Yale Poorvu Center

Department of Mechanical Engineering

Anjali Agrawala

How I Wrote My Prospectus

I started formulating my prospectus by identifying the key advantages of the technology I had already been developing. Seeking to leverage those specific advantages and expand on my prior research project, I brainstormed potential applications for the technology and kept notes on possible research pipelines to reach the end goal application. With these initial ideas, I then completed an extensive literature review focusing on 1) technologies similar to the one I was working on and the different applications in which they are employed, 2) current challenges in my chosen application space of interest, and 3) technologies usually used within my application space of interest. By organizing and consolidating this information, I was able to identify a potential use case for my technology within my application of interest, assess the limitations of the technology within this application, and begin developing a hypothesis. Given that my project was interdisciplinary, I also built connections with technical experts in another department and the School of Medicine to gain further insights and feedback on the feasibility of and requirements for accomplishing my proposed project. From this point, it was helpful to divide my planned research into objectives or tasks, all of which help collectively build towards addressing the proposed hypothesis. Doing so included predicting potential challenges or pitfalls and planning to address them. In designing tasks and experiments, I also tried to guide the project such that I would obtain specific skills and/or knowledge I was interested in. As I started converting this prospectus outline into paragraph form, I also made sure to obtain approval from my advisor on my plan and consulted with senior graduate students and/or postdocs for further feedback. Although I didn't end up having enough time for it, I would recommend saving some time to get feedback on the writing draft itself.

Advice for Prospectus Writers

Writing your prospectus is a great opportunity to take some time to plan out what you want to work on for your dissertation. For me, developing your project idea and planning can often be the most stressful part of research. By writing the prospectus you can gain some structure and clarity over your research direction early such that you can successfully set yourself up to plan and complete experiments, coursework or any other training you may need within a reasonable timeline. Additionally, since your prospectus will be your original, intellectual work and property, it serves as an opportunity to formally establish your project ideas and develop them. By setting these expectations for research and essentially creating an agreement with your advisor and your department, you can help prevent potential conflicts of interest and set expectations for your scientific contributions upon graduation. Keep in mind that your prospectus is only a proposed plan of research. It is expected to evolve and will function as a guideline or framework to help break down your dissertation into smaller, completable tasks. It is more important to be on the same page as your advisor than it is to stick strictly to your prospectus. However, your prospectus can also act a bit as a contract in the sense that it can help you and your advisor assess and agree on when you are ready and expected to give your thesis defense.

Detection of REM Sleep Behavior Disorder using Wearable Textile Sensors

A Prospectus
Presented to the Faculty of the Graduate School
of
Yale University
in Candidacy for the Degree of
Doctor of Philosophy

by
Anjali Agrawala

Research Advisor: Professor Rebecca Kramer-Bottiglio
Committee: Prof. Corey O'Hern, Prof. John Onofrey, and Prof. Madhu Venkadesan
Area Exam: 29 November 2022. Planned Final Examination: July 2026.

Contents

Abstract

Rapid Eye Movement Sleep Behavior Disorder (RBD) is a powerful and early indicator of Parkinson's Disease, but current diagnosis requires an expensive and labor-intensive clinical sleep study. In recent years, there has been a growing interest in the use of alternative technologies to unobtrusively measure nocturnal movement during sleep. The growing field of wearable sensing has led to the development of soft stretchable sensors, which show promise for this application. When these sensors undergo uniaxial stretching, the electrical output of the sensor changes, thus allowing them to be attached to human joints to describe their bending. However, more work is needed to create implementable strain sensors that prioritize wearability. Here, I introduce a new textile-based capacitive sensor that can be seamlessly integrated into commercial activewear and is designed for user comfort. I characterized the performance of our sensor including its breathability, strain response, cyclic stability, and electromechanical response to environmental conditions and washing. Real-time data collection from sensors placed on the elbows, hips, and knees is also demonstrated. I discuss ongoing challenges with implementation of the sensor and how I plan to mature the technology. Finally, I propose to use this new sensor technology in a human subject sleep study to investigate its utility in providing an objective method of RBD detection.

1 Introduction: Sleep-based Predictor of Parkinson's Disease

PD is a slow progressing and age-related neurodegenerative disease, with an increasing rate of incidence, that is projected to affect over 14 million people worldwide by 2040[1]. Well-established hallmark motor symptoms of PD such as resting tremor, bradykinesia (slowness of movements), rigidity, and postural instability[2,3,4] typically appear in advanced stages of disease progression[1]. Since standard diagnosis of PD is still dependent on clinical evaluations, including an interview, physical exam, and review of medical history[4], diagnosis relies on these visually identifiable symptoms. However, such motor symptoms and clinical diagnosis are preceeded by years of known prodromal symptoms including nonmotor symptoms such as hyposmia, depression, and rapid eye movement (REM) sleep behavior disorder (RBD)[1]. RBD is characterized by loss of normal muscle atonia during REM sleep. On average, RBD has been shown to precede clinical diagnosis of neurodegenerative diseases by over 10 years[5], with a 2017 study demonstrating that 74% of RBD patients showed probable prodromal PD compared to 0.3% of controls using the criteria set by the Movement Disorder Society[6]. Thus, RBD has potential as a powerful predictor of PD. The loss of muscle atonia during REM sleep in RBD patients leads to dream enactment, typically, in the form of purposeful, complex motor behaviors[7]. In my thesis, I plan to address the need for an unobtrusive, convenient method of objectively analyzing abnormal movements during sleep to identify RBD by developing a wearable sensor for tracking joint movement during sleep.

1.1 Sleep Studies: State-of-the-art

Although polysomnography is the gold standard in sleep disorder diagnosis, demand for a more convenient and easily repeatable test has led to more recent investigations into other methods of sleep behavior analysis. Here I summarize the capabilities and limitations of the gold standard test and the more recent technologies.

Polysomnography (PSG) involves multiple modes of measuring physiologic activity to comprehensively and continuously monitor sleep overnight. Sleep staging is determined with the use of electroencephalography (EEG) - which measures electrical brain activity, electrooculography (EOG) - which measures eye movement, and electromyography (EMG) - which measures muscle tone[8]. Standard PSGs will typically also include sensors to measure heart rhythm, respiration, and oxygen saturation and video monitoring to provide contextual information and assess complex motor behaviors[8]. RBD diagnosis specifically requires PSG, with the two criteria including evidence of REM sleep without muscle atonia and video-based confirmation of dream enactment[8].
Limitations. PSG is an expensive and labor-intensive test[9]. Due to these challenges with PSG, they are usually based on single night tests which may not accurately reflect a patient's normal sleep behavior. The accuracy of single-night PSG exams are limited due to the inter-night variability normal of sleep disorders, and/or the first-night effect in which patients experience a change to their normal sleep architecture as a result of adaptation to the obtrusive sensor equipment, unfamiliar environment, and knowledge of being video recorded[8,9,10]. Although portable PSG equipment is available for at-home use, its use requires high patient cooperation and can have lower accuracy relative to in lab tests[9].

Actigraphy is an alternative method of sleep study in which an accelerometer is worn on an individual's wrist like a watch. The actigraph outputs an analog voltage signal during movement, which is analyzed to obtain information about movement frequency, amplitude, and duration[9]. Typically, actigraphy data uses movement data as a surrogate for investigating sleep quality rather than examining nocturnal movements themselves[9]. However, the convenience, low cost, and ability to perform multiple night tests in the patient's normal environment has led to recent investigations in the use of actigraphy for assessing nocturnal movements.
Limitations. Of these recent studies, one has shown that using multiple actigraphs simultaneously by attaching them to different limbs may be necessary to circumvent the accuracy challenges of single actigraphy tests. Comparison of upper and lower limb actigraphy data has shown a tendency of single actigraphy studies to over- and under- estimate nocturnal movements in PD patients due to the tendency for aymmetric symptoms across the body[9]. In fact, although actigraphy results have been compared with PSG data in evaluating specific sleep parameters, the accuracy of actigraphy in measuring nocturnal movements has not been assessed[9]. Studies have also shown that actigraphy is limited in its ability to isolate limb movements from axial turns and therefore may not be capable of independently identifying either complex nocturnal limb movements or voluntary motor activity during stages of wakefulness[9]. Perhaps for related reasons, actigraphy has shown low sensitivity in identifying primary sleep disorders like RBD[9].

Inertial Measurement Units (IMUs) have recently started being explored for their application in detecting abnormal nocturnal movement. Like actigraphy, IMUs are a low-cost and less labor-intensive alternative to PSG. Single IMUs were placed on the waist, trunk, or abdomen in most studies[9]. Some studies also utilized multiple devices at a single time, placed on wrists, ankles, trunk, and/or abdomen, to create a body area network to help reconstruct the individual's body position[9]. These wearable devices, that thus far have only been used by a small number of authors, include on-board gyroscopes, accelerometers, and sometimes magnetometers that collect data about body motion and position. Specifically, authors used velocity, acceleration, and degree

of axial turns to investigate body rotations[9]. Some authors also studied body position and number of limb movements[9].
Limitations. IMUs present some technical challenges including potential magnetic interferences[9] and positional drift in continuous long-term measurements[11] such as those that would be required for a sleep study. The devices are also made of rigid components, which can challenge user comfort and become obtrusive when multiple units are used across the body. Thus far, no authors have measured the accuracy of IMUs in assessing abnormal nocturnal movements in PD[9] or used IMUs to investigate specific sleep disorders like RBD.

The use of sensors for detecting abnormal nocturnal movements during sleep is a growing area of interest within PD research. However, few studies using wearable sensing technologies such as actigraphy and IMUs focus on sleep disorders including RBD. My thesis will focus on filling the existing technology gaps by developing an alternative sensing method for unobtrusively identifying and quantifying nocturnal motor behaviors characteristic of RBD.

1.2 Soft Strain Sensing: State-of-the-art

Soft, stretchable sensors have been developed with the goal of creating comfortable and non-intrusive technology for tracking human motor activity. By leveraging the high elasticity and compliance of elastomeric materials, soft strain sensing has already shown promise for measuring human performance in personalized healthcare[12,13], sports[14], and human-machine interfacing[15]. In such soft sensors, elastomeric materials including Ecoflex[16-18], Dragon Skin[19-21], and PDMS[21-23] serve as insulating host or substrate materials while a variety of different conductive materials have been used to create electrodes including carbon black[24,25], carbon nanotubes[26], silver nanowires[27], graphene[25,28], liquid metals[29,30], and/or ionic fluids[31]. Despite the advantages these sensors offer, including the ability to conform to curved and irregularly shaped surfaces and withstand normal skin deformations of human joints (range of 40-55% strain[32-35]), there remain ongoing challenges in the field regarding breathability, measurement frequency, and integration. To address these issues with current soft strain sensors, I developed a new capacitive sensor technology. These issues will therefore be addressed in context throughout Section 2. In Section 3, I will briefly describe my plan to further the implementation of this sensor for real-time in-situ data collection. Lastly, Section 4 will outline my plan to investigate the use of this technology in monitoring motion during sleep, an application that soft strain sensing has not yet, to my knowledge, been used for.

2 Comfortable, Wearable Sensors for Tracking Joint Movement

2.1 Sensor Design and Fabrication

To enable long-term wearability, the design of the sensor focused on prioritizing user comfort. Therefore, the parallel plate capacitive sensor was composed of common fabric materials used in everyday clothing. The sensor electrodes were made from commercially available conductive fabric and three different textiles - cotton, polyester, and nylon - were used as dielectric materials to investigate the resulting sensing properties of each. The electrode fabric layers were bound to the dielectric fabric layers in a heat press using a breathable, thermoplastic adhesive film in a stacked-assembly method (refer to Fig. 1). Two configurations of sensors were made: a 3-layer and a 5-layer. The 5-layer sensors are designed with the external electrode encasing the internal electrode and separating dielectric layers, allowing the entire outside of the sensor to be grounded and effectively shielding the electrical response of the sensor from being affected by contact with human skin. The 5-layer sensor configuration is therefore better suited for wearable applications and will be the focus for the remainder of this section.

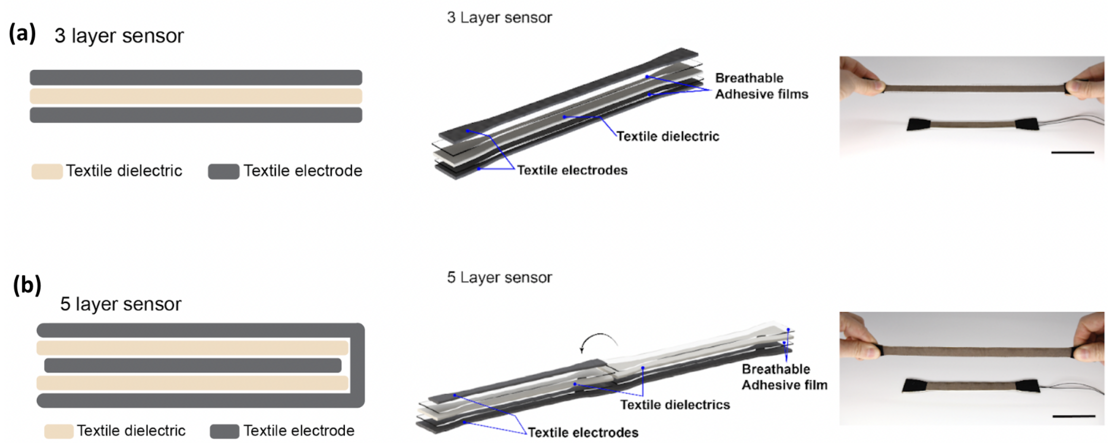

Figure 1. **Textile Sensor Overview.** Schematic representation of (a) 3- and (b) 5- layer sensor constructions.

2.2 Breathability

Composing the sensors out of textile materials has given them an added degree of comfort relative to elastomeric strain sensors due to both their light weight and breathability. In two previous works[36,37], knit textiles were used to create capacitive strain sensors. While these two works may similarly achieve tactile comfort, silicone is used between fabric layers, compromising the breathability and thus, thermophysiological comfort of the sensor. Breathability can be characterized by two parameters - air permeability and water vapor transmission rate (WVTR). We characterized these properties for all the fabrics tested in our sensors, both bare and laminated with adhesive (Fig. 2). Although air permeability of the laminated fabrics was reduced relative to the bare fabrics, the laminated dielectric fabrics all showed air permeabilities greater than 100 L/m^{2}s. On average, the laminated conductive fabric showed an air permeability slightly less than this value. However, the order of attachment of the adhesive to the fabric likely plays a role in the porosity of the fabric-bonded adhesive and air permeability of the overall composite. Thus, attaching the adhesive to the dielectric fabric first enables the air permeability of composite layers to be greater than 100 L/m^{2}s and similar to that of normal clothing[38]. Our results also showed a slight reduction in WVTR for laminated fabrics relative to bare fabrics. However, the WVTRs of the laminated fabrics were measured to be between 38-41 g/m^{2}h, which are still greater than and within the ranges transepidermal water loss (TEWL) rates of adults during resting conditions (5-10 g/m^{2}h)[39,40] and during sweating (6-66 g/m^{2}h)[40] respectively. For further comparison, a non-porous 8 µm film[41] and a porous 40 µm film of PDMS[42], an elastomer commonly used in strain sensors, were shown to have WVTRs of 5-6 g/m^{2}h and 20.3 g/m^{2}h respectively. Both these values are less than those of the laminated fabrics.

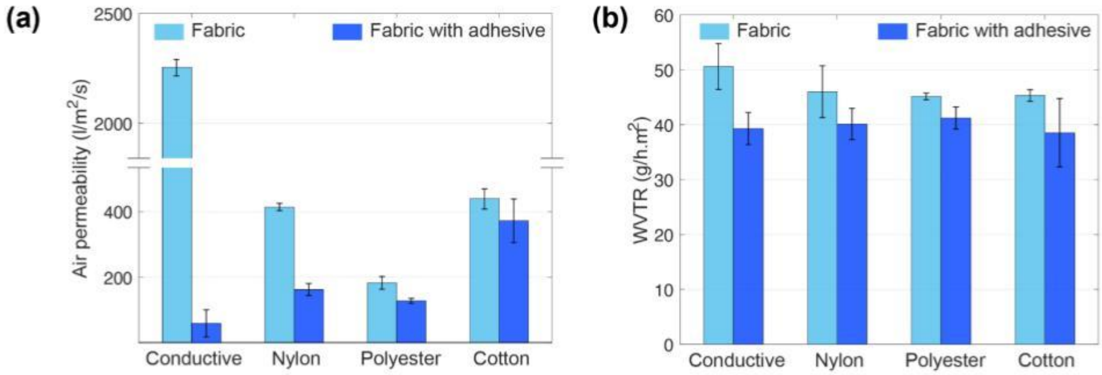

Figure 2. **Constituent sensor material breathability.** (a) Air permeability and (b) Water vapor transmission rate of bare and laminated fabrics. An average of three samples is shown for each.

2.3 Electromechanical Characterization

Our sensors all exhibit a monotonic increase in capacitance in response to increasing strain. This response is largely related to the geometrical changes of the sensor with strain, which are governed by the Poisson's ratio of the sensor[43]. The Poisson's ratio of the sensor, its strain dependence[43], and strain-dependent changes in the dielectric fabric mesostructure including changes in porosity[44], partial alignment of fibers[45], and compressive deformation[43,46] may explain the nonlinearity of the normalized relative capacitance. The sensitivities of the 5-layer sensors of each dielectric material are shown by the slope, S, within three strain ranges ($\varepsilon < 25\%$, $25\% < \varepsilon > 50\%$, $\varepsilon > 50\%$) in Fig. 3a-c. These three linear strain ranges were chosen to allow for calculation of the sensor sensitivities and subsequent comparison to similar textile-inclusive literature. Both the nylon and polyester dielectric sensors show similar sensitivities to other capacitive strain sensors utilizing textile materials. Comparatively, the cotton dielectric sensors show the lower sensitivities in all three strain regions. This result can be attributed to the reduced elasticity of the cotton fibers (which have only 5% spandex relative to 20% in the nylon and polyester fabric), the relative permittivity of cotton fibers, and the fabric density.

The sensors proved to be mechanically robust, withstanding 5,000 consecutive strain cycles from 5% to 60% strain without failure (Fig. 3d-f). The polyester sensors showed a stable normalized relative capacitance of about 0.43% at ~60% strain. The nylon sensors demonstrated a similar stability, with a slight drift in the normalized relative capacitance ($\Delta C/C_o$) from about 0.51% to 0.45% at ~60% from the first cycle to the last. The cotton sensors demonstrated the lowest electromechanical stability likely due to factors such as slower rearrangement of the cotton fibers[47], the reduced spandex percentage and elasticity, and the hygroscopic nature of cotton.

The measurement frequency of capacitance is positively related to the excitation frequency of the applied voltage signal[48]. We found that the capacitance of nylon and polyester dielectric sensors was stable at excitation frequencies up to 1 MHz, while cotton dielectric sensors showed a more monotonically changing capacitance at higher frequencies (Fig. 3g-i). This effect is most likely a result of cotton's relatively higher moisture content and higher content of bound water. These results indicate that the capacitance of the nylon and polyester dielectric sensors, in particular, can be accurately measured at a higher range measurement frequencies which has been a challenge in previous sensors due to their high resistance electrodes[48]. We then assessed that the nylon and polyester dielectric sensors exhibit the most favorable characteristics for a wearable strain sensor, prompting further characterization of these sensors as described in Sections 2.4 and 2.5.

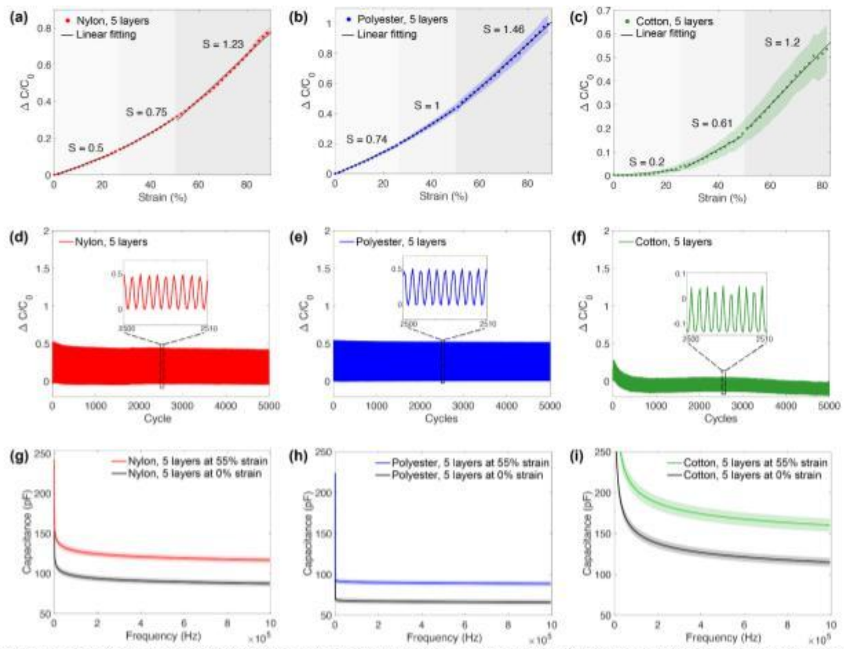

Figure 3. **Electromechanical characterizations of 5-layer textile sensors.** (a-c) Average relative change in capacitance as a function of strain for five sensors with dielectric (a) nylon, (b) polyester, and (c) cotton. Three sensitivity regimes are shown. (d-f) 5000 strain cycles to ~60% for a representative sensor with dielectric (d) nylon, (e) polyester, and (f) cotton. (g-i) Average change in capacitance as a function of frequency for five sensors with dielectric (g) nylon, (h) polyester, and (i) cotton.

2.4 Electromechanical Dependence on Temperature and Humidity

We also tested the response of the nylon and polyester sensors to different ambient temperature and humidity settings. Although the sensors continued to show a monotonic increase in capacitance with strain (Fig. 4a-b), both showed reduced sensitivities as a result of higher humidity (Fig 4c-d). When introduced to higher humidity, the sensors uptake water to establish dynamic equilibrium. This uptake causes an increase in the relative permittivity of the dielectric fabrics and thus, the sensor capacitances because water replaces air in the fabric structure and water (ε_r = 78 at 2.45 GHz and 25℃)[49] has a much higher relative permitivity than air ($\varepsilon_r \approx 1$)[49,50]. However, this increase in capacitance was not fully accounted for by normalizing the data as there is an obvious reduction in sensitivity for both sensor types as a result of higher humidity. Higher temperatures also resulted in a slightly higher sensor sensitivity which is thought to be the effect of increased drying at higher temperatures, reducing water uptake and effectively counteracting the effects of higher humidity. These results present a practical challenge in the implementation of the sensor, which will be discussed further in Section 3.1.

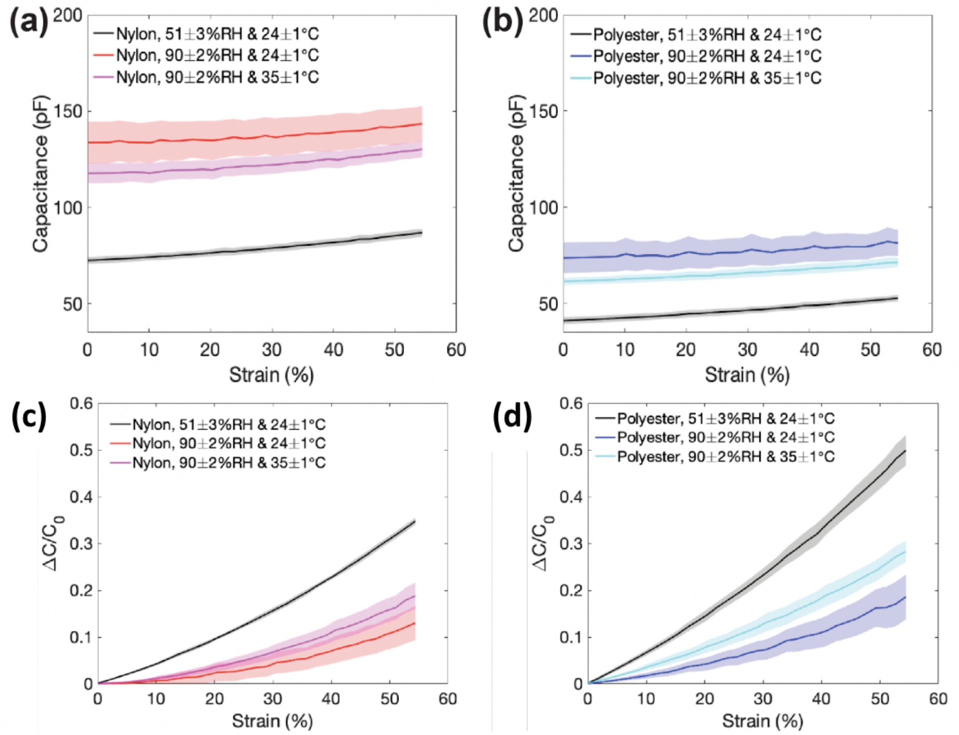

Figure 4. **Effect of temperature and humidity on the electromechanical response of 5-layer sensors.** (a-b) Average capacitance as a function of strain, under varying temperatures and humidities, for five sensors with dielectric (a) nylon and (b) polyester. (c-d) Average relative change capacitance as a function of strain, under varying temperatures and humidities, for five sensors with dielectric (c) nylon and (d) polyester.

2.5 Response to Washing

The nylon and polyester dielectric sensors were washed, dried, and tested three times, showing an initial decrease in sensitivity with strain after the first washing cycle and no clearly distinguishable differences after repeated washing cycle (Fig. 5). This result supports reusability of the sensors. The initial sensitivity reduction may have been the result of permittivity changes in the dielectric fabrics induced by changes to the fiber contact network and tightness of the fabric knit structure after fibers swelled with water and then dried.

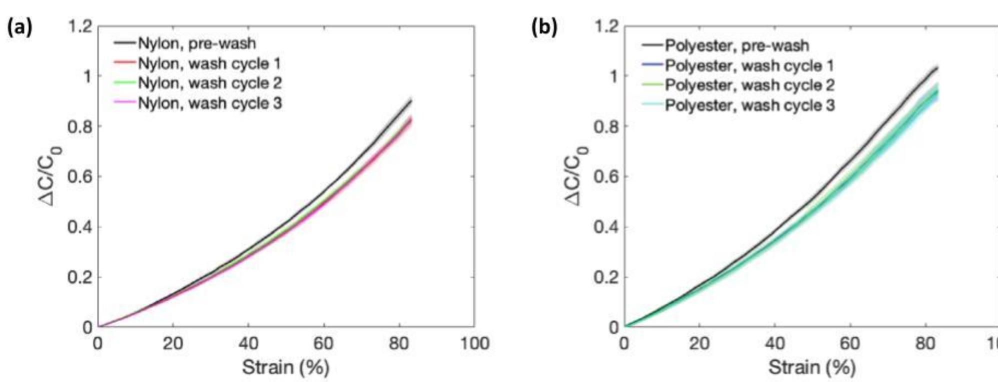

Figure 5. **Effect of washing on the electromechanical response of 5-layer sensors.** (a-b) Effect of washing on the electromechanical response of five sensors with dielectric (a) nylon and (b) polyester.

2.6 Sensor Integration and Demonstration

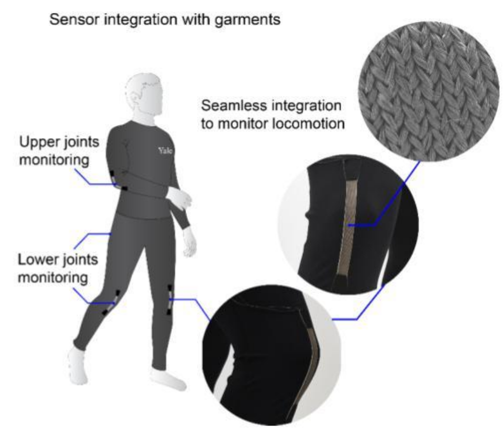

Figure 6. **Sensors integrated into garments for human motion monitoring.**

Previous sensor literature has demonstrated the challenge of modularly mounting sensors[51,52] on joints such as the difficulty with ensuring sensors stay on the joint rather than laterally slipping off and that the sensor does not shift up or down during motion, reducing the strain experienced by the sensor. To circumvent these challenges, we developed a method of semlessly integrating the sensors into commercial activewear using the thermoplastic adhesive film to bond the sensor layers to the garment and incorporating the garment as one of the dielectric layers in the sensor construction (Fig. 6). As shown in Fig. 7, I demonstrate a sensing suit capable of providing real-time in-situ data describing elbow, hip, and knee joint motion. Thus far, I have detailed the creation and characterization of comfortable, stretchable sensors capable of describing human motor activity unobtrusively.

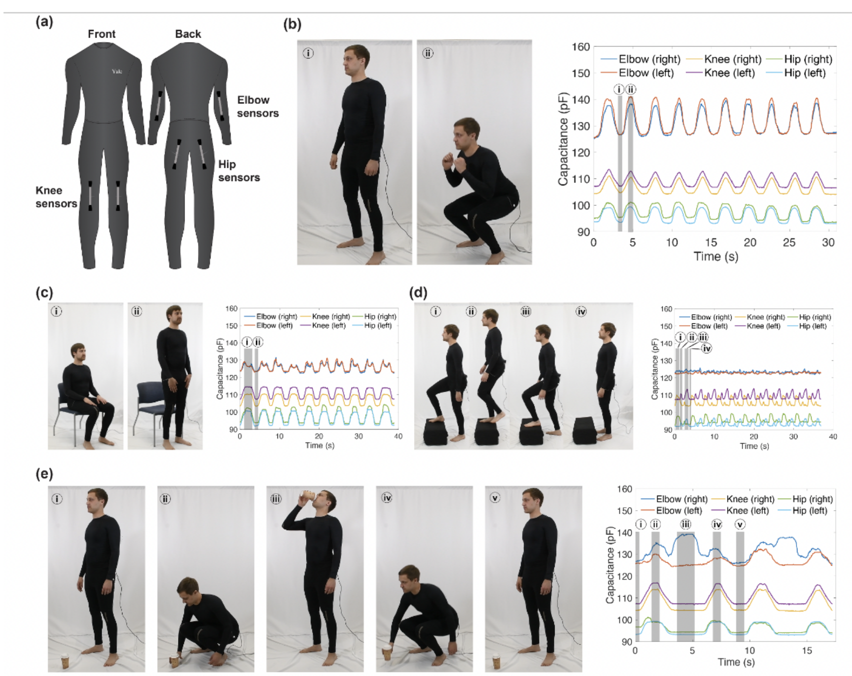

Figure 7. **Sensorized smart garment capable of monitoring the movement of body joints.** (a) Sensor placement on the knees, elbows, and hips of the garments. Photographs and capacitance responses of the sensors during the following human motions: (b) squats (10 cycles), (c) sit–to–stand cycles (10 cycles) (d) step–ups (10 cycles) and (f) retrieving an object from the floor (2 cycles).

3 Dynamic Interpretation of Sensor Output

In this section, I introduce my plan to address the ongoing challenges of our sensor technology and further develop the initial prototype sensing suit. In Section 3.1, I propose my plan to address the challenge of sensor sensitivity to environmental effects and sweat. In Section 3.2, I discuss the next steps in improving the current sensor interfacing and wire management. This will be followed by a description of my proposed plan for relating sensor output to joint bend angles.

3.1 Compensating for Ambient and Local Conditions

In order to mature our sensor technology, it is important to further our understanding of its sensing mechanism and identify both the variables and the process by which the sensor response may be affected. As established by the results presented in Section 2.3, our sensor shows a nonlinear increase in capacitance with increasing strain. This increase in capacitance can likely be explained by strain-related effects from both geometrical changes (overlapping electrode areas and dielectric thickness) and changes in the dielectric fabric mesostructure. I hypothesize that changes in the dielectric fabric structure are reflected in a strain-dependent change in relative permittivity of the dielectric fabric. I plan to simultaneously measure the strain, capacitance, overlapping electrode area, and dielectric thickness of the sensors to isolate and calculate the dynamic Poisson's ratio of the sensors and the dynamic relative permittivity of the dielectric layers. This experiment will be completed in multiple temperature and humidity settings to gain information about how the strain dependence of both Poisson's ratio and relative permittivity changes with humidity and temperature. The expected output of these results is a greater understanding of the underlying sensing mechanism and how it is affected by ambient environmental conditions.

From there, I will begin investigating the effect of local conditions, namely sweat. As users wear the sensing suit and sweat overtime, this biofluid will be absorbed into the fabric and will most probably affect the sensor response. As such, I will augment the aforementioned experiments by repeating these experiments with different simulated sweat conditions. Such experiments will include independent variables such as ionic concentrations, sweat rate/volume, and the coupled effect of sweat with ambient conditions (temperature and humidity). I will also test the recovery of the sensing signal after washing and drying previously sweat-saturated sensors.

Using the collected data, I plan to develop an empirical model to account for the change in sensor response as a result of these external conditions to enable accurate interpretation of the capacitance output. I predict that the result of this empirical model will require the integration of temperature, humidity, and possibly sweat sensors to provide enough input information to determine the expected capacitance-strain response. This model will be important in calibrating the sensors as described in Section 3.2 below.

3.2 Relating Sensor Output to Joint Motion

In order to translate our sensor output to a physiologically relevant metric, I plan to create a calibration curve relating sensor capacitance to joint bend angle. First, I will refine the sensing garment we currently have by untethering the data acquisition electronics using a bluetooth enabled module. I will then focus on improving wire management by substituting the current flexible, silicone-sheathed wires (30 AWG) with a commercially available, solderable, and machine sewable insulated conductive thread that can be better integrated into the garment. After making these improvements, I plan to use motion capture to assess joint bend angle while also collecting capacitance sensor data. Previous works[53-56] have used similar such measurements to directly relate joint bend angle to strain. My end goal of this section is to be able to use an input of sensor capacitance to obtain the desired output of joint bend angle. After simultaneously collecting motion capture joint bend angle data and sensor capacitance data, I will use our known relationship of capacitance and strain to relate the joint bend angle to strain and create a calibration curve. Note that this may require me to retest the relationship of capacitance and strain using the exact activewear material as a dielectric sensing layer. Application of the known relationship of capacitance and strain will likely also require use of the multimodal sensing and the empirical model as described in Section 3.2. I will then conduct additional data collection trials using both motion capture to measure joint angles and the capacitive sensor. The sensor data will be used to calculate joint bend angle based on the developed calibration curve and the accuracy of the calculated joint angle will be determined using the motion capture joint angle data.

4 Rapid Eye Movement (REM) Sleep Behavior Disorder Detection

4.1 Conducting Human Subject Sleep Studies

To test the use of my matured sensor technology in its ability to nonintrusively measure motor activity during REM sleep, I will prepare sensing suits to be used in conjunction with additional sensing technologies common in PSG. In particular, I plan to use a headband or cap device with EEG, EOG, and EMG sensing capabilities to collect physiologic activity data during sleep. The use of EEG and EOG will enable sleep stage determination, satisfying the need to identify when the subject has entered REM sleep. EMG will be used to determine muscle atonia during REM sleep as loss of normal muscle atonia during REM is a criteria for RBD diagnosis. For the purpose of testing the sensors as an objective RBD detection technology, I will conduct these sleep studies with three types of human subject groups - (1) health subjects, (2) PD patients with RBD, (3) PD patients without RBD.

4.2 Analysis of Human Subject Motion Data During REM Sleep

Here I will focus on the analysis of the sensor data collected during the sleep studies. First, I will use the EOG and EEG data to pinpoint the time intervals in which subjects are in REM sleep. I will then use the EMG data to determine whether muscle atonia was observed or not and record the data as a binary classifier. I will begin by using the strain sensor outputs (converted to joint angles) to find the speed, frequency, duration, and amplitude of motions for each joint. I will then apply a machine learning approach to find existing correlations between these metrics of motor activity in each joint and the EOG, EEG, and EMG data. This approach will then be used to train a model with the goal of classifying individuals as part of the correct human subject group. If this method proves successful in this goal, it would be worth investigating how many and which strain sensors are necessary to maintain classification accuracy.

5 Overview of Project Outlook

5.1 Potential Impact

This work has the potential to contribute to wearable strain sensing, sleep study technology, and RBD diagnosis. The work proposed in Section 3 aims to mature our current sensor technology to have use beyond simple demonstration. The end result would introduce a sensing suit with embedded, breathable capacitive sensors that are suited for long-term wearable applications. Beyond the specific application included in this thesis prospectus, this sensing technology has potential uses in other health monitoring applications, rehabilitation, sports performance, virtual reality, and animation. The application of this project in particular also has the potential to improve the current state of RBD diagnosis by introducing an unobtrusive sensing system that can potentially be adopted as an alternative at-home sleep behavior diagnostic tool.

5.2 Essential Resources

Dr. Rebecca Kramer-Bottiglio's lab currently has all the required resources needed to complete the research described in Section 3, including a materials testing system (Instron 3345), an environmental chamber (ETS, Model 5500-8485), a LCR meter (Keysight Technologies, E4980AL) and a motion capture system (Phasespace, Inc.). The research described in Section 4 will require an approved Institutional Review Board (IRB) protocol and collaboration with the sleep medicine department within the Yale School of Medicine. I have already identified and started a collaboration with Dr. Brian B. Koo for access to the necessary facilities and human subjects, clinical equipment, and guidance regarding IRB procedures and data interpretation.

5.3 Provisional Timeline

	Q1 2023	Q2 2023	Q3 2023	Q4 2023	Q1 2024	Q2 2024	Q3 2024	Q4 2024	Q1 2025	Q2 2025	Q3 2025	Q4 2025	Q1 2026	Q1 2026
Objective 1: Compensating for Ambient and Local Condition Effects														
Characterizing Effects of Ambient Conditions on Sensor Response														
Characterizing Effects of Local Conditions on Sensor Response														
Develop an Empirical Model to Predict Sensor Response														
Integration of Ambient and Local Condition Sensors														
Objective 2: Relating Sensor Output to Joint Motion														
Untethering Sensor Data Collection and Improving Wire Management														
Simultaneous Motion Capture and Strain Sensor Data Collection														
Objective 3: REM Sleep Behavior Disorder Detection														
Human Subject Sleep Sleep Studies														
Analysis of Motion Data During REM Sleep														
Thesis Defense														

5.4 Publications

- **A. Agrawala***, L. Sanchez-Botero*, and R. Kramer-Bottiglio. Stretchable, breathable, and washable textile sensor for human motion monitoring. *Advanced Functional Materials*. (Under Review) (*Authors contributed equally)
- W. Johnson, **A. Agrawala**, X. Huang, J. Booth, and R. Kramer-Bottiglio. Sensor Tendons for Soft Robot Estimation. IEEE Sensors, 2022. Best Paper Award.

- E. Porte, S. Eristoff, **A. Agrawala**, and R. Kramer-Bottiglio. Characterizing elastomers and elastomer-based sensors in extreme environments. (Under Preparation)
- S. Woodman, L. Sanchez-Botero, D. Shah, M. Landesberg, **A. Agrawala**, and R. Kramer-Bottiglio. Stretchable single-board computers. (Under Preparation)

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