Replicating the haptic perception capability of the human hand is an indispensable goal for intelligent robots and human–machine interactions. Multifunctional electronic skin (e-skin) sensors can be an ideal candidate to bridge the gaps among humans, robots, and the environment. Mutual interference of multistimuli and unconformable spatial distribution impedes the application of e-skin sensors. Hence, a large-area, hand-covering elastomeric e-skin sensor is proposed to imitate the human hand for multifunctional detection. Five multifunctional sensing units are designed on the finger-tips, and 15 pressure-sensing units are distributed on the finger phalanxes and palm to cover the main sensory area of the hand. A multilayer architecture is designed to improve the sensing performances and reduce the coupling interference during multifunctional detection. The e-skin sensor exhibits similarity to the human hand not only in shape but also in functionality, possessing pressure sensitivity of 0.025 V kPa$^{-1}$ in 0.1–120 kPa and temperature sensitivity of 0.38% °C$^{-1}$ in 20–70 °C. The performance of the e-skin sensor can meet the requirements of daily manipulations. Experimental studies on grasping objects with different grasping modes and object properties demonstrate the potential applications of the e-skin sensor for grasping haptic perception and human–robot interactions.

1. Introduction

Human hands take over the uppermost tasks in our daily life to feel and perceive rich information from the external environment by touching and grasping diverse objects and simultaneously inferring their properties.$^{[1–3]}$ Although remarkable progress has been achieved on understanding the physiological and neural mechanisms of the human sensation process,$^{[4]}$ replicating such tactile perception abilities of the human hand for robots and prosthetics is still undergoing great challenges. Flexible electronics have been proven to be a considerable candidate to break through the pivotal obstacles of interactions between humans and robots and are widely used in artificial intelligence,$^{[5,6]}$ human–machine interfaces,$^{[7,8]}$ virtual reality/augmented reality (VR/AR),$^{[9–11]}$ and health monitoring.$^{[12–14]}$ Among them, multifunctional electronic skins (e-skins) are attractive for their capabilities to simultaneously detect pressure, temperature, humidity, texture, etc.$^{[15–17]}$ One significant factor hindering the development of such e-skins is the compliance and conformability of the device to intimately attach to the surfaces of humans and robotics. Researchers have involved 2D and 3D network structures such as bulk,$^{[18]}$ helical,$^{[19]}$ and kirigami$^{[20]}$ structures to improve the stretchability of metal- and semiconductor-based electronic sensors. However, due to their inherent rigidity and brittleness, such devices are still suffering from insufficient conformability. The burgeoning of functional elastomers may be used to overcome these difficulties with their prominent performances of stimulus sensing, electrical conductivity, and intrinsic stretchability.$^{[7,21,22]}$

Most elastomers like thermoplastic polyurethanes (TPU),$^{[23]}$ polydimethylsiloxane (PDMS),$^{[24]}$ and fluorine copolymer$^{[25]}$ have generally good flexibility and stretchability but are usually...
intrinsically insulating. Diverse nanomaterials including graphene,[24] carbon nanotubes (CNTs),[26] metallic nanoparticles,[16] and nanowires[27] have been utilized to endow the elastomers with high electrical conductivity and sensing capability. To achieve multifunctional sensing, these functional elastomers have to be integrated into multiple sensing mechanisms, such as piezoresistive,[24] capacitive,[28] piezoelectric,[11] and triboelectric[29] principles for pressure-sensing, and thermal-resistive[30] and thermoelectric[31,32] principles for temperature sensing. For example, Park et al. reported a PDMS-coated microporous poly(pyrrrole) /graphene foam (PDMS/PPy/GF) composite and utilized its piezoresistive and thermoelectric effects to detect pressure and temperature, respectively.[33] The PPy/GF composite was fabricated as electrodes and worked as a supercapacitor for the power supply of the sensor. A unitary-material strategy to use only one material for multiple stimuli detection has also been proposed in a few studies.[16,34,35] Zou et al. synthesized a dynamic covalent thermost (polyimine) doped with silver nanoparticles and fabricated an e-skin to detect pressure, temperature, flow rate, and humidity.[16] Pressure sensing was conducted through capacitive changes and the other parameters were measured by the resistance response of the composite. However, the resistance changes of the e-skin for temperature, humidity, and flow rate were very close so it was difficult to discriminate the multiple stimuli effectively by the response of the developed e-skin. Hence, the coupling interference during simultaneous multifunctional detection is critical for elastomer-based multifunctional e-skins.

The structure design of the e-skin is also crucial as it greatly affects its sensing performance. 2D spiral[36] and meandering[17] patterns and 3D microspheres,[18] micropillars,[38] and nanorods[39] have been proposed for the structural design of e-skins to enhance the sensitivities. Different kinds of architectures also can be incorporated into one sensor to achieve multifunctional sensing. For instance, Khatib et al. designed several 2D line patterns using different materials as sensing components to detect pressure, temperature, and pH.[15] Jung et al. utilized 3D micropyramid arrays coated with CNT ink for pressure sensing and printed thermoelectric flat paper-based 2D patterns to measure temperature.[40] Most studies are focusing on the sensitivity enhancement with micro- and nanoscale architectures; however, there is still a lack of research on multistimulus decoupling structures for multifunctional e-skins. Circuitry-based strategies like differential feedback circuit[41] and Wheatstone bridge[42] have been integrated into the e-skin to achieve decoupling of pressure and temperature signals while having complicated structural design and fabrication. A more straightforward and efficient method is to propose a novel structure design that can help separate the pressure and temperature stimuli, so the coupling interference of the e-skin can be remarkably reduced. Moreover, current works of e-skins have mainly involved single- or few-point detection and some have reported large-scale rectangular-shaped arrays,[18,32,34,41,44] which are difficult to be conformedly integrated with the human or robotic hand. The dense but nonuniform distribution of the mechanoreceptors in human hand is a good inspiration for the distribution design of sensor arrays, and researchers have made efforts in imitating the mechanoreceptors to fabricate tactile sensor arrays with high density.[45] While such sensor arrays usually involved complicated fabrication processes and measurement equipment, the redundant tactile information is also a significant concern. Thus, a proper distribution design of the sensor array that can obtain adequate tactile information with a small number of sensing units will be attractive for the design of e-skin sensor. In general, a large-area, hand-imitating stretchable e-skin sensor with properly distributed and multifunctional sensing capabilities is urgently needed and remains to be further studied.

Herein, we propose a large-area hand-covering fully elastomeric e-skin sensor for distributed and simultaneous detection of pressure and temperature during human grasping applications. Multifunctional sensing units, each containing a pressure component and a temperature component, are designed on the fingertips of the hand-shaped e-skin sensor. There is one pressure unit, which consists of a single pressure component, on each finger phalanx and an array of seven pressure units on the palm. Thus, the e-skin sensor contains five multifunctional sensing units and 15 pressure units to cover the main sensory area of the hand. For enhancing the sensing performance and decoupling of the multifunctional detection from the structural perspective, the multifunctional sensing unit is designed as a multilayer architecture. The middle patterned sensing components are sandwiched by the upper truncated circular cone bump and lower indurated patterned substrate. Graphene/CNT/ silicone rubber (SR) nanocomposite is prepared and utilized to fabricate the pressure component, and silver nanoflake (AgNF)/SR nanocomposite is used to fabricate the temperature component and stretchable electrodes. The proposed e-skin sensor exhibits pressure sensitivities of 0.025 V kPa−1 in 0.1–60 kPa, and 0.008 V kPa−1 in 60–120 kPa, and temperature sensitivity of 0.38° C−1 in 20–70 °C. The fast response and dynamic stability make the e-skin sensor able to meet the demands for tactile perception during grasping manipulations in daily life. Besides, the coupling interference between the pressure and temperature components can be reduced to less than 15%. Further wearable applications have been implemented by integrating the e-skin sensor with the human hand. Diverse experimental studies are conducted to grasp objects with different grasping modes, different properties, and temperatures. The proposed e-skin sensor is demonstrated to be feasible and reliable for multifunctional detection and exhibits great potential in the research of human grasping motions and human–robot interactions.

2. Results and Discussion

2.1. Large-Area Hand-Covering Fully Elastomeric E-Skin Sensor

Most haptic perceptions are achieved through the sensory receptors distributed on the human hand, where fingertips are the upmost region to perceive diverse information like contact, pressure, temperature, moisture, etc.[13] Inspired by the structure and characteristics of the human hand, we proposed a fully elastomeric e-skin sensor with multiple sensing units, which can cover the whole hand to achieve large-area tactile sensing. As shown in Figure 1, the multifunctional units with pressure- and temperature-sensing capabilities are arranged on five fingertips. Each multifunctional unit contains one pressure component in the center and one temperature component around it. On the
other regions of the hand-shaped e-skin sensor, there is one pressure unit consisting of a single pressure component in the central area of each finger phalanx and an array of seven pressure units in the palm. Thus, the designed e-skin sensor has five multifunctional units and 15 pressure units. Figure S1, Supporting Information, shows the dimensions of the units in the e-skin sensor, indicating that the units possess a similar size to the fingers, so the distributed e-skin sensor can properly cover the main tactile-sensing area of the fingers and palm of the human hand.

Comparing some related works that reported pressure sensor arrays with higher unit density,[2,45] the proposed e-skin sensor can meet the demands of distributed multifunctional tactile sensing for daily object grasping and operating with fewer units.

For the structure of each unit, taking the multifunctional unit on the fingertip as an example, as shown in Figure 1b, it consists of a truncated circular cone-shaped PDMS bump, sensing layer, indurated embedded layer, and PDMS substrate. The pressure component is a pattern of concentric circular arcs, imitating the fingerprint of the human hand to get the utmost of the sensing area. The temperature component is a series of serpentine short arcs, which can help decrease the strain of the component when the unit is pressed and deformed. The indurated pattern is designed to enhance the sensing performance of the multifunctional unit. The indurated pattern consists of two parts: four indurated beams under the pressure component can enhance the pressure sensitivity, and the indurated annular pattern beneath the temperature component is used to improve temperature-sensing performance. The structure of the pressure unit on the phalanx and palm is almost the same but it has no temperature component and the corresponding indurated annular pattern, as shown in Figure 1c.

To achieve the high conformability and multifunctionality of the e-skin sensor, elastomeric nanocomposites are utilized to perceive multiple stimuli and avoid the usage of traditional metal electrodes. The bump and substrate comprise polydimethylsiloxane (PDMS). In our previous work, the addition of nano-SiO2 has been proven to increase the Young’s modulus of PDMS.[44] So, the indurated embedded patterns are made of nano-SiO2-doped PDMS (PDMS–SiO2) to increase the hardness (Young’s modulus is shown in Figure S2, Supporting Information). The pressure component consists of graphene, CNT, and SR (GR/CNT/SR) nanocomposite to achieve a high piezoresistive effect under tensile strains. The temperature component and electrodes are made of AgNF and SR nanocomposites, which exhibit high conductivity and thermal resistance effect.

The pressure- and temperature-sensing principles of the multifunctional unit are shown as in Figure 1d,e. As shown

![Figure 1. Schematic view of the large-area hand-covering fully elastomeric e-skin sensor. a) The structure of the e-skin sensor. b) The structure of the multifunctional unit. c) The structure of the pressure unit. d,e) The pressure- and temperature-sensing principles of the multifunctional unit, respectively.](image-url)
in Figure 1d, according to Saint-Venant’s theory, when the pressure is applied to the multifunctional unit through the grasped object, the pressure is transmitted and mainly distributed in the center area of the lower surface of the truncated cone-shaped bump, and then mostly applied to the pressure component, whereas the temperature component in the outer area of the bump will not be greatly affected by the applied pressure. To achieve pressure sensing, the piezoresistive mechanism is selected for the pressure component to convert outer pressure into resistance changes. The concentric long arcs of the pressure component will be squeezed into the air gaps of the lower four indurated beams, as shown by the yellow region in Figure 1d, and generate apparent tensile strains. Thus, the resistance of the pressure component changes, the pressure can be detected, and the pressure sensitivity is improved. For temperature sensing, when the temperature is applied to the top surface of the bump, the heat will be converted to the temperature component through the bump to induce the thermal resistive effect of the nanocomposites of the component, as shown in the red region in Figure 1e. When high temperature is applied to the component, the conductive nanocomposite of the component will swell due to thermal expansion effect, and the nanofillers inside the nanocomposite will separate from each other to break the conductive networks and increase resistance. At the same time, the intrinsic resistance of the nanofillers will also increase. Thus, the temperature rise will lead to the increases in the resistance of the temperature component. The temperature component is sandwiched by the upper bump and lower indurated annular pattern, so it is difficult for the pressure to deform the temperature component. Thus, the temperature component will mainly be affected by temperature, and the temperature can be well detected. As for the pressure component, although it is influenced by both temperature and pressure, the resistance change induced by pressure is much greater than that induced by temperature, so the influences of temperature can be negligible. Therefore, the multilayer structural design of the multifunctional unit that we proposed can not only enhance the pressure sensitivity but also reduce the coupling interference between pressure and temperature detection. As for the stretchable electrode, it has the same material as the temperature component, but the connecting area between the electrodes and component is very small (the rectangular ends of the component shown in Figure 1), so the temperature change will not generate much effects on its resistance.

To unite the separate units into an integrated array for the convenience of measurement, stretchable electrodes are utilized to connect all these sensing units, as shown in Figure S3, Supporting Information. In addition, the terminals of the electrodes have been designed as a flexible printed circuit (FPC) to connect with peripheral measurement equipment.

2.2. Characterization of the Conductive Nanocomposites

The conductive nanocomposites were prepared using a simple mechanical blending method. As shown in Figure 2a, GR, CNT, and SR with a mass ratio of 4:2:100 were uniformly blended in a planetary mixer to obtain a homogeneous and pasty mixture. A 100 μm-thick layer of PDMS was coated onto a glass wafer covered by polyimide (PI) film. Then, the above GR/CNT/SR mixture was coated onto the cured PDMS film by the screen-printing method. After the mask was removed and the mixture was cured under heating, the specimen with dimensions of 20 × 1 × 0.2 mm was cut and peeled off. The AgNF/SR nanocomposite specimen with a mass ratio of 2:1 was also prepared in the same way and the detailed preparation process is shown in the Experimental Section.

The morphologies of the nanocomposites were investigated through a scanning electron microscope (SEM). Figure 2b,e shows the cross-sectional image of the GR/CNT/SR and AgNF/SR nanocomposites under a magnification of 1000×. The neat PDMS substrate and nanocomposite layer can be easily distinguished, and there are almost no delamination boundaries between the two layers, indicating that the nanocomposites can be tightly bonded with the substrate. Figure 2c,d,f,g shows the morphologies of the GR/CNT/SR and AgNF/SR nanocomposites under different magnifications, respectively. As shown in Figure 2c, we can see that GR plates are uniformly distributed in the SR matrix without obvious aggregations. Under a higher magnification of 10 000× in Figure 2d, GR plates of several micrometers can be seen to be wrapped by the SR matrix. Among the GR plates, CNTs with nanoscale diameter and length of micrometers are embedded and connect the gaps between the adjacent GR plates like a bridge, resulting in denser and more uniform conductive networks. From Figure 2f,g, it is obvious that the AgNFs of several micrometers are uniformly dispersed in the SR matrix as well.

Then the electromechanical properties of the GR/CNT/SR and AgNF/SR nanocomposites were tested. The specimens were stretched under a uniaxial universal testing machine and their electrical properties were recorded by a multimeter. Before stretching, the GR/CNT/SR nanocomposite shows a conductivity of 0.035 ± 0.013 S cm⁻¹, which is higher than the GR/SR nanocomposite with a mass ratio of 4:100 reported in our previous study. The AgNF/SR nanocomposite possesses a much higher conductivity of 200 ± 13 S cm⁻¹. Under stretching, the resistances of GR/CNT/SR and AgNF/SR nanocomposites both monotonically rise with the increase in tensile strains, as shown in Figure 2h. However, the GR/CNT/SR nanocomposite performs with a much greater increasing rate than the AgNF/SR, that is, the GR/CNT/SR nanocomposite has a larger gauge factor (GF). The GF can reflect the sensitivity of the nanocomposites under tensile or compress strains and can be calculated as $GF = (\Delta R/ R_0) / \varepsilon$, where $\Delta R$ is the relative change in resistance of the specimen, $R_0$ is the original resistance, and $\varepsilon$ is the applied tensile strain. From the curves in Figure 2h, the GF can reach about 3210 at the maximum strain of 80% and a relatively high GF at a small strain of 30% is 42 for the GR/CNT/SR nanocomposites. The detailed electromechanical properties of the AgNF/SR nanocomposite are shown in Figure S4, Supporting Information. The AgNF/SR nanocomposite can be stretched to about 300%, and the resistance changes of the electrode are shown in Figure S4b,c, Supporting Information. The nanocomposite shows a very small resistance change, and the GF is 1.8 at $\varepsilon = 30\%$, and 7.5 at $\varepsilon = 100\%$, demonstrating that the nanocomposite can maintain high conductivity under stretching. When the nanocomposite is stretched to over 200%, the resistance increases dramatically as the conductive network breaks.
During the applications of the electrodes in the e-skin sensor, it will not be stretched so obviously, so the resistance can be relatively stable and small during practical applications. Figure S4d, Supporting Information shows the cyclic test of the AgNF/SR nanocomposite, showing good reproducibility. The difference between the conductivity and GF of these nanocomposites demonstrates the feasibility to use the GR/CNT/SR nanocomposite for the pressure component and the other for the stretchable electrodes. Thus, the resistance change of the AgNF/SR electrodes will not generate much influence on the GR/CNT/SR pressure component. In addition, we can choose the AgNF/SR nanocomposite to work as the temperature component due to its thermal resistance effect, which has been proven by some related works.[15,16] Figure 2i shows the relative change in resistance of the GR/CNT/SR nanocomposite under cyclic test.

2.3. Fabrication Process of Large-Area Hand-Covering E-Skin Sensor

To fabricate the e-skin sensor, a layer-by-layer screen-printing procedure was developed. Taking the multifunctional unit as an example, Figure 3 shows the entire fabrication procedure. In Figure 3a, the upper bump layer and sensing layer were first fabricated. First, the uncured PDMS (the mass ratio of the base resin and curing agent is 10:1) was poured onto a customized aluminum mold of the upper bump and defoamed in a vacuum chamber. After the PDMS was cured at 80°C, a steel mask with hollows of the pressure component was covered onto the bump, and the prepared GR/CNT/SR nanocomposite was poured on the mask and screenprinted into the hollows of the mask with a scraper. After removal of the mask, the pressure component was left on the surface of the bump and cured at 80°C. Then,
the temperature components and stretchable electrodes were successively screen printed in the same way. Thus, the upper bump layer with the sensing layer was fabricated. Second, as shown in Figure 3b, the bottom substrate layer was mold casted through the corresponding substrate mold with PDMS (10:1). Indurated PDMS was prepared by mixing PDMS base resin, curing agent, and nano-SiO$_2$ with a mass ratio of 5:1:0.5. Higher mass ratio of the curing agent can also increase the hardness of the PDMS. The indurated pattern layer was fabricated by screen printing indurated PDMS onto the cured substrate. After curing, PDMS (10:1) was screen printed to fabricate the bonding layer between the sensing layer and the substrate layer. The PDMS was heated at 80°C for 0.5 h to reach a half-cured state. Then, the whole substrate layer was peeled off from the mold, aligned with the upper sensing layer, and then assembled. The half-cured PDMS remained highly sticky to bond these two layers intimately. Finally, the fabricated e-skin sensor was peeled off from the mold, and its optical image is shown in Figure 3c. Figure 3 shows the schematic view of the fabrication process; the actual molds and steel masks used are shown in Figure S5, Supporting Information.

Furthermore, to distinguish the multiple units in the e-skin sensor in the following discussions, the units have been named according to their positions on the hand. As shown in Figure 3c, the temperature components on five fingertips are named as $U_1$, $U_2$–$U_5$, from the thumb to the little finger. The pressure components on the thumb are labeled as $T_1$ and $T_2$. Similarly, the pressure units on the index, middle, ring, and little finger are $I_1$–$I_5$, $M_1$–$M_3$, $R_1$–$R_3$, and $L_1$–$L_2$, respectively. The units in the palm are named as $P_1$–$P_7$.

2.4. Multifunctional Sensing Performance of Large-Area Hand-Covering E-Skin Sensor

The multifunctional sensing performance of the fabricated e-skin sensor was investigated under a customized calibration system, which is shown in Figure S6, Supporting Information. The multifunctional unit of the e-skin sensor was mounted on a displacement platform, and a 3D-printed loading bar was fixed with a high-precision load cell on the z-axis platform to apply pressure to the unit. The e-skin sensor was connected with a peripheral FPC connector through its electrode terminals. Through the FPC connector, the pressure components in the e-skin sensor were connected to a scanning circuit, and the temperature components were connected to a multimeter. The connection configuration and principle of the scanning circuit are shown in Figure S7, Supporting Information. The pressure components were linked with the stretchable electrodes to form a 2 x 10 resistance array for convenience to connect with the scanning circuit. The circuit can convert the resistance change of the pressure components into voltage outputs. The detailed information will be illustrated in the Experimental Section.
The multifunctional sensing performance of the e-skin sensor is shown in Figure 4. First, the pressure sensing performance of the sensing unit is shown in Figure 4a–f. The pressure calibration result is shown in Figure 4a, the unit performs with two linear-stage sensitivities: \( S_{P1} = 0.025 \text{ V kPa}^{-1} \) from 0.1 to 60 kPa and \( S_{P2} = 0.008 \text{ V kPa}^{-1} \) from 60 to 120 kPa. The lower sensitivity under higher pressure can be caused by the GR/CNT/SR patterns of the pressure component in Figure 1 in contact with the bottom substrate after being squeezed into the gaps among the indurated beams under external pressure, and the strain of the component will have a slower increase, indicating that it gradually gets into the saturation region. For the e-skin sensor without the indurated layer, its pressure sensitivity is also shown in Figure 4a, which is much lower with a small value of \( S_C = 0.006 \text{ V kPa}^{-1} \), verifying the effectiveness of the proposed indurated layer structure. A cyclic test from 0.1 to 120 kPa was conducted, and results are shown in Figure S8, Supporting Information; the loading and unloading output curves of the 1st, 50th, 100th, and 200th cycles have good consistency, so the sensing unit can work steadily in the whole sensing range. Figure 4b,c shows the responses of the unit under cyclic loads with different magnitudes and frequencies, respectively. Under different pressures of 10, 20, 50, and 100 kPa, the unit showed regularly changed voltages, which are generally consistent with the calibration results. During each loading process, when the pressure is applied to the unit, the pressure maintains about 1 s and the voltage shows a small decline, which may be induced by the viscoelasticity of the SR and will be discussed later. In Figure 4c, the unit response has a modest rise with the increasing frequency from 0.1 to 2 Hz but still matches with the tested values in Figure 4a. This slight frequency-dependent phenomenon mainly results from the greater internal stress of such polymers under higher strain rates, as the mobility of the polymer chains reduces with increasing strain rates and the stiffness of the polymer increases due to the chain entanglement effect.[26,48] Generally, this phenomenon will not generate much influence, and the detection frequency range of 0.1–2 Hz can meet the major demands in our daily object’s manipulation. Hysteresis is also an important parameter for pressure sensors. For the proposed e-skin sensor, as shown in Figure 4d, the...
hysteresis is relatively larger under high-pressure loads. The maximum value is about 23% under 100 kPa, and it is much smaller under lower pressures. Figure 4e shows the dynamic response performance of the unit. A square-wave pressure of 50 kPa is repeatedly applied to the unit, the load maintains 3 s, and there is also a 3 s pause after each cycle. As shown in this figure, the square load (blue curve) reaches 50 kPa and slightly declines to a plain stage, mainly caused by the stress relaxation of the flexible structure of the unit. The unit performs with a considerably fast response with a loading time of \( T_L = 100 \text{ ms} \) and unloading time of \( T_{UL} = 200 \text{ ms} \). It is worth noting that during each pressure-maintaining stage, there is an overshooting of voltage, which can be mainly attributed to the viscoelasticity of the SR matrix. The later voltage relaxation effect is also observed and can be induced by the inherent creep behavior of such elastomers. After unloading, the output of the unit does not recover to its original value, and the unloading time is also slightly longer than the loading process. This can result from the hysteresis of the nanocomposites and irreversible destruction of the conductive networks inside the nanocomposites after pressure loading. The above phenomena have also been reported and illustrated in other research.[26,50] Finally, a multiple loading test is conducted for 1000 cycles under 20 kPa. As shown in Figure 4f, the unit exhibits good repeatability with a negligible decrease in the peak and original voltage values, which have been illustrated as the creep behavior of the viscoelasticity nanocomposites.

The temperature-sensing performance has also been tested through a heating platform. The resistance of the temperature component was measured and recorded by a multimeter. We have calibrated the temperature-sensing performance of the components with different AgNF ratios to select an optimized nanocomposite. As shown in Figure 4g, the components with low AgNF contents show relatively higher temperature sensitivity, but the resistance change during heating is not stable and the linearity is relatively poor. We have chosen the ratio of SR:AgNF as 1:2 for its higher sensitivity and better linearity. The selected temperature component exhibited a sensitivity of 0.38% °C⁻¹ in a wide range of 20–70 °C to cover the normal object manipulation demands, higher than some multifunctional e-skins that have been reported.[16,18,35] A multicycle test was also implemented to demonstrate the sensing stability and accuracy. We can see from Figure 4h that the resistance of the component circularly changes when the temperature varies from 20 to 40, 55, and 70 °C, respectively. The relative change in resistance for every group is consistent with the calibration curves. Figure 4i shows the hysteresis of the temperature component under different temperatures. The almost coincident curves of the heating and cooling process indicate that the component has very small hysteresis, ensuring temperature-sensing accuracy. A 50-cycle test from 20 to 70 °C was conducted and the result is shown in Figure 4j, demonstrating that the temperature component exhibits good reproducibility and stability during long-term experiments. Furthermore, the pressure and temperature-changing tendency during the entire testing process for a single cycle from 20 to 70 °C is shown in Figure 4l. As the platform has a fast heating rate but a very slow cooling rate, the component performs with almost the same changing trend, which shows that the component has good reliability of temperature sensing.

We have also tested the resistance changes of the temperature component under different pressures. As shown in Figure S9, Supporting Information, the component with the annular indurated pattern exhibits much lower resistance changes under pressures from 0 to 120 kPa, indicating that the indurated layer can reduce the influences of the pressure on the temperature component.

For such multifunctional e-skin sensors, the coupling interference between pressure and temperature has always been a significant concern, which greatly affects the sensing accuracy during practical applications. We apply different temperature and pressure loads to the multifunctional unit to investigate the sensing performance change of the unit. The results are shown in Figure S10, Supporting Information. In Figure S10a, Supporting Information, we tested the performance of the pressure component under different temperatures, and the pressure sensitivity has small decreases of 7.8% under 45 °C and 15.1% under 70 °C. The temperature-sensing performance is also calibrated when external pressures are applied. As shown in Figure S10b, Supporting Information, the temperature sensitivity exhibits a modest reduction of 8.4% under 40 kPa and 16.9% under 80 kPa. The coupling interferences between pressure and temperature are beyond 15% under large detection magnitudes, but it is still considerable under most cases, which is attributed to the structural design of the multifunctional sensing unit. Such results have also verified the effectiveness of the proposed multifunctional e-skin sensor.

A comparison of the performances of e-skin sensors in our study and some other related works is conducted. As shown in Table S1, Supporting Information, the sensing mechanism, distributed array structure design, multifunctional sensitivity, sensing range, and response time are selected to be compared. We can see that the proposed e-skin sensor exhibits the main innovation on the human-imitated distributed design with multifunctional sensing capabilities, which performs with favorable multifunctional sensitivities and sensing ranges.

2.5. Object Grasp Applications with Different Grasping Modes

To further demonstrate the multifunctional sensing performance of the proposed e-skin sensor, we have conducted wearable applications of the e-skin sensor to grasp different objects. Figure S11, Supporting Information, shows the optical image of the e-skin sensor worn on the human hand. The e-skin sensor was connected with an FPC connector to link with the peripheral scanning circuit (Figure S7, Supporting Information) for unit output measuring. From this image, we can see that the units in the e-skin sensor can properly cover the main sensory regions of the hand. Pressure is the main as well as the most direct information during the haptic perception, so we have conducted pressure-sensing experiments by grasping different types of objects first. According to current studies, human grasping modes can be classified into 33 types by the features of different grasping motions.[31] In our research, we have selected the five most common grasping modes to grab different typical objects. As shown in Figure S5, we implemented large-diameter (LD) grasping mode to grab a big cylinder, power-sphere (PS) mode for a ball, parallel-extension (PE) mode for a cube, ring (RI) mode...
for a round plate, and quadpod (QP) mode for a smaller ball. The output voltages of all the pressure units have been recorded by the scanning circuit and the values at the steady-grasp stage were plotted as a hand-shaped nephogram to intuitively reflect the pressure distribution, as illustrated in the lower half of Figure 5.

For the LD mode grasping a cylinder, most of the pressure units except P₅ at the bottom of the palm were in contact with the cylinder when the hand steadily grasped it. As can be observed from the nephogram in Figure 5a, the main contacting units are T₁ on the thumb, M₂, M₃ on the middle finger, R₁ on the ring finger, and P₁, P₆ on the palm, as these units generate larger voltages. Thus, we can infer that the primary regions that we used the LD mode to grasp a cylinder are the thumb, middle, and ring finger, and the upper half of the palm. In the same way, the main area when we used PS mode for a ball is distributed on the thumb, index, middle, ring finger, and the top right part of the palm. When we grasped a cube with PE mode, the applied pressure was uniformly distributed on the fingertips of the thumb and the distal two phalanxes of the other four fingers. In addition, due to the smaller weights of the round plate and small ball, the pressure for RI and QP modes is also smaller. From the nephograms of different modes, we can intuitively distinguish the differences among the modes and objects.

Then, we further investigated the LD grasp process for the cylinder, and the output voltages of all the units are shown in Figure 6a. The upper row of Figure 6a plots the output curves of the units T₁₋₂, I₁₋₃, M₁₋₃, and P₁₋₂, whereas the lower row shows the curves of R₁₋₃, I₁₋₂, and P₃₋₇. It is inconvenient for us to analyze the whole change trends of all the units clearly, so we selected nine featured moments of the grasping process to plot the nine nephograms in Figure 6b, that is, nine typical frames from T₂ to T₉. In these three frames, the releasing process took about 1 s, and the distal two phalanxes of each finger got out of contact earliest with the object, and the palm got separated at last. In the last end motion frame T₉, the voltages of the units did not fully recover to the original zero state with small magnitudes of 0–0.15 V, indicating the slight irreversible destruction of the conductive networks inside the nanocomposites of the units, as mentioned earlier. However, this will not generate much interference for us to study the grasping motions. Moreover, besides the above information on the quasistatic contact situation during grasping, more dynamic details can be learnt from the nephograms. The darker colors in the nephogram indicate that the corresponding regions on the hand apply greater grasping force, so we can consider these areas as the main contact region (MCR), as shown by the red dashed circles in Figure 6b. During the contact process (T₂–T₄), the MCR started from Region I at T₂ and then transmitted upward to Region II at T₃. In this period, the transmission of MCR can demonstrate that the cylinder object in the hand has an upward slip or slip tendency. From the MCR transmission of I→II→III, we can preliminarily analyze the relative movement tendency between the contact regions of the hand and the object. Furthermore, if we continue to extract more and denser frames from T₂ to T₉, this tendency can be studied more meticulously in a dynamic and intuitive perspective. Comparing those reported methods on detecting slippage by single-pixel tactile sensors, this can be also a decent strategy for slippage detection. Generally, the proposed e-skin sensor has

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**Figure 5.** Wearable grasping experiments and the corresponding voltage nephograms with different grasping modes. a) Large-diameter grasping mode for a cylinder. b) Power-sphere grasping mode for a ball. c) Parallel-extension grasping mode for a cube. d) Ring grasping mode for a round plate. e) Quadpod grasping mode for a small ball.
great potentials to acquire more abundant and lossless distributed tactile information, such as contact, pressure, and slippage, to get into deep research on the human-grasping mechanism.

### 2.6. Object Grasp Applications with Different Object Properties

Discrimination of object properties like texture (smooth/coarse), stiffness (hard/soft), and temperature is a significant function for human hand grasping. Based on the proposed multifunctional e-skin sensor, we have implemented manifold grasp experiments with different object properties to demonstrate its multifunctional sensing performance. In Figure 7a–d, plastic ball, rubber ball, tennis ball, and hollow rubber ball are selected as the grasp objects and they have successively descending stiffness. The QP mode was utilized to grasp the objects for twice and the output voltages of the units T1, I1, M1 on the thumb, index, and middle fingers were plotted. As shown in Figure 7a, the voltage in the steady grasp stage (indicated by the blue dashed rectangle) exhibits slight relaxation as the strain rate of the nanocomposites during the grasp is much slower than the calibration test in Figure 4. When the ball was released, the voltage directly decreases. In Figure 7b, when we grasped a little softer rubber ball, there is a more obvious voltage decline in the steady state, as the softer ball induced stress relaxation effect, and the creep behavior of the pressure component became notable. In addition, there is a small rise in voltage when the grasp was finished (indicated by the blue dashed circle), which can be attributed to the shape springback of the deformed rubber ball. The voltage relaxation in the steady stage and rise in the releasing stage is even more apparent when the tennis ball and hollow rubber ball with lower stiffness were grasped in Figure 7c,d. It is noteworthy that during the grasp of the hollow rubber ball, as the ball is quite soft, the voltage relaxation of the units is so prominent that it begins to decline with almost no steady stage. Furthermore, the voltage rise when releasing the ball is also remarkable as the ball withstood large deformation during grasp and exhibited obvious springback, as shown in the image in Figure 7d. In overall consideration, the voltages detected by the e-skin sensor will

![Figure 6. The whole process of the LD mode for grasping a cylinder. a) Output curves of all the 20 pressure components in the e-skin sensor during the grasping process. b) Extracted nine typical frames of the grasping process.](image-url)
be beneficial for us to discriminate the stiffness of the grabbed objects. During daily practical applications, the unit array in the sensor can help us achieve more accurate and reliable perception through multiple results of all the contact units, as there may be some deviations in the results of only one or few units.

Grasping applications for simultaneous detection of pressure and temperature have been conducted as well. As shown in Figure 7e, we grabbed an empty beaker, a beaker with 50 mL of room temperature (RT), and a beaker with 100 mL of 50 °C hot water. When the empty beaker was grasped, the pressure components $T_1$ and $I_1$ on the thumb and index fingertips generate small output voltages with slight relaxation, consistent with the results of grasping the plastic ball. Meanwhile, the temperature components $U_1$ and $U_2$ at the same position exhibit tiny fluctuations, indicated by the blue and green curves in Figure 7c. When we grasped the beaker with 50 mL RT water, the pressure increases as the weight of the beaker is larger, and the undulation of the temperature curves of $U_1$ and $U_2$ is mainly caused by the pressure and deformation of the temperature components during the grasp but is negligible. Finally, when the beaker with 100 mL hot water was grabbed, the larger pressures are detected and the two temperature components exhibit response of resistance change to the higher contact temperature. The values are a little smaller than the temperature calibration results, as there is slight coupling interference induced by pressure during the grasping process, which is shown in Figure S10, Supporting Information.

From all the earlier wearable applications of grasping diverse objects with different modes, different stiffnesses, and different temperatures, the results have effectively demonstrated the multifunctional sensing performance of the proposed e-skin sensor. The abundant distributed information that we obtain from the grasping process can help us to better study human grasping motion characteristics and human–robot interactions.

3. Conclusion

In summary, we presented a novel large-area hand-covering fully elastomeric e-skin sensor for simultaneous distributed pressure

**Figure 7.** Wearable applications to grasp objects with different properties. Grasping sphere objects with sequentially decreasing stiffness: a) plastic ball, b) rubber ball, c) tennis ball, and d) hollow rubber ball, respectively. e) Grasp beakers with water of different temperatures.
and temperature sensing. The multifunctional sensing units contain a pressure component and a temperature component and are in the five fingertips of the hand. Single pressure units are distributed on the finger phalanges and palm to cover most sensory areas. A sandwiched structure, containing upper bump, middle-sensing layer, and lower indurated patterned substrate, is proposed for the sensing units to improve the sensitivity and reduce the coupling interferences between pressure and temperature. The GR/CNT/SR nanocomposite was used for pressure sensing with ultrahigh GF of \( \approx 1210 \), whereas the AgNF/SR nanocomposite served as the temperature component and stretchable electrodes due to its extremely low only GF of 10 at a large tensile strain of 120%. Based on the structural design and nanocomposites, the e-skin sensor exhibited a two linear-region pressure sensitivity of 0.025 V kPa\(^{-1}\) in 0.1–60 kPa and 0.008 V kPa\(^{-1}\) in 60–120 kPa and temperature sensitivity of 0.38% °C\(^{-1}\) in 20–70 °C. The e-skin sensor performed with dynamic fast response, good repeatability, and low mutual interference. Plentiful experimental studies were conducted by integrating the e-skin sensor with the human hand to grasp different objects. The object grasping was implemented under five typical grasping modes and with diverse object properties. The results have demonstrated that the proposed e-skin sensor was capable of simultaneously detecting pressure and temperature and showed the possibility to discriminate the properties of the objects. The proposed approach to design and fabricate the e-skin sensor can construct a human-hand-imitated sensor array with the distributed multifunctional sensor array. The current limitation mainly lies in that the single-unit structures and dimensions have to be further optimized to get an optimal multifunctional sensing performance with higher sensitivity, wider sensing range, and lower coupling influence, which will be the focus of our future work. In overall consideration, the e-skin sensor has provided fine feasibility for studies of human-grasping mechanisms and human–robot interfaces.

4. Experimental Section

**Materials and Reagents:** Polydimethylsiloxane (PDMS, Sylgard 184) was purchased from the Corning Co., Ltd. The SR used as the matrix of the nanocomposites was bought from the Zhonghao Chenguang Research Institute of Chemical Industry. Graphene was supplied by the Sixth Element Materials Technology Co., Ltd. The CNT, AgNFs, and nanosilicon dioxide were supplied by the Nanjing XFNANO Materials Tech Co., Ltd. During the nanocomposite preparation, tetrahydrofuran (THF, Sinopharm Chemical Reagent Co., Ltd) and polyvinylpyrrolidone (PVP, average molecular weight of 10 000, Sigma-Aldrich Trading Co. Ltd.) were selected as the solvent and dispersant to adjust the viscosity of the nanocomposites and promote the dispersion uniformity of the nanofillers.

**Preparation Method of the Conductive Nanocomposites:** For the GR/CNT/SR nanocomposite, the graphene and CNT were added into 0.5 g SR with a mass ratio of GR:CNT:SR = 4:2:100. 0.25 g THF was added and the mixture was uniformly blended in a planetary mixer (AR 100, Thinky Corporation) for 3 min to obtain a homogenous and pasty mixture. To fabricate the GR/CNT/SR specimen, PDMS with a mass ratio of base resin: curing agent = 10:1 was mixed and degassed in the planetary mixer. As shown in Figure 2a, a glass wafer covered by a PI film was coated with a 100 μm layer of PDMS with a film applicator. The PDMS film was cured at 80 °C for 2 h. A 100 μm steel mask with rectangular hollows of 20 × 1 mm was covered on the PDMS film. The aforementioned GR/CNT/SR nanocomposite was coated onto the mask with a scraper blade. After removal of the mask, a 100 μm layer of the GR/CNT/SR nanocomposite was left on the PDMS film and cured at 80 °C for 1 h. Then, the specimen was cut and peeled off together with the PDMS substrate from the glass wafer. For the AgNF/SR nanocomposite, AgNF and PVP as the dispersant were added into SR with a mass ratio of 200:2:100 and blended in the planetary mixer. The AgNF/SR specimen was fabricated by the same method as the GR/CNT/SR specimen.

**Characterization of the Conductive Nanocomposites:** The morphologies of the two nanocomposites were obtained by an SEM (S-3700N, HITACHI) with an accelerating voltage of 3000 V. The electromechanical properties were studied using a uniaxial universal testing machine (UTM 2203, Shenzhen Suns Technology Stock Co., Ltd.) Conductive tapes were linked to the terminals of the specimens with silver paste to connect with a digital multimeter (34465A, Keysight) for resistance measurement.

**Characterization of the E-skin Sensor:** The multifunctional sensing performance of the e-skin sensor was tested on the customized calibration system, as shown in Figure S6, Supporting Information. The sensor was mounted on the displacement platform. A high-precision load cell (Nano 17, ATI Industrial Automation, force resolution of 0.01 N) was fixed on the z-axis platform and a 3D-printed loading bar was fixed on the load cell to apply pressures to the units in the e-skin sensor. The e-skin sensor was connected with an FPC connector and the pressure component electrodes were connected to a customized scanning circuit, while the temperature component electrodes were linked to the multimeter. As shown in Figure S7, Supporting Information, the pressure components were linked as a 2 × 10 array for convenience to connect with the scanning circuit. The circuit was designed to transform the resistance of each unit into voltage through the voltage divider principle (Figure S7a,c, Supporting Information). A digital signal processor (DSP, TMS320F2812, Texas Instruments) was utilized to obtain the voltages and send the data to PC as well as control the analog switch to select the units sequentially for scanning measurement. A 3.3 V voltage was input to the analog switch by the DSP to measure the voltages of the corresponding unit. As for temperature sensing, the multifunctional unit was mounted on a heating platform and a thermocouple for real-time temperature measurement was attached to the unit. The temperature of the unit was recorded by a multichannel thermometer (AT 4708, Changzhou Applent Instruments) through the thermocouple and the resistance was measured by the multimeter.

The experiments involving human subjects have been performed with the full, informed consent of the volunteers.

**Supporting Information**

Supporting Information is available from the Wiley Online Library or from the author.

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**Conflict of Interest**

The authors declare no conflict of interest.

**Data Availability Statement**

Research data are not shared.
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conductive nanocomposites, electronic skins, human–machine interfaces, large-area sensor arrays, multifunctional sensing

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