wearable system, sleeves were augmented with pouch sensors (a) and resistive sensors (b). Sensors streamed pressure data from the forearm and upper arm during two experiments (c). Shaded regions interpolate ground truth measured at times indicated by black rectangles. The system also monitors adherence to usage prescriptions (d).

velcro strap is attached so it can be worn on a belt as shown in Figure 1. P4V pressure transducers are mounted on the sleeve near each pouch; note that future deployments could use a smaller transducer to reduce obtrusiveness.

The microcontroller samples each sensor at 1 Hz. Every minute, it computes the mean and standard deviation of the ADC values. It then uploads these statistics to Google Sheets via WiFi, where a real-time dashboard provides pressure and usage information. The current deployment assumes constant internet connectivity, achieved via a mobile hotspot.

C. Experiments and Results

Two experiments were conducted using the above integration methodology. The first experiment used a single Class I size II sleeve worn on the left arm. It had a resistive sensor on the middle of the outer forearm, a pouch sensor on the outer forearm near the wrist, and a pouch sensor on the outer upper arm. It was worn for 71 hours over 7 days. The second experiment used two Class II sleeves as shown in Figure 1, each with a sensor on the outer forearm near the wrist and on the outer upper arm. A size II sleeve worn on the left arm had pouch sensors, and a size I sleeve worn on the right arm had resistive sensors. These were worn for 45 hours over 3 days. A single able-bodied right-handed subject participated in these preliminary tests of feasibility.

1) User Experience: The sleeve systems successfully demonstrated continuous monitoring of both pressure and usage. The sensors were qualitatively unobtrusive, not causing noticeable discomfort beyond what the sleeve alone induces. The sensors produced reliable data throughout a variety of activities of daily living including working at desks, cooking, eating, walking, and commuting a few miles per day using a kick scooter. The experiments also spanned multiple ambient temperatures; indoor temperatures were regulated, while outdoor temperatures averaged 21.3° C during the first experiment (range 17.2–26.1° C) and 10.7° C during the second experiment (range 5.0–20.0° C). The battery lasted a full day without recharging; this lifetime could also be substantially extended by reducing the frequency with which WiFi is used and by using low-power microcontroller modes.

2) Monitoring Pressure: Figure 7c shows the inferred pressures from each sensor. The Juzo Pressure Monitor was also inserted under the sensors to estimate ground truth at the end of the first experiment and throughout the second experiment. Minimum and maximum values were recorded while varying arm pose, and used to generate the shaded regions in Figure 7c. These bounds were averaged to estimate ground truth. One such estimate in each experiment was used to perform a single-point calibration for each sensor.

During the second experiment, which included more frequent ground-truth measurements, calibrated pouch sensors had an average error of 1.6 ± 1.4 mmHg while the sleeve was worn (1.7 ± 1.4 mmHg and 1.6 ± 1.5 mmHg on the upper and lower arm, respectively). 89.4% of the readings were within the shaded ground-truth regions. For calibrated resistive sensors, errors averaged 2.8 ± 2.1 mmHg (1.8 ± 1.3 mmHg and 3.9 ± 2.2 mmHg on the upper and lower arm, respectively), and 64.2% of readings were within the ground-truth regions.

Results suggest that the pouch sensors generally achieved the accuracy goals for evaluating compression therapy, and that they yielded lower errors than the resistive sensors ($p < 0.001$ using a one-sided t-test). Both sensors successfully reflect a pressure gradient between the forearm and upper arm that the sleeves are designed to exert. The observed variability in Juzo measurements also lends credence to the variability of sensor readings throughout each day.

3) Monitoring Adherence to Prescribed Usage Duration: Based on the pressure measurements, a threshold can be used to determine whether the sleeve is being worn. Figure 7d shows the results, illustrating that the recommended dosage was achieved on 60% of days. The current system is capable of measuring adherence with approximately 1-minute resolution.

VI. Discussion

Both sensing modalities explored in this work demonstrate promise for continuously monitoring pressure and usage duration. They each have trade-offs regarding accuracy, integration within the sleeve, and robustness.

The pouch sensors exhibit a linear response, yield accurate results even without calibration, and can be rapidly fabricated

and customized using commodity materials. Importantly, they are also soft and thus well-suited to wearable applications. However, the transducer or tubing must be mounted on the sleeve which may slightly increase obtrusiveness, and the transducer requires an outlet to the atmosphere which may complicate encapsulation for protection and waterproofing. Popping and leakages are also concerns.

The resistive sensors can yield accurate results, do not require a separate transducer mounted on the sleeve, and are generally physically robust. However, they require semi-rigid sheathing to appropriately distribute pressure onto the sensor, which introduces edges on the person's skin under high pressures. They also exhibit higher sensitivity to ambient temperature, and characterization results suggest that at least a single-point calibration routine is required to obtain sufficient accuracy. The response curves can also be highly sensitive to where pressure is exerted on the sensor, so may change over time due to the sheath moving or to the sensor properties.

All together, the trade-offs and experimental results indicate that the pouch sensors are promising for wearable applications including compression therapy assessment. They exhibit comparable or improved performance compared to resistive sensors, minimizing calibration routines while achieving medically relevant pressure accuracies and usage duration resolutions. They are also soft and thus safer for prolonged human contact. Further work can improve robustness, and miniaturize and protect the transducer.

VII. Conclusion and Future Work

This paper describes a custom soft pneumatic sensor, a method for integrating it into textiles, and systematic evaluations of this sensor and a commercially available resistive sensor. Experiments explore sensitivity to ambient conditions, characterization, expected accuracy, and required calibration. The pouch sensors have no sharp edges, which is important for long-term wearing. Preliminary deployments are promising for continuous pressure and usage monitoring.

Future work will include longer deployments with a larger subject pool and close collaboration with clinical staff. Additional characterizations can also be performed with more instances of each sensor, and additional design parameters such as materials can be explored. Future work can also improve fabrication and integration, such as sewing conductive threads into the sleeve and extending battery life.

This work takes a step towards new insights for lymphedema treatment, and towards smart wearables that provide doctors and patients with real-time personalized feedback.

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