PDMSkin: On-Skin Gestures with Printable Ultra-Stretchable Soft Electronic Second Skin

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ABSTRACT

Innovative enabling technologies are key drivers of human augmentation. In this paper, we explore a new, conductive, and configurable material made from Polydimethylsiloxane (PDMS) that is *capillary* doped with silver particles (Ag) using an immiscible secondary fluid to build ultra-stretchable, soft electronics. Bonding silver particles *directly* with PDMS enables inherently stretchable Ag-PDMS circuits. Compared to previous work, the reduced silver consumption creates significant advantages, e.g., better stretchability and lower costs. The secondary fluid ensures self-assembling conductivity networks. Sensors are 3D-printed ultra-thin (<100µm) onto a pure PDMS substrate in one step and only require a PDMS cover-layer. They exhibit almost stable electrical properties even for an intense stretching of ≫200%. Therefore, printed circuits can attach tightly onto the body. Due to biocompatibility, devices can be implanted (e.g., open wounds treatment). We present a proof of concept on-skin interface that uses the new material to provide six distinct input gestures. Our quantitative evaluation with ten participants shows that we can successfully classify the gestures with a low spatial-resolution circuit. With few training data and a gradient boosting classifier, we yield 83% overall accuracy. Our qualitative material study with twelve participants shows that usability and comfort are well perceived; however, the smooth but easy to adapt surface does not feel tissue-equivalent. For future work, the new material will likely serve to build robust and skin-like electronics.

CCS CONCEPTS

• **Hardware → Emerging interfaces**; • **Human-centered computing → Gestural input**.

KEYWORDS

skin interface, novel material, strain sensing

1 INTRODUCTION

New technologies often enable innovative augmented human applications. A research space that is, for example, driven by new materials surrounds on-skin interfaces, actuators, and devices. Particularly, highly flexible materials can be useful to create interactive surfaces that apply directly to or even under the skin of the user. Previous work has shown that such surfaces potentially enable diverse input modes such as touching, grabbing, pulling, pressing, scratching, shearing, squeezing, or twisting [2]. A primary benefit of the paradigm lies within the user always *wearing* the input device and also the possibility to present information to the user with vibrotactile cues or passive haptic learning (e.g., [5, 20]).

However, a fundamental problem of on-skin sensors, actuators, and interfaces is blending the technology in with the user's body. Ideally, a skin interface does not affect the user's behavior, is supple and integrates well with the skin and does not impair movement capabilities. For example, Holz et al. [9] "implanted" different sensors and actuators into artificial skin and onto the arm to enable enjoyable interactions that blend in with the user's body.

Polydimethylsiloxane (PDMS) based materials have great potential for achieving these objectives in real-world use cases. We expect that PDMS-based sensors can be manufactured ultra-thin, ultra-stretchable, and also implantable. These unique characteristics could ultimately enable a natural *second skin*. In this paper, we present our initial exploration of using a novel type of silver doped, 3D-printable PDMS, so-called *capillary Ag-PDMS* [25]. The material is different from binary Ag-PDMS, as it incorporates an additional immiscible fluid to form self-assembling conductive networks, thus resulting in novel properties. Overall, compared to other existing materials, it has several advantages for building highly flexible, skin attachable interfaces: First of all, interfaces can be freely 3D-printed using COTS 3D printers. Second, building sensors requires only two steps (printing and protective cover) without significant manual effort even in a batch size one process. Thirdly, the material has easily configurable properties like stretchability and thickness, which allows designing soft and smooth skin-covering interfaces in various configurations (details are presented in section 2 and 3.1).

Based on the new material, this paper makes the following contributions:

- we show an initial successful application of the new material as on-skin interface with ten subjects and a six-class machine learning classifier
- we report on the system design and qualitative experience when using the new material for constructing interface strain sensor devices
- we evaluate the comfort and ease of use for the new material with twelve subjects

2 BACKGROUND AND RELATED WORK

Skin interactions are a heavily researched field today. The survey paper by Bergström and Hornk [2] gives a general overview of the current state of the art, including input types (e.g., touch or deformation) and possible locations (e.g., arm or nose). They also look into the underlying technologies for enabling on-skin input (e.g., optical or electrical sensing) and providing feedback (e.g., visual or tactile). The article by Steimle [23] covers the overall field as such and stresses the potential of on-skin interfaces in mobile computing. The idea of electronic skin started already 20 years ago (e.g., [19]). Therefore, past research looked into various applications, principles, and materials for implementing such devices.

Enabling skin interactions is a challenging task, mostly requiring specialized equipment. Existing skin interfaces, for example, did measure waves propagating through the body when touching, which is susceptible to body composition (e.g., [8]). Others have used infrared line sensors as well as 3D or depth cameras attached on shoulder or wrist, which all require a line of sight making them impractical for some day-to-day use cases (e.g., [16], [7], [22]).

More recent approaches apply flexible interfaces and surfaces directly onto human skin. For instance, very similar to regular non-permanent tattoos, Kao et al. [10] used gold leaves to create interactive and fashionable on-skin interfaces. A key advantage of the approach is that devices can be manufactured ultra-thin and have many useful application areas. Similarly, in additional studies Lo et al. [18] presented how such tattoos can have many different configurations and looks. However, for long term usage where the second skin permanently covers the arm, these approaches are not durable enough. Additionally, they might be challenging to use in some scenarios, e.g., when covered by the shirt of the user.

Good alternatives are, therefore, more durable materials based on stretchable Polydimethylsiloxane (PDMS) doped with conductive particles. Past on-skin applications of different composites included silver (Ag) [15], carbon nanotubes (CNT) [27], or carbon-filling (cPDMS) [28]. Combining the conductive doped PDMS material with non-doped PDMS substrate and cover layers allows building sensors, actuators, and other circuits. Processing the materials is possible by mechanical cutting or lithography/screen printing (e.g., [31]). Some approaches rely on adding an optional brittle silver ink or carbon layer on top of the flexible conductive material, which improves conductivity under stress [30]. Overall, all of these approaches usually rely on multiple cutting and lithography steps combining doped and non-doped materials, which creates manufacturing overhead. Seminal work by Kawahara et al. [11] and also Khan et al. [12] uses simple 3D-printing technology to implement flexible electronics, which is more accessible for non-experts. We take inspiration from these approaches for our on-skin interface.

For detecting touches and other interactions, the majority of skin-applied related work uses one of two principles, capacitive or resistive sensing. Capacitive measurements yield sensitive matrices with high spatial resolution in various configurations [6]. Alternatively, resistive measurements help to build a touch-sensitive surface with lower spatial resolution but better flexibility for detecting squeeze and stretch.

An approach making use of the formerly mentioned fundamental concepts is, e.g., iSkin [28]. The principle uses conductive cPDMS and plain PDMS in five layers to implement a touch-sensitive skin system based on capacitive measurement. Manufacturing the devices requires multiple laser cutting steps, resulting in 190-700μm thick sensors. Another example by the same group also presented a setup that uses multiple material combinations, utilizing PVC as dielectric and PET or tattoo paper as a substrate in a screen or inkjet printing process [24]. The overall system builds up a sensitive matrix consisting of six layers in total, whereas a simple resistor setup would require four layers.

Besides the hardware aspect, past research thoroughly studied on-skin interactions as a paradigm (e.g., types of gestures). We do not contribute to this research area but are motivated by the field as such for our evaluation study. Some early work covering interaction possibilities dates back almost ten years [17]. The implementation that we present in this paper ties in with more recent research on secondary artificial skin devices that mimic the properties of human skin [26] which we, at some point. would also like to achieve on top of human skin.

All in all, we wanted to follow the existing research on the subject but envisioned a more stretchable, more robust, and more simple-to-set-up underlying principle for manufacturing on-skin interfaces. Therefore, we use *capillary Ag-PDMS*, which is PDMS doped with silver particles mixed with an immiscible fluid to create self-assembling conductive networks. Sun et al. [25] initially introduced capillary Ag-PDMS as a fundamental material for building ultra-stretchable devices. The new material has, depending on the configuration, superior stretchability (up to 1000%) compared to cPDMS (<50%), and binary Ag-PDMS (<200%). Due to its unique conductivity properties (self-assembly networks), capillary Ag-PDMS is less brittle than existing silver-based stretchable conductors and does not require support layers to achieve electrical stability [30]. More notably, capillary Ag-PDMS returns to its initial electrical properties even after intense stretching. Additionally, instead of cutting or screen printing, we can freely print the material in any shape, requiring only three layers for a simple resistor compared to up to four layers in related work [24]. To our knowledge, we present the first application of the new material as an on-skin interface based on resistive measurements and a machine learning classifier. With the well definable material affixing onto the soft skin of the user, we can observe sharp peaks in the sensor signal when force is applied, leading to a sensitive touch system with excellent force detection.

3 WORKING PRINCIPLE

In order to collect experience and validate the properties of our *capillary Ag-PDMS-based* artificial skin, we have built a rather simple interaction prototype. The interface attaches to the skin of the forearm with medical adhesive tape and supports a set of typical control gestures including swiping, squeezing and stretching (see Figure 5). The hardware configuration consists of six ultra-stretchable 3D-printed strain resistors (see Figure 4). Gestures are classified using a brute-force machine learning approach, which yields promising results for all six interactions and a neutral class.

3.1 Material Properties and Layering

The primary new material that we use for our sensing principle is Polydimethylsiloxane (PDMS) that is *capillary* doped with silver particles (Ag) using an immiscible secondary fluid. Readers should refer to the work by Sun et al. [25] for detailed explanations of the manufacturing process of capillary Ag-PDMS. The forces introduced by the secondary fluid support the self-assembly of conductive particle networks (see Figure 1) [13]. This principle results in less than two thirds of silver consumption compared to the current state-of-the-art and follows superior stretchability while maintaining excellent electrical capabilities even for an intense stretching of $\gg$100–200%.

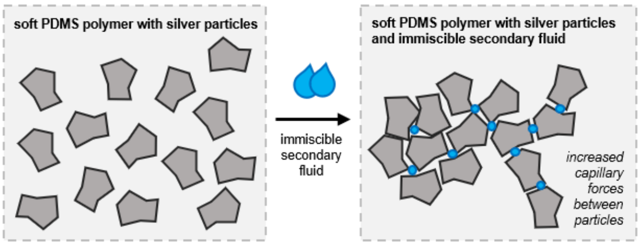

Figure 1: The schematic drawing shows how the secondary immiscible fluid improves the capillary forces between silver particles. The illustration was adapted from the work by Sun et al. [25].

3.1.1 System Cost vs. Sensor Properties.

The 15 vol% Ag-PDMS that we use in this paper costs around $60 per 100 grams. We consider it to be at the sweet spot between conductivity (1300 S/cm) and stretchability. Another configuration would be, for example, 21 vol% Ag-PDMS that has a conductivity of 3000 S/cm. However, this severely impacts the stretchability of the material (< 50 %) and is, therefore, less suitable for sensors but applies for, e.g., flexible circuits. Additionally, we can, in general, increase the conductivity by raising the curing temperature, which again comes at the cost of decreased stretchability. According to the literature, the decomposition temperature of PDMS is up to 550 °C [14]. Therefore, human-centered applications guarantee thermal stability.

Silver is more robust than carbon as doping material. In general, the costs of silver are higher than those of carbon, but as with *capillary Ag-PDMS*, the amount of silver reduces to around 30% compared to state-of-the-art binary Ag-PDMS. Thus there is not much difference in price compared to CNT-PDMS or cPDMS anymore (e.g., cPDMS costs $60 per 100g, which is equally cheap as capillary 15 vol% Ag-PDMS). Because we 3D-print the conductive traces used for the sensor, there are only simple and few processing steps to be performed in contrast to the standard use of mechanical cutting or lithography/screen printing processing steps. Thus the overall system costs are likely to be cheaper than traditional setups with carbon-based PDMS settings.

3.1.2 Sensor Printing, Layering and Resulting Properties.

Due to the viscosity of the novel material, it can be flexibly and freely 3D-printed without the application of a thickening agent, which would decrease conductivity. Using a transparent PDMS substrate and printing *capillary Ag-PDMS* onto it results in inherently highly flexible devices such as precise circuits, sensors, and also actuators. Using PDMS substrate and PDMS-incorporated-ink also secures lamination and bonding stability as similar polymers have excellent compatibility. As printed Ag-PDMS objects are sensitive to moisture and would oxidize at room temperature, in general, it makes sense to cover the printed ink with a thin protective layer (e.g., PDMS). Furthermore, PDMS is an organosilicon compound that is viscoelastic and bio-compatible once cured. Today it is already in use for medical and cosmetic products and even food (as, e.g., E900). Therefore, devices making use of *only* this material could potentially be implanted under the skin to form a fully body-embedded interface.

3.2 Sensor Implementation

To summarize, the properties allow the implementation of tiny, highly sensitive structures. A possible sensor configuration is a stretchable resistor (see Figure 2 and 3) that changes its resistance when force is applied. Using the well definable material affixed onto the soft skin of the user, we can observe significant peaks in the sensor signal even when light force is applied, leading to a sensitive touch system. The overall sensor consists of two PDMS layers with a printed *capillary* Ag-PDMS sensor in between (see figure Figure 2). The total height is mostly defined by the thickness of the PDMS substrate and the cover layer. Therefore it should be configured to the needs of the application.

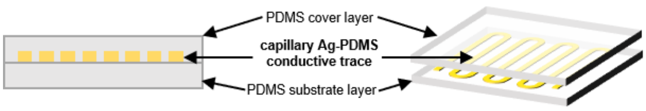

Figure 2: The illustration shows the cross-section of the capillary Ag-PDMS structure, including the substrate, conductive trace, and cover layer (left). The top view on the right shows how the conductive trace can be freely configured and is encapsulated between layers.

3.3 Sensor Design

To manufacture the on-skin interface device, we have printed six strain sensors using a *Voxel8 Developer's Kit* printer equipped with a dispensing tip on a syringe controlled by air pressure. With capacitive touch sensing, we could achieve a higher spatial resolution. Nevertheless, this would also require at least three layers to create a capacitor and two additional layers as covers, which increases the manufacturing complexity. Therefore, we apply resistive strain sensors for our initial testing of *capillary* Ag-PDMS, as these sensors can be easily read out and are more straightforward to manufacture. We chose a serpentine-shaped, 128 mm long *Ag-PDMS* trail (similar to the sensor shown in Figure 3). This approach results in higher resistance changes when strain is applied while keeping a small footprint. The traces are 150 to 200 μm thick and 300 μm wide. The gap between traces was set to 1 mm. We apply the material onto a pure PDMS substrate material with a 200 μm diameter nozzle. In our evaluation setup, we choose a 400 μm PDMS substrate layer as a compromise of high robustness and slinkiness to the skin. Our

Figure 3: The upper left image shows a resistor made from capillary Ag-PDMS. The top right image displays how the material applies to the arm, leaving almost no gap between the PDMS base layer and the skin. Even long hair tightly fits between the substrate and the arm. The two images in the bottom show the stretchability of the material, which keeps its electric properties for an intense stretch of ≫100–200%.

goal was to reduce the gap between the PDMS and the skin surface as much as possible. The resulting material is then cured in the oven at 200 °C for two hours.

3.4 Interactive System

For our evaluation, the sensors stick onto a patch of deformable *3M 2476P* adhesive medical silicone tape, which we also use to fixate the device firmly onto the user's forearm. We arrange four sensors vertically on the arm and two sensors horizontally and below the others. Figure 4 (left) illustrates the arrangement of the sensors and Figure 4 (right) displays a user who is wearing the device. The resulting interactive area is 7 x 7 cm, whereas the outer boundary of the patch is 15 x 15 cm, which ensures sufficient hold. To avoid that users accidentally interact outside the bounds of the interactive surface, we outline the input area using a water-resistant pen.

For reading the resistance of the six strain sensors, we use an *Arduino Uno*. We build a simple voltage divider with $R_{Ref} = 1kOhm$ as a known reference resistor and a strain resistor. We apply an input voltage of $V_{in} = 5V$. Every voltage divider then connects to one of the six available analog inputs of the *Arduino* device. Our firmware reads the voltage values (V_{out}) from these pins every 10 ms for every sensor. With the known input voltage V_{in} and known resistor R_{Ref} we can calculate the resistance of R_s (the strain sensor) as follows:

$$R_s = R_{Ref} * (\frac{V_{in}}{V_{out}} - 1)$$

Even though we can configure our material precisely, small variations remain because of the printing process. Therefore, we have to take into account that the resistance can vary between the different sensors. To mitigate the issue, we perform a calibration by collecting 150 resistance measurements of each sensor in a relaxed state. We then choose the mean of those samples as R_0, which characterizes the sensor's base resistance. The base resistance values that we could observe for our sensors vary between 180 and 460 Ohm, whereas it can go as high as 5 kOhm until the sensors are no longer conductive (more electrical properties in 3.1). We compute the final sensor reading as R_s/R_0 for every sensor and expose the resistance measurements by printing them to the serial port of the *Arduino* device.

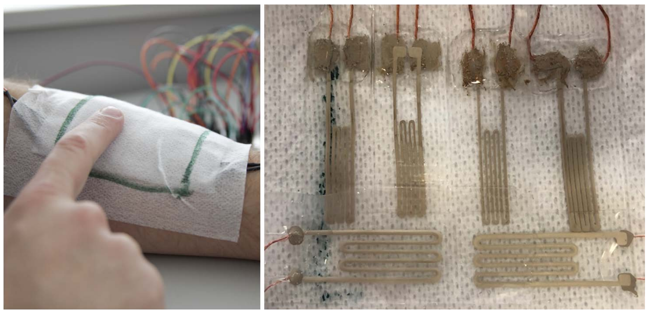

Figure 4: Our prototype device consists of six stretchable sensors that we affixed with medical adhesive tape.

Previous work has thoroughly looked into the design of on-skin gestures for different parts of the arm [29]. In contrast to the usability focused view on skin interactions, our main design objective was to enable simple input gestures that highlight different sensor aspects. Namely, this includes showing the feasibility of both pressure and strain-based applications. All interactions should be target free and independent from specific locations within the on-skin interface (compared to, e.g., a virtual button). This approach simplifies the data acquisition process and improves fault tolerance. The first prototype sensors that we manufactured from the novel material still have a rather low spatial resolution. We, therefore, have to take into account that we have limited accuracy for the locality of touches, which limits the types of possible interactions. As users can reach their arm on one side comfortably with the hand on the other side, we decided to use the forearm as an interactive surface for our explorative experiments.

Taking all previous aspects into account, we designed six distinct gestures that could serve as generic controls for different applications. Figure 5 illustrates the different interactions. They include swiping left and right with the index finger as well as swiping up and down with all fingers except the thumb. More advanced gestures include squeezing and also stretching the skin with the index finger and thumb vertically. By letting users execute the different gesture types on a skin applied prototype, we will be able to understand if the novel material can function as a pressure (swiping) and also strain (squeezing and stretching) soft surface input device.

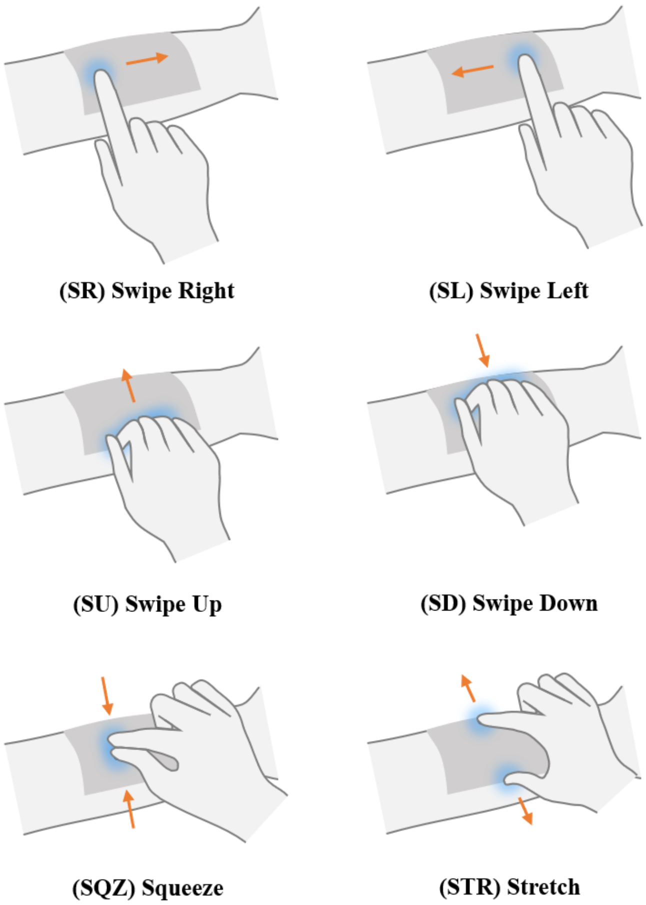

Figure 5: The different gestures that we conceptualized include swiping right (SR) or left (SL) with one finger, swiping up (SU) and down (SD) with four fingers, and also squeezing (SQZ) or stretching (STR) with two fingers vertically.

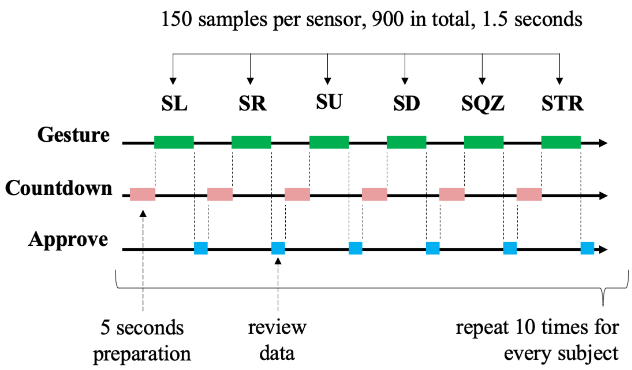

Figure 6: The graph shows the structure of the data acquisition session. Participants perform each gesture after a five-second preparation time. We review the recorded samples immediately and then manually start the next countdown.

3.5 Interaction Recognition

Interaction recognition was carried out by a classifier that we trained on the initial collection of scripted gestures. For training, we set up a data collection session with ten users that we recruited through a sample of convenience from our lab. In detail, each session looked as follows:

(1) we explain the experiment to the participant and ask them to provide their informed consent
(2) we apply the device, introduce them to every type of interaction and also resolve questions
(3) a custom-built tool provides the instructions to the user, and we record gestures and non-gestures (relaxed states between two interactions)
(4) every gesture is performed ten times in a round-robin fashion in which we give users 1500 ms to perform every interaction after a five-second countdown (see Figure 6)
(5) we collect data for a neutral, non-interaction state by recording a snapshot after performing every gesture once and after moving the arm four times
(6) after completing all tasks we collect information on the participant's gender, age, and arm circumference

The experiment took place in a dedicated room at our lab. We applied the prototype to the left arm. We asked users to avoid touching the interface before performing the next gesture. Additionally, we instructed participants to rest their arms comfortably on the table. During data collection, one person visually observed the recordings to ensure that all sensors are working in order and also that participants executed the interactions within the 1500 ms time frame. We offered participants to take a break after the first half of the data acquisition phase was completed. After the session, they were rewarded with a bag of candy for their participation.

In total, we collected 700 unique gesture recordings, each containing 900 resistor readings (150 per sensor). Six participants were male and four female. The mean age was 27.7 years (min: 22, max: 40), and the mean arm circumference where the device was applied to was 24 cm.

Figure 7 illustrates typical patterns for the different gestures that participants performed during the session. The distinct peaks that occur when a user applies pressure to a sensor can be well observed. For example, the graphs of the "swipe left" and "swipe right" gesture show how the rightmost sensor is triggered first, and the leftmost last, and vice versa. Similarly, the "swipe up" gesture results in a peak for the bottom sensors, and only then the upper sensors trigger (again, reversed for swiping down). Finally, squeezing results in a simultaneous strain of multiple sensors caused by the deformation of the surrounding skin. Stretching, however, only triggers a smaller subset of sensors. The signals are well separated in time and have distinct shapes, which suggests that machine learning will yield good results.

3.5.1 Classification. To classify the different gestures (including the neutral class), we apply a simple brute-force machine learning approach in Python. We use *tsfresh* for feature extraction that works based on scalable hypothesis testing and can calculate an extensive list of features on time-series data [4]. It includes simple features such as min/max values or the number of peaks and also

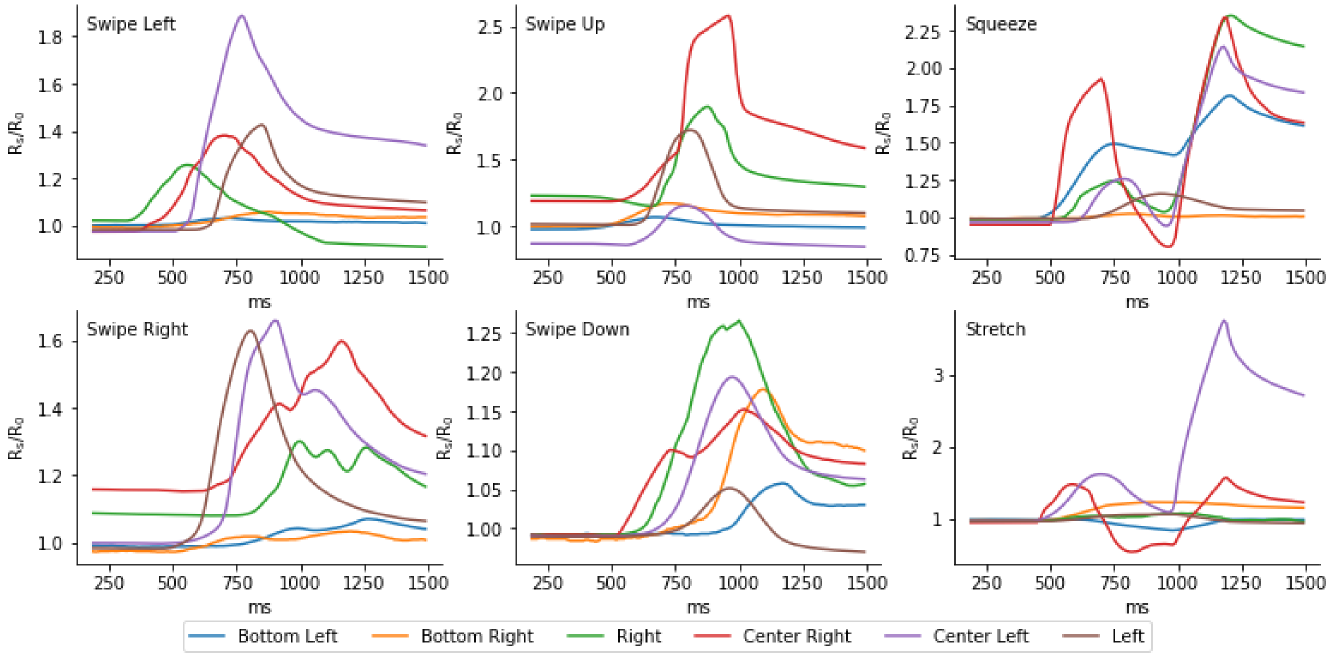

Figure 7: Sensor patterns for the different gestures swipe left, right, up, and down as well as squeeze and stretch. The peaks are well separated in time and occur in line with what one would expect for the related gesture. The improve the spikes of the plot, we have applied a moving average filter of 20 samples, which we do not use during classification.

	Leave one subject out cross-validation Overall Accuracy: 0.83			**Leave one measurement out cross-validation** Overall Accuracy: 0.84		
	Precision	**Recall**	**F1 Score**	**Precision**	**Recall**	**F1 Score**
Neutral	0.99	0.99	0.99	0.92	0.93	0.93
Swipe Left	0.85	0.79	0.82	0.75	0.81	0.78
Swipe Right	0.82	0.93	0.87	0.82	0.75	0.78
Swipe Up	0.81	0.83	0.82	0.81	0.81	0.81
Swipe Down	0.84	0.85	0.85	0.84	0.87	0.86
Squeeze	0.78	0.59	0.67	0.84	0.83	0.83
Stretch	0.75	0.86	0.80	0.89	0.86	0.87
Average	0.83	0.83	0.83	0.84	0.84	0.84

Figure 8: The table above shows the results assessed in two different settings. The left part presents the results for a leaving one subject out based cross-validation, which illustrates the generalization performance. The right side shows the results for within subject, leave one measurement out cross-validation, which yields similar good results.

more sophisticated features such as FFT coefficients. Extracting the features results in a vector with 4,728 entries for each measurement, which we use to train gradient boosted decision trees with XGBoost in its default configuration [3]. One might argue that a more transparent approach, e.g., peak detection combined with a rule-based classification principle, would create a more simple pipeline. However, to maximize the accuracy and avoid extensive feature and algorithm engineering for a preliminary prototype that will likely change in the future, we decided to go for a black-box approach.

3.5.2 Validation. For validating our model, we picked two different setups. We perform a "leave one subject out" and also a "leave one measurement out" cross-validation. For the first approach, in each iteration, we train the classifier using nine subjects and leave the remaining one out for testing. Therefore, every subject serves as the test dataset once. We iterate over all results to compute mean outcomes for every gesture and also the overall score, as shown in Figure 8. For training and testing within single subjects, we pick nine measurements of each gesture (63 in total) and use the remaining seven for testing and repeat this ten times per person. This means we perform a 10-fold cross-validation individually and then compute the overall scores.

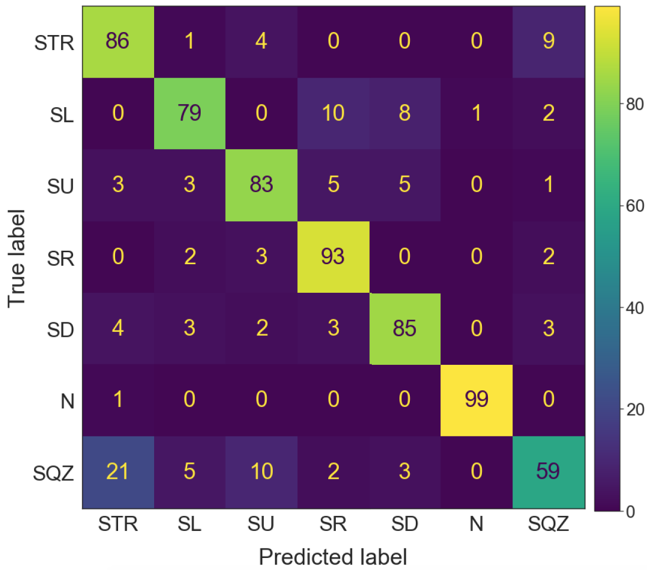

Figure 9: Leave one subject out confusion matrix showing that squeezing and stretching is most challenging to classify in our limited resolution sensor setup.

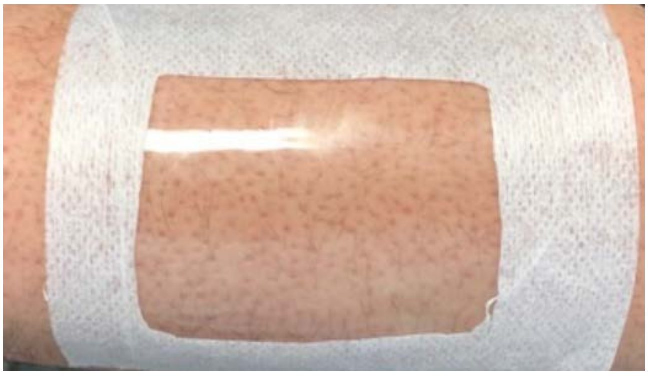

Figure 10: The patch firmly pulls the PDMS substrate material onto the user's skin on the forearm to enable a soft interactive surface that naturally blends in.

3.5.3 Results and Discussion. Figure 8 shows that we can achieve satisfying results for all gestures. Unsurprisingly, we yield high precision for the neutral state, whereas accuracy improves with samples collected from varying subjects compared to only training within users. Intuitively, more data supports overcoming small variations in the neutral class. We only observe small differences between the overall scores for the remaining six gestures. However, taking the confusion matrix shown in Figure 9 into account, results reveal that squeezing seems to be harder to distinguish from stretching. Based on the sensor readings of the falsely classified samples, we can see more strain sensors showing activation than expected. The phenomenon is likely to occur when the squeeze gesture is not performed vertically on the forearm. As this is an artifact of the low spatial accuracy of the strain sensors, we consider our results acceptable for further investigations. For example, a capacitive on-skin interface made from the same new material could improve the resolution of the device. Both approaches "leave one subject out" and "leave one measurement out" show similar good results, suggesting that a pre-trained model and an individually trained model can both achieve satisfactory performance.

4 MATERIAL EVALUATION

To further understand the applicability of the ultra-stretchable PDMS substrate and cover material, we were interested in evaluating how users perceive it. Previous work has explored the use of different textures and materials for a secondary artificial skin device [26]. However, the authors did not place the synthetic skin on the user's arm. Even though our study design is different, the questions that we asked took inspiration from their work.

We conducted a controlled lab study where we apply a patch combined with a 5 x 7 cm PDMS cutout to the participant's forearm and let them interact with it. The sample has a similar thickness to the sensors that we used earlier for our interaction experiments. We ask participants to rate five-point Likert scale items, which includes questions about if the material feels like human skin, how comfortable and easy it is to execute gestures on the surface, and also if it looks like an electronic device. We also ask if the material feels comfortable on the forearm. Finally, we ask them to provide ratings on the scales fake/natural, humanlike/machinelike, and also artificial/lifelike, which is commonly used to assess anthropomorphism [1]. We also collect age and gender. In detail, the study includes the following steps:

(1) we explain the experiment to the participants and ask them to provide their informed consent
(2) a patch with a cutout for pure PDMS material is applied to the user's forearm (see Figure 10), similar to the interactive device that we proposed earlier
(3) we walk participants through the different gestures that we designed and also present them the illustration in Figure 5
(4) participants execute the gestures we previously showed them and are also allowed to interact with the material freely
(5) participants fill in the questionnaire

4.1 Results

In total, we recruited twelve participants to volunteer in our study (sample of opportunity; three females and nine males; age range: 19 to 56 years old; median age: 26.4 years). We ran the study in a dedicated room, in which one instructor was present during the entire test to oversee the steps involved.

We refer readers to Figure 11 for the results of the questionnaire. Overall, participants report that they can easily and comfortably perform gestures. Additionally, the material feels comfortable on the skin and does not look like a device. We suspect that the close attachment supports the overall perception – however, as we did not print a circuit onto the substrate material, Ag-PDMS patches might blend in less subtly. Study participants do not think that the device feels like human skin, most likely due to the smooth texture of the substrate. We could solve that problem by configuring the PDMS with a different surface structure (e.g., by pouring it in a skin-like negative cast). Additionally, to be blend in with human

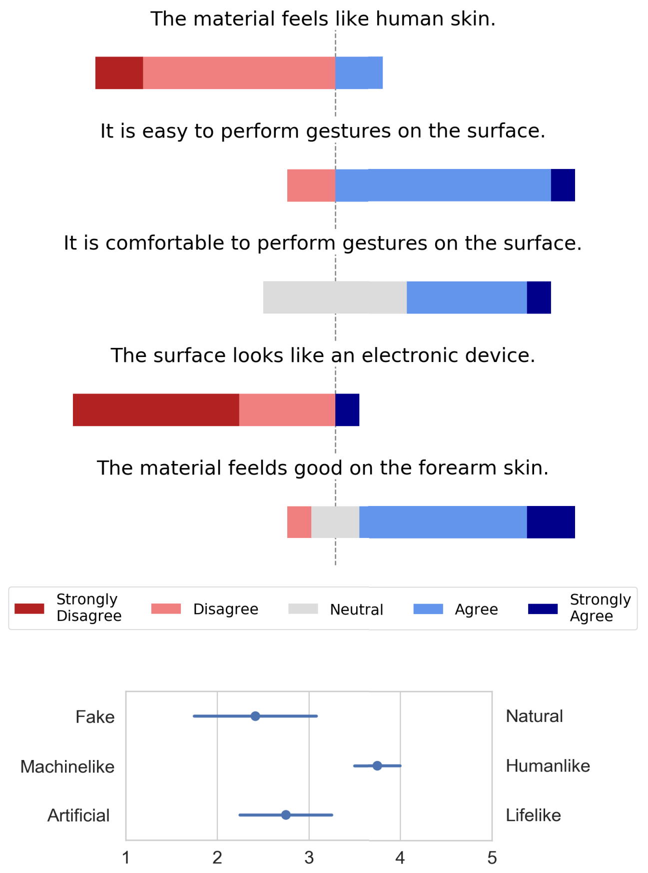

Figure 11: Results of the questions that we asked participants during our study and also the results of the anthropomorphism rating scales.

skin, the thickness of the PDMS should be further reduced, or the color could be changed. These modifications will likely also improve the items of the anthropomorphism scale shown in Figure 11 that currently does not fulfill the expectations that we have for a fully conformal and subtle on-skin interface.

5 DISCUSSION AND LIMITATIONS

Our initial results are promising. With our proof-of-concept on-skin interface, we showed that it is possible to build a stable wearable interaction system based on the novel *capillary Ag-PDMS* material. The study that we present shows that the strain sensors can serve as an input channel for simple gesture recognition tasks. Overall, we achieved 83% accuracy with our prototype. Currently, due to the low spatial resolution, we still have considerable confusion, e.g., between stretching and squeezing. With more data, higher spatial resolution, and hand-crafted features specialized for the respective gestures, we expect a higher detection accuracy. Optimizing the hyperparameters of our model or using a more sophisticated model altogether might further improve the performance. Past research has suggested a set of interactions that we did not consider in our study, which we, therefore, will include in future gestural interface evaluations. Since our results also show that the material is highly sensitive to touch induced forces, there is potential to develop more complex interaction gestures that utilize the high force resolution unveiled by our experiments (see Figure 7).

The interaction area of our prototype is currently small, and due to the preliminary attachment with medical adhesive tape does not exploit the full potential regarding flexibility and conformability with skin. As the surface is composed of individual sensors, the sensory surface also lacks uniformity. Ideally, we would print multiple sensors onto the same substrate to avoid displacements of sensors and achieve a fully integrated second skin. For fixating the sensors, we would instead use skin-friendly glue or lock the surface in place mechanically (e.g., sleeve). Even though the texture of the PDMS patch is generally rated as comfortable, it is not perceived like human skin, which is one of our long-term objectives. Instead of using a smooth top layer, we think a cover layer should be structured or be combined with a second material to achieve tissue-equivalence [21]. Limitations in our qualitative evaluation are introduced by not having any conductive material printed onto the substrate which will affect the perceived looks of the secondary skin device.

Finally, due to the novelty of the material and simple implementation, we were only able to get initial results so far. Both the material and application scenario, are somewhat conservative and far behind the possible limits (e.g., maximum stretchability). Nevertheless, the presented results are encouraging and indicate that the material shows great potential for creating very robust and natural user interfaces.

6 CONCLUSION

Our paper presented the first artificial skin interface based on ultra-stretchable 3D-printed *capillary Ag-PDMS* sensors. We have initially explored the novel material and its use as a strain and pressure sensor for on-skin input interfaces. With an accuracy of 83% for six distinct gestures, we were able to show the general applicability of resistance based skin interfaces with ten subjects and low spatial resolution. Using this principle, instead of capacitive sensing requires less manufacturing overhead (three layers compared to five layers for capacitance-based setups). The overall approach shows systematic advantages considering that we can easily configure it. Consequently, resulting devices are generally more straightforward to implement, can be ultra stretchable and thin, and are fully bio-compatible due to the purely PDMS-based materials used.

We also conducted a qualitative material evaluation study. PDMS material, as such, feels comfortable and is easy to perform gestures on; however, we will have to adapt the surface texture for achieving tissue-equivalence.

In conclusion, we want to push the application limits further by making use of the full potential of capillary Ag-PDMS. The superior properties of the material and the manufacturing principle of devices are beyond the current state-of-the-art. Therefore, we believe that it will serve as the foundation for many new use cases and skin-like electronics.

7 FUTURE WORK

For future work, we believe that several research opportunities arise, and open questions remain.

A primary concern will be the development of more accurate interfaces. This topic includes both an improvement of the spatial resolution as well as classification accuracy. First, we could print higher-resolution resistance measurement pads. Potentially, capacitive on-skin touch sensor interfaces might be used as they have proven applicability and did work well in previous research with higher spatial resolution. Even though our classifier might be sufficient for day-to-day usage depending on the use case, we should expect a level of frustration introduced by having to repeat missclassified gestures. Therefore, we will investigate other, more robust classifiers for the interaction system. Nevertheless, we should recognize that the accuracy of our existing classification pipeline would likely also improve with higher resolution.

Another aspect concerns the wearability and attachment principle of the on-skin devices. We want to push the height limits of both cover and substrate PDMS layers and also conductive traces. Achieving minimum thickness will improve the slinkiness of devices and, ultimately, will make skin-applied interfaces more conformal. Additionally, we suggest adapting the cover layer to mimic the haptic feeling of human skin. Devices could almost be not perceivable by the user. Subsequently, we want to get rid of the medical adhesive tape by manufacturing wearable sleeves, which will be possible due to the ultra-stretchability of the new material.

Finally, more applications of the new material are interesting. For example, printing conductive traces in three dimensions could create a set of exciting applications. We are also interested in implanting a prototype device in a medical setting. More specifically, we are looking into using the approach for open, non-healing wounds. We might be able to achieve both the supervision of the healing process and also the active support of the healing itself. However, many open questions from a medical, as well as ethical standpoint, have to be resolved first.

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