

Flexible Pressure Sensors for Objective Assessment of Motor Disorders

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Monitoring body motion is relevant to motor control disorders as well as assessment of fine motor skills in child development. Furthermore, motion tracking is necessary for rehabilitation monitoring and injury prevention and benefits both sick and healthy individuals. Flexible pressure sensors based on resistors, capacitors, inductors, or transistors are reviewed in the context of healthcare measurements, ranging from physiological signals to body movement characteristics such as grip and gait. To demonstrate the use of flexible pressure sensors for motor assessment, a touch sensing glove that evaluates fine motor skills in autism research is developed. The results show that autistic children perform fewer taps per minute compared to typically developing children, with larger variations in tap durations. In a second example, a force and motion sensing glove is developed to assess spasticity, a neuromuscular disorder that causes muscle stiffness/resistance and jerky movement. Analyses of force versus velocity show movement-dependent muscle resistance in a patient with spasticity. Through these flexible sensor systems, the shift from subjective scores to objective measurement will promote better diagnosis and dramatically improve the accuracy in tracking patient response to therapy.

1. Introduction

Monitoring body motion offers clues to a person's health status and aging process.^[1] Motion tracking studies have provided guidelines to prevent sports injuries^[2] and to enhance occupational well-being.^[3] Continuous monitoring of body movements can provide feedback on injury rehabilitation^[4–6] and enable early detection of disorders that affect motor control, for example, Parkinson's disease and multiple sclerosis.^[7,8] Furthermore, assessment of fine motor skills is one of the screening tests for conditions such as epilepsy^[9] and autism spectrum disorder (ASD).^[10,11] However, the current approach to evaluate motor skills is often subjective; that is, clinicians would observe

patients doing certain movements, and then clinicians rank each patient's level according to qualitative descriptions in benchmark classification scales.^[12,13] The subjective scores can be inconsistent between raters and do not capture fine-level changes in a patient's progress in response to therapy. Therefore, there is a critical need to tackle the issue of imprecise assessment of motor disorders.

With recent advances in flexible sensors and innovations in tactile sensing,^[14–19] we have new low-cost technologies that are prime to facilitate quantitative evaluation of motor control. This progress report presents current developments in wearable sensor systems applicable to motor disorders, so that consistent, objective metrics become available to accurately track whether a treatment effectively relieves symptoms. The wearable sensors are not limited to placement on patients but can also be worn by clinicians or caregivers to assist them in taking meas-

urements during patient interactions.^[20,21] The point-of-care sensor systems will allow frequent monitoring, which is highly desirable to offer a better understanding of the patient's short and long-term response to therapies, to tailor treatment and improve outcome and quality of life for patients.

Other reviews^[14–19] have already extensively covered the flexible materials and devices used in tracking vital signs and electronic skin applications. In light of that, this progress report focuses on the applications in monitoring motor skills, in particular to implement pressure sensor designs for wearable systems with diverse form factors, for instance, gloves,^[20,22,23] epidermal tags tags,^[24,25] and shoe insoles.^[26,27] We will discuss the mechanisms behind different pressure sensor designs

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1 in Section 2, the material choices in Section 3, and healthcare
2 applications based on body motion sensing in Section 4. Spe-
3 cifically, we will show the progress of our own work for two use
4 cases: i) tracking fine motor skills in autism research and ii)
5 quantifying spasticity, a debilitating condition with increased
6 stiffness of the limbs and inability to produce fluid movements
7 as a result of brain injury, stroke, cerebral palsy, and other dis-
8 eases. We will present our perspectives on sensor designs and
9 limitations and our approaches to engineer systems toward
10 practical clinical use.

11 2. Sensor Designs and Mechanisms

12
13
14
15 The transduction of mechanical forces into electrical signals
16 has been realized in flexible devices that measure strain, pres-
17 sure, torque,^[28,29] or acceleration.^[30,31] Here we focus our dis-
18 cussions on pressure sensors; but we note that the designs
19 may be generalizable to other mechanical sensor types because
20 strain and stress (i.e., pressure) are connected by elastic
21 moduli, and the forces measured by pressure sensors can be
22 extracted to infer torque or acceleration. We categorize pres-
23 sure sensing mechanisms into equivalent-circuit components
24 as shown in **Figure 1**, represented by passive components
25 (resistors, capacitors, and inductors) and active amplification
26 devices (transistors). Individual sensor performance metrics,
27 such as the detection limit, sensitivity/resolution, linearity, and
28 response time, are compared across the categories, along with
29 system-level considerations of power consumption and envi-
30 ronmental stability. **Table 1** summarizes the main transduction
31 mechanisms, materials, advantages, and limitations for four
32 different types of pressure sensor.

33 2.1. Resistive Pressure Sensors

34
35
36
37 Resistive pressure sensors consist of an elastic conduc-
38 tive or a semiconductive layer in contact with two electrodes.
39 The electrodes can be in a coplanar interdigitated geometry
40 (Figure 1a)^[32] or they can be vertically aligned as a sandwich
41 structure (Figure 1b).^[33] The elastic layer is patterned to tune
42 the elastic modulus; some microstructure examples include
43 sponges,^[34] pyramidal structures,^[35–37] and nanowires^[38,39]
44 which specifically allow sensors to be anisotropic and selectively
45 respond only to a certain pressure direction.^[40] Resistance R is
46 given by $R = \rho L/A$, where ρ is the resistivity, L is the length,
47 and A is the cross-sectional area of the elastic layer. In the case
48 of constant resistivity ρ , the dimensional changes in L and/
49 or A under pressure determine the change in R . Alternatively,
50 the resistivity ρ of the elastic layer could vary with pressure. For
51 instance, in a piezoresistive semiconductor the change in band
52 structure under pressure would shift ρ , or in a composite mate-
53 rial pressure decreases the interparticle separation distance and
54 thereby reduces ρ . The contact resistance between the elec-
55 trodes and the elastic layer may be affected by pressure, adding
56 to the change in the overall measured resistance.

57 Resistive pressure sensors offer wide dynamic range span-
58 ning orders of magnitudes, relatively simple device structures
59 and fabrication processes, and a detection limit down to



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0.6 Pa.^[41] At very low pressure range < 10 Pa, high sensitivity of
44 9000% change kPa^{-1} could be reached.^[42] Due to their small
45 footprint, resistive sensors can be integrated at high density
46 in pixelated sensor arrays. However, the disadvantages seen
47 in resistive sensors are that they show higher drift over time
48 and are more sensitive to temperature compared to capacitive
49 and inductive devices, when accounting for thermal expansion
50 and thermally induced dielectric constant variation.^[43,44]
51 Polymer resistive sensors are limited to slow millisecond
52 response time and display hysteresis between loading and
53 unloading responses because of the viscoelasticity in polymers.
54 The viscoelastic properties affect other sensor mechanisms as
55 well, to be discussed below.

52 2.2. Capacitive Pressure Sensors

53
54 The structures of capacitive pressure sensors can be in a
55 parallel-plate geometry (Figure 1c)^[45] or a coplanar form with
56 interdigitated electrodes (Figure 1d).^[46] For capacitive sen-
57 sors the elastic layers between the electrodes are made of
58 insulators, in contrast to (semi)conducting composites in
59 resistive sensors. On the other hand, similar to resistive

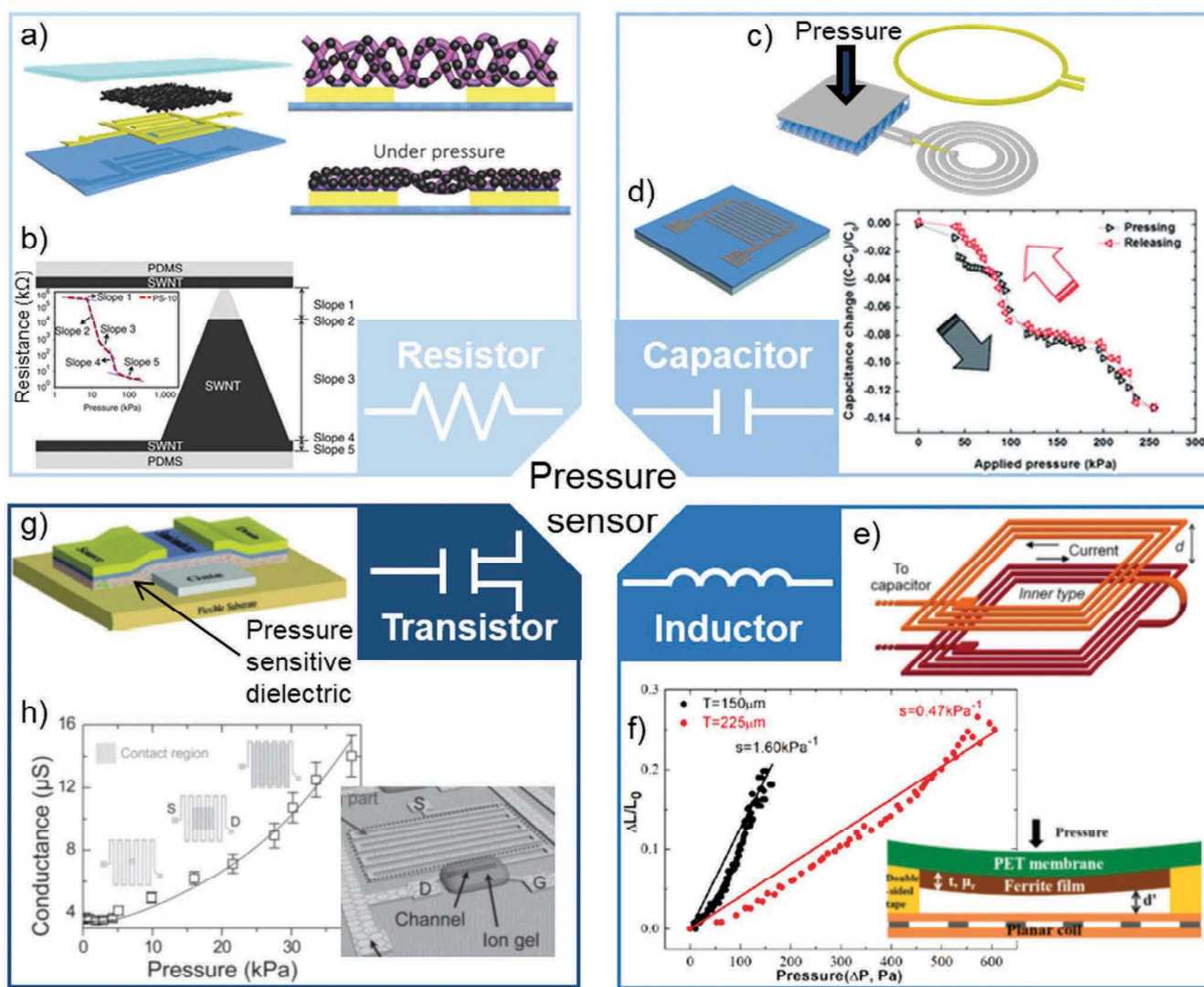


Figure 1. Pressure sensor designs. a) Coplanar interdigitated resistive pressure sensor. Reproduced with permission.^[32] Copyright 2015, Wiley-VCH. b) Vertical sandwich structure of resistive pressure sensor with a pyramidal elastomer composite. Reproduced with permission.^[33] Copyright 2015, Springer Nature Publishing. c) Parallel-plate capacitive pressure sensor with a fixed inductor to allow wireless communication. Reproduced with permission.^[43] Copyright 2018, IOP Publishing. d) Coplanar capacitive sensor scheme. Reproduced with permission.^[46] Copyright 2016, The Royal Society of Chemistry. e) Inductive pressure sensor with spiral coils placed on flexible membranes in a folded geometry. Reproduced with permission.^[68] Copyright 2009, Elsevier. f) Inductive pressure sensor tuned by a magnetic core material. Reproduced with permission.^[69] Copyright 2019, MDPI. g) Transistor-based pressure sensor with a piezoelectric dielectric. Reproduced with permission.^[76] Copyright 2011, American Chemical Society. h) Transistor-based pressure sensor with gel dielectric that spreads over a wider area under pressure. Reproduced with permission.^[78] Copyright 2014, Wiley-VCH.

sensors, the compressibility of elastic dielectrics is tuned by microstructuring^[47–49] and adding air voids,^[45,50,51] to lower elastic moduli in capacitive pressure sensors. Hierarchical structures^[52] further enhance capacitive sensor arrays to detect the direction of the applied pressure.

The capacitance C is given by $C = \epsilon_0 \epsilon_r A/d$, where ϵ_0 is the permittivity of vacuum, ϵ_r is the relative permittivity of the material, A is the area of the electrode plates, and d is the distance between the plates. Changes to the geometric parameters (d and A) are major contributors to the capacitance change under pressure in parallel-plate capacitors. Another contribution to the capacitance is from the relative permittivity ϵ_r of the dielectric composite,^[53] which changes as the volume ratio of air to polymer varies under pressure. The coplanar design

relies on tuning permittivity and is widely used for capacitive touch sensors,^[54,55] which are essentially simplified pressure sensors. When the fringe fields between the electrodes are disturbed by an object, the capacitance is changed to indicate a touch event. The capacitance change due to relative permittivity changes can also be attributed to change in nanoparticles density inside the dielectric layer.^[54] It should be noted, that as the capacitive sensors are elastic, dimensional change of the coplanar capacitor could also occur and attribute to the overall capacitance change.^[46]

Capacitive sensors exhibit the advantages of low sensitivity to temperature and humidity variations and low power consumption compared to resistors. Pressure sensitivity of 42% change kPa⁻¹ for range <1.5 kPa and a detection limit of 1 Pa

Table 1. Pressure sensors mechanisms, materials, advantages, and limitations.

Pressure sensing mechanism	Material of active sensing layer	Advantages	Limitations
Resistive	Conductive or semiconductive	<ul style="list-style-type: none"> Wide dynamic range Relatively simple device structures and fabrication High sensitivity at very low pressure range Best detection limit Can be integrated at high density in pixelated sensor arrays 	<ul style="list-style-type: none"> High drift over time More sensitive to temperature changes Polymer resistive sensors limited by slow response time and hysteresis
Capacitive ^{a)}	Insulating	<ul style="list-style-type: none"> Low sensitivity to temperature and humidity variations Low power consumption 	<ul style="list-style-type: none"> Relatively low linear dynamic range Susceptibility to electromagnetic interference and parasitic coupling to the surroundings If using polymers, may be limited by slow response time and hysteresis
Inductive	Conductive	<ul style="list-style-type: none"> High environmental stability Enables wireless communication designs 	<ul style="list-style-type: none"> Larger device footprint Relatively complex fabrication
Transistor	Insulating, semiconductive, or conductive	<ul style="list-style-type: none"> Built-in current amplification Enable scaling to large-area active-matrix arrays with high spatial resolution 	<ul style="list-style-type: none"> Drift and bias stress instability Need for complex optimization of contact resistance and semiconductor transport properties

^{a)}Sub-types include piezoelectric and triboelectric pressure sensors.

are reported for microstructured capacitive sensor.^[56] The combination of a structured pyramid dielectric layer that is also comprised of porous material resulted in very high sensitivity of 4450% change kPa^{-1} (range < 100 Pa), a detection limit of 0.14 Pa, and was insensitive to a temperature change of up to 100 °C.^[42] Yet capacitive pressure sensors show relatively low linear dynamic range and susceptibility to electromagnetic interference and parasitic coupling to the surroundings. The response time and hysteresis may be limited by viscoelastic polymer dielectrics. Nonetheless, for two sub-types of capacitive transducers—piezoelectric or triboelectric sensors, they offer fast response time and can operate as energy-harvesting devices.

2.2.1. Piezoelectric Pressure Sensors

Piezoelectric capacitors use piezoelectric materials as the dielectric. The piezoelectric materials produce a voltage proportional to the applied pressure, due to breaking of symmetry in the chemical structure by external forces.^[57–59] The equivalent circuit model of a piezoelectric pressure sensor consists of a capacitor in series with a voltage source dependent on pressure.^[60] Common piezoelectric polymers, such as polyvinylidene fluoride and its derivatives, have demonstrated fast microsecond response time, high sensitivity of 14 V kPa^{-1} , and a detection limit of 15 Pa.^[58] On the other hand, these materials are temperature sensitive, and the generated voltage dissipates under static conditions. Thus, piezoelectric sensors are suitable for measuring dynamic stimuli but not for monitoring constant pressure.

2.2.2. Triboelectric Pressure Sensors

Triboelectric pressure sensors are based on contact electrification and electrostatic induction;^[61–63] namely, they use

mechanical rubbing between insulators to induce charges on electrodes. The electrostatic charges generated by periodic contact and separation of two insulator surfaces produce an alternating potential and current. A wide variety of triboelectric device structures are available through simple fabrication, as insulating thin polymer films on electrodes are mounted in configurations that draw on mechanical friction.^[57,64–66] A triboelectric device is represented by a voltage source, which originates from the separation of the charges, in series connection with a capacitor, which originates from the capacitance between the two surfaces.^[67] Self-powered triboelectric pressure sensors have been shown to detect pressure from as high as 450 kPa down to 5 kPa with a sensitivity of 0.5 V kPa^{-1} .^[66] The main limitation is that triboelectric sensors cannot monitor a constant pressure, as they produce electrical signal only under movement from pressure changes.

2.3. Inductive Pressure Sensors

The structures of inductive pressure sensors use planar spiral coils, placed on flexible membranes in a folded geometry (Figure 1e)^[68] or close to a core material (Figure 1f)^[69] that changes the coil inductance under pressure. The inductance expressions for spiral inductors are highly dependent on the selected shapes (square, hexagonal, circular, etc.), and we refer readers to ref. [70] for numerical and analytical solutions. In general, the total inductance L has a fixed self-inductance component and a pressure-variable component that is the mutual inductance, which depends on the gap space between the conductor segments or between the coil and a nearby metal core plate. In the case with a magnetic core material like ferrite, the coil inductance is also modulated by changes in effective permeability, as the magnetic core moves toward the coil under pressure.^[69,71–73]

Inductive pressure sensors show high environmental stability since the materials are inert conductors and substrates,

1 and they enable wireless designs. An example shows good sensi-
2 tivity of 160% change kPa^{-1} below 180 Pa and a detection limit
3 of 14 Pa.^[69] Inductive coupling allows wireless communications
4 between the sensor circuit and the readout system that meas-
5 ures the sensor resonance frequency. The full sensor circuit is
6 an LCR oscillator with resonance frequency $f = 1/(2\pi\sqrt{LC})$. The
7 sensor can be an inductor or a capacitor component, such as
8 combinations^[45,74] of a pressure-sensitive capacitor with a fixed
9 inductor or vice versa. The use of pressure-sensitive inductor is
10 not as common as resistive and capacitive sensors, presumably
11 because of larger device footprint and more complex fabrication
12 incorporating magnetic materials or multiple layers of support
13 structures.

16 2.4. Transistor-Based Pressure Sensors

17
18 Thin-film transistors (TFTs) can serve as sensors converting
19 input pressure to output current. Based on the gradual channel
20 approximation model, the TFT current–voltage characteristics
21 operating in the linear regime is described by $I_{SD} = (W/L)(\mu C)$
22 $[(V_G - V_T)V_{SD} - V_{SD}^2/2]$, where I_{SD} is the source–drain current,
23 W is the channel width, L is the channel length, μ is the effec-
24 tive carrier mobility, C is the dielectric capacitance, V_T is the
25 threshold voltage, and V_G and V_{SD} are the control gate voltage
26 and the source–drain voltage, respectively. To turn a TFT struc-
27 ture into a pressure sensor, the gate dielectric is modified with
28 pressure-sensitive materials, using strategies such as micro-
29 structured insulators^[47,75] or composites with piezoelectric
30 particles (Figure 1g)^[76] for which the dielectric capacitance
31 C changes under pressure. There is also a suspended flexi-
32 ble polyelectrolyte dielectric structure^[77] that modulates the
33 gate capacitance in response to pressure. Another approach
34 modulates the channel ratio W/L , as the gel dielectric spreads
35 over a wider area under pressure (Figure 1h).^[78] In addition, the
36 resistance between the source and drain electrodes has been
37 tuned under pressure by using a conductive rubber to vary the
38 transistor current.^[79]

39 TFTs are essential for pixel addressing and TFT-based
40 pressure sensors are integrated structures that enable scaling
41 to large-area active-matrix arrays with high spatial resolu-
42 tion for obtaining pressure maps.^[80] This type of sensors has
43 reached sensitivity of 102% change kPa^{-1} .^[58] There is built-in
44 current amplification in TFT sensors dependent on the applied
45 V_G and V_{SD} biases, providing a knob to tune the current gain
46 in exchange with power consumption. The caveats with using
47 integrated TFT pressure sensors are that stability issues^[81,82] of
48 TFTs may introduce drift in pressure measurements and sig-
49 nificant engineering efforts are needed to optimize TFT contact
50 resistance^[83] and transport properties.^[84]

53 3. Sensor Materials

54
55 A suite of materials ranging from conductors, semiconductors,
56 to insulators are developed to achieve flexibility and stretcha-
57 bility in wearable health monitors.^[85] Figure 2 summarizes our
58 view of the common approaches to simultaneously optimize
59 mechanical and electronic characteristics.

1 Highly elastic and stretchable conductive materials are
2 needed for the electrodes and interconnects. Examples include
3 composite percolation networks^[86,87] of conductive parti-
4 cles and elastomers (Figure 2a), and inherently stretchable
5 materials (Figure 2b) such as liquid eutectic metals^[88,89] or
6 conducting polymers^[90] demonstrated to maintain a conduc-
7 tivity $\approx 100 \text{ S cm}^{-1}$ under 800% tensile strain. Thin film metals
8 have been made into highly conformal, stretchable conductors
9 by geometry optimization using serpentine^[91,92] or cilia struc-
10 tures (Figure 2c).^[93]

11 Semiconducting materials (Figure 2d–f) are essential in
12 current-modulation switches, ranging from transistor-based
13 sensors to bioinspired synaptic devices.^[94,95] 1D carbon
14 nanotubes^[96] and atomically thin 2D materials^[97,98] have been
15 integrated as semiconductors in flexible transistors. Semicon-
16 ducting polymers with dynamic hydrogen bonds enable robust
17 transistors that are stretchable and repairable by solvent and
18 thermal treatments.^[99]

19 Electrical insulators constitute the dielectrics in capacitors,
20 inductors, and transistors. Many elastic polymers are avail-
21 able and their mechanical moduli are easily tunable by varying
22 crosslinker ratios and hydrogen bonding like in hydrogel^[100]
23 (Figure 2g).^[100] Particular properties such as piezoelectric
24 responses^[101] can be incorporated using a polymer composite
25 with ferroelectric perovskite particles or a ferroelectric polymer
26 such as polyvinylidene fluoride (Figure 2h). The internal struc-
27 tures of dielectrics are often patterned to modify the mechan-
28 ical modulus for the targeted pressure range (Figure 2i).

29 In addition to electrical and mechanical properties, other
30 criteria to consider when choosing materials for wearable sen-
31 sors include the materials processing requirements and safety
32 for human use. As many of the aforementioned polymers and
33 composites are processable as solutions, they are conducive
34 to low-cost printing fabrication^[102–106] to make sensors widely
35 deployable. The materials nontoxicity and biocompatibility is
36 a crucial requirement, since pressure sensors for monitoring
37 motor disorders require contact with the human body. Currently
38 sensors are encapsulated with biocompatible materials^[107] to
39 interface between the sensor and the human body. In order to
40 minimize electronic waste buildup and reduce environmental
41 impacts, biodegradable materials, such as paper-based pressure
42 sensors,^[108,109] could be disintegrated in solution (e.g., phos-
43 phosphate buffer solution) or by incineration. By choosing materials
44 that combine biocompatibility and biodegradability, a class of
45 bioresorbable materials have been used in implanted transient
46 pressure sensors that are safely broken down inside the body,
47 eliminating the need for invasive removal procedure.^[110,111]

48 To achieve durable sensors, self-healing materials that are
49 capable of repairing mechanical fractures autonomously under
50 ambient conditions are desired, particularly for flexible pres-
51 sure sensors that undergo continuous mechanical deformation
52 during their usage.^[112–114] The self-healing mechanisms are
53 categorized as either intrinsic or extrinsic.^[112,115] Intrinsic self-
54 healing materials is based on molecular interactions between
55 crosslinking groups embedded onto the polymer chains. They
56 display fast healing times and multiple healing cycles but
57 require extensive molecular designs and synthesis. Extrinsic
58 self-healing materials are based on monomers and catalysts
59 prepacked in capsules or vessels that are released when the

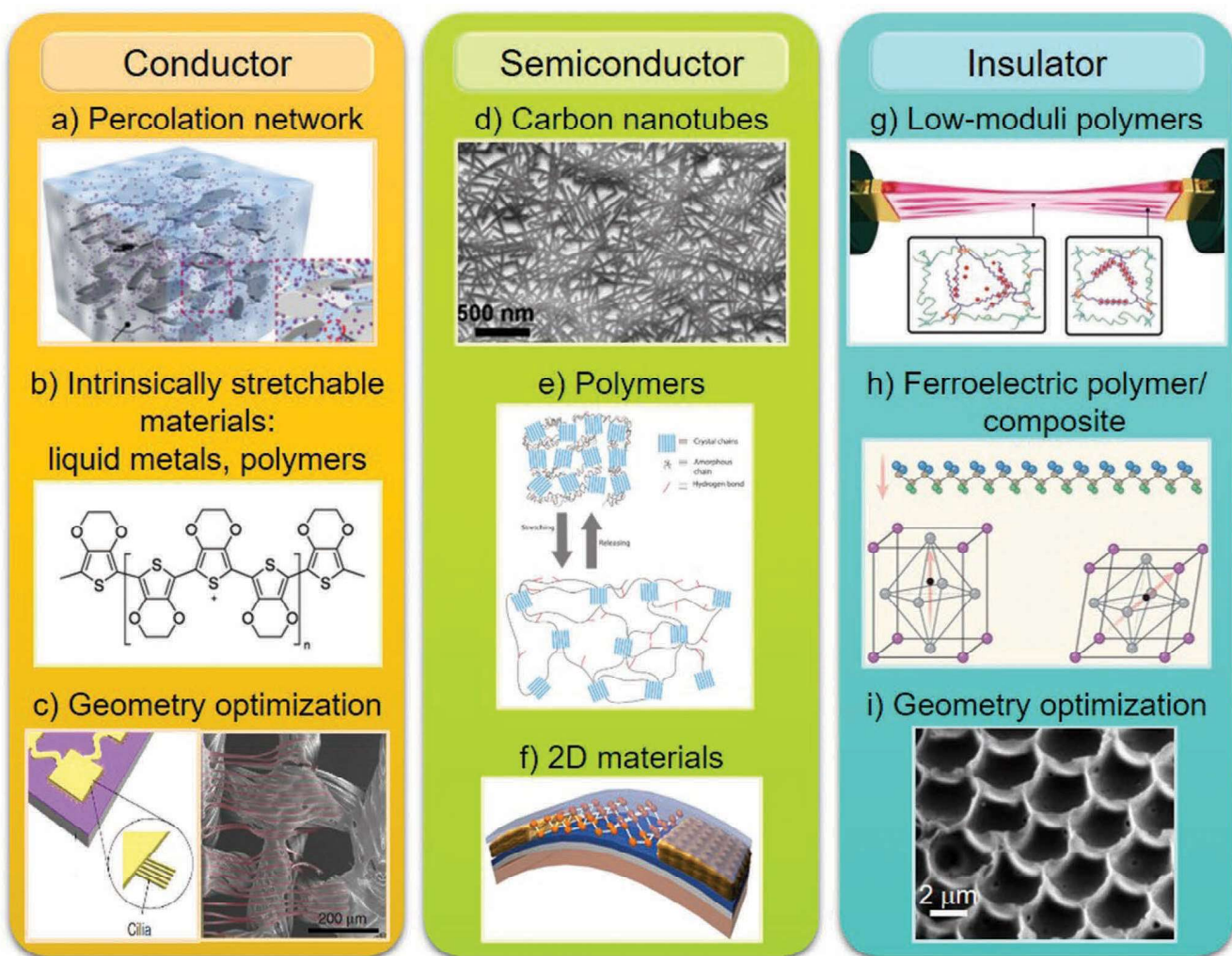


Figure 2. Material choices for flexible pressure sensors. a) Composite percolation network that forms conducting paths with Ag flakes. Reproduced with permission.^[86] Copyright 2017, Springer Nature Publishing. b) Chemical structure of poly(3,4-ethylenedioxythiophene), a conducting polymer. c) Metal patterned with cilia to form stretchable structures. Reproduced with permission.^[93] Copyright 2016, Springer Nature Publishing. d) Single-walled carbon nanotubes. Reproduced with permission.^[166] Copyright 2013, American Chemical Society. e) Schematics of semiconducting conjugated polymers with dynamic hydrogen bonds to improve stretchability. Reproduced with permission.^[99] Copyright 2016, Springer Nature Publishing. f) 2D semiconductor. Reproduced with permission.^[167] Copyright 2015, American Chemical Society. g) Low-moduli, highly stretchable hydrogel. Reproduced with permission.^[100] Copyright 2017, American Association for the Advancement of Science. h) Chemical structure of poly(vinylidene fluoride-co-trifluoroethylene) (top) and lead zirconate titanate (bottom). Reproduced with permission.^[101] Copyright 2018, Springer Nature Publishing. i) Polyurethane patterned into nanoneedle structures for a dielectric layer. Reproduced with permission.^[47] Copyright 2012, American Institute of Physics.

polymer matrix is damaged. They enable large-volume self-healing, yet display slower healing times, and can be more complex to process than intrinsic self-healing materials.

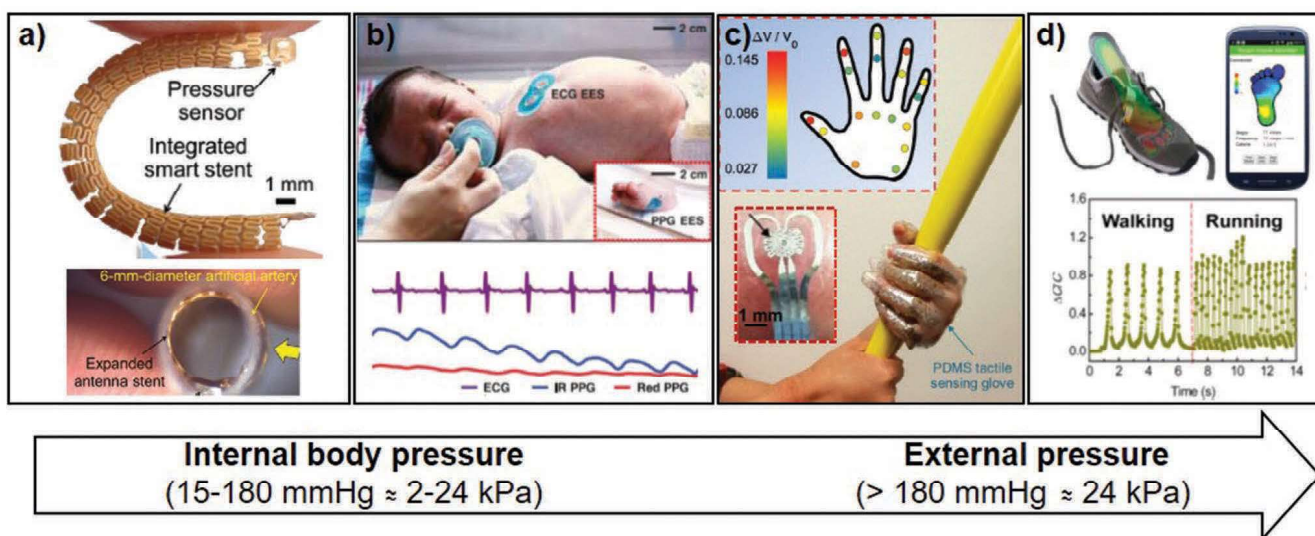
4. Applications of Pressure Sensors in Monitoring Body Motion

Here we categorize body pressure measurements into two ranges in **Figure 3**: i) internal body pressure corresponding to physiological signals, e.g. heart pulse, intraocular pressure, around 15–180 mmHg (≈ 2 –24 kPa); ii) external pressure exerted in motor control studies, exceeding 180 mmHg (>24 kPa) in grip or gait signals. In Sections 4.1–4.3, we present some of

the research achievements in flexible wearable electronics. While these prior work discussions are not comprehensive, they provide the context leading to our own work presented in Sections 4.4 and 4.5.

4.1. Monitoring Internal Body Pressure

Flexible pressure sensors have been widely deployed for continuous monitoring of vital signs (e.g., heart rate and respiratory rate) and other physiological signals (e.g., blood pressure and intraocular pressure). The devices have been placed as implants directly inside the body (Figure 3a)^[116] or noninvasively outside the body (Figure 3b).^[25] Cardiovascular diseases



Internal body pressure
(15-180 mmHg \approx 2-24 kPa)

External pressure
($>$ 180 mmHg \approx 24 kPa)

Figure 3. Devices for monitoring body motions. a) Stent with capacitive pressure sensors to monitor blood flow. Reproduced with permission.^[116] Copyright 2018, Wiley-VCH. b) Epidermal ECG and PPG sensors. Reproduced with permission.^[25] Copyright 2019, American Association for the Advancement of Science. c) Tactile sensing glove measuring grip pressure. Reproduced with permission.^[123] Copyright 2017, Wiley-VCH. d) Insole with capacitive pressure sensors. Reproduced with permission.^[50] Copyright 2016, American Chemical Society.

are the main cause of mortality worldwide, and especially in people with stents to treat artery obstruction, close monitoring of the blood flow is important to prevent heart attacks and strokes. The stents used to widen arteries will become narrow over time due to endothelial tissue growth or deposits. Thus, a study on stent technologies has incorporated micro-scale capacitive pressure sensors with wireless interface for in vivo continuous monitoring to detect flow obstruction (Figure 3a).^[116] In another example of continuous monitoring based on implants, the intraocular pressure, i.e., the fluid pressure inside the eye, was monitored wirelessly by an LC resonator sensor.^[117] The intraocular pressure is a primary indicator of glaucoma, the second leading cause of blindness. Sensors that offer wireless, continuous measurement are crucial because peaks in intraocular pressure might occur during sleep time, and not necessarily during daytime checkup.^[118] A flexible bioresorbable piezoresistive sensor implant was used to monitor physiological forces in vivo in mice.^[110] The system monitored the diaphragmatic contractions to detect breathing patterns, while avoiding invasive removal procedure of the implant.

Changes in internal body pressure are transmitted to the body surface, and noninvasive monitors are capable of inferring many different physiological signals, offering unintrusive health tracking not just for the sick but applicable to wide populations. For example, to avoid surgeries required with implants, flexible sensor technologies are being developed for contact lenses to measure intraocular pressure and detect eye problems.^[119,120] Surface measurements of arterial pulse waves that are indicative of the heart health were demonstrated with different pressure sensors, such as a triboelectric sensor^[65] or a liquid-capsule platform with a resistive sensor.^[121] In another use case, muscle vibrations from the vocal cord were recorded using triboelectric pressure sensor or hydrogel resistive sensor.^[65,122] These signals have applications in Parkinson's

disease monitoring, as studies found deterioration in phonation due to Parkinson's disease.^[7]

4.2. Monitoring Pressure Exerted by Body Movement

Large-area flexible pressure sensor arrays have been integrated into portable systems with high spatial resolution for research in musculoskeletal biomechanics. For instance, grip and gait are common measurements used in therapy and rehabilitation as well as for sports training. Gloves were embedded with resistive sensors for tactile mapping of human grasps.^[23] Figure 3c shows hand-grip mapping with microfluidic diaphragm pressure sensors.^[123] These resistive diaphragm sensors exploit a Wheatstone bridge circuit to measure tangential and radial strain. Under an applied pressure, a decrease in the tangential cross-sectional area due to tension leads to an increase in the tangential bridge resistance; on the other hand, compression around the periphery results in an increase in the radial cross-sectional area, leading to a decrease in the radial bridge resistance. When the glove is worn by a patient, the grip data indicate upper body and overall strength of the patient.^[124] Alternatively, a physician wearing the instrumented glove^[20] would be equipped to track forces from different maneuvers while interacting with the patient.

Quantifying plantar pressure using unintrusive, flexible insoles (Figure 3d)^[50] provides information regarding gait and posture, which are used for diagnostics in many areas including orthopedic problems, neurological disorders (e.g., multiple sclerosis and Parkinson's disease), sport biomechanics, and injury prevention and rehabilitation.^[4,27,125,126] A resistive pressure sensor for monitoring gait uses hydrogels with self-healing capability to extend the sensor life time.^[114] Within gait monitoring and analysis research, there is an urgent need to serve diabetic patients by identifying and preventing high pressures

1 on the soles of their feet. The high pressure points would result
2 in ulcerations, and their relief is important to not exacerbate
3 sore spots. Hence, real time monitoring during daily activities
4 is highly desirable to manage such chronic problems.^[127]

4.3. Multimodal Monitoring Systems

9 Multimodal systems that combine several sensors types pro-
10 vide comprehensive analyses for vital signs,^[25] chemical
11 biomarkers,^[128,129] and musculoskeletal conditions.^[5] Com-
12 paring pressure signals from arterial pulses and electrocar-
13 diograms^[130] (ECGs) enable extraction of pulse transit times
14 to determine the systolic blood pressure.^[121] Alternatively,
15 the blood flow motion during heart beats can be monitored
16 indirectly in photoplethysmograms^[131–133] (PPGs) which use
17 flexible photodetector technologies.^[134–137] Notably, a complete
18 epidermal system that allows wireless, battery-free, continuous
19 sensing of PPGs and ECGs on neonates has been achieved with
20 performance comparable to clinical-grade monitoring systems
21 (Figure 3b).^[25] Flexible bioresorbable implants that eliminate
22 the need for a second surgery to remove them from the body
23 were demonstrated in vivo on rats to monitor the electrocor-
24 ticography (ECoG) and intracortical pressure signals.^[111] The
25 implanted sensors recorded dynamic changes in brain signals
26 for different stages of epilepsy and simultaneously tracked
27 swelling of the cortex during and after the operation.

28 The combination of pressure sensors with inertial mea-
29 surement units^[138,139] records parameters of force, power, and
30 range of muscle movements, in order to gather complementary
31 descriptions of motor characteristics in biomechanic studies.
32 Moreover, the use of multimodal sensors has been extended
33 to behavioral studies in child development research. Simulta-
34 neous monitoring of body movements and physiological status
35 has been carried out to track stressor response^[140] and then
36 predict imminent aggression in children with autism spectrum
37 disorder.^[141] Wearable sensor systems that facilitate similar
38 automated tracking would benefit individualized, real-time
39 interventions in the future.

4.4. Objective Assessment of Fine Motor Skills in Children with Autism Spectrum Disorder

4.4.1. Motivation

47 Our first application example focuses on evaluating fine-motor
48 skills to identify sensory-motor dysfunction, a symptom of neu-
49 rological disorders (e.g., Alzheimer's, Parkinson's, epilepsy,
50 and autism). The routine screening method used in clinical
51 settings is the finger tapping test (FTT).^[142] This simple test
52 scores the number of taps per minute by a patient tapping
53 one's finger on the thumb or on a counter key. The average
54 number is a measure of motor speed. The performance differ-
55 ence of an individual over time can track disease progress or
56 regression.^[143] Currently FTT is done by manual counting or
57 is recorded by video camera to be later processed by computer
58 vision algorithms (Figure 4a). There are also computer inter-
59 face devices for FTT,^[142,143] but the existing button interface is

isolated and not designed for combination tests that examine
coordinated motor and cognitive functions. Thus, FTT can be
improved by a hardware redesign using flexible sensors.

Specifically, we designed a low-cost (USD\$ < 20), intuitive
touch sensing glove that can be modularly integrated with tests

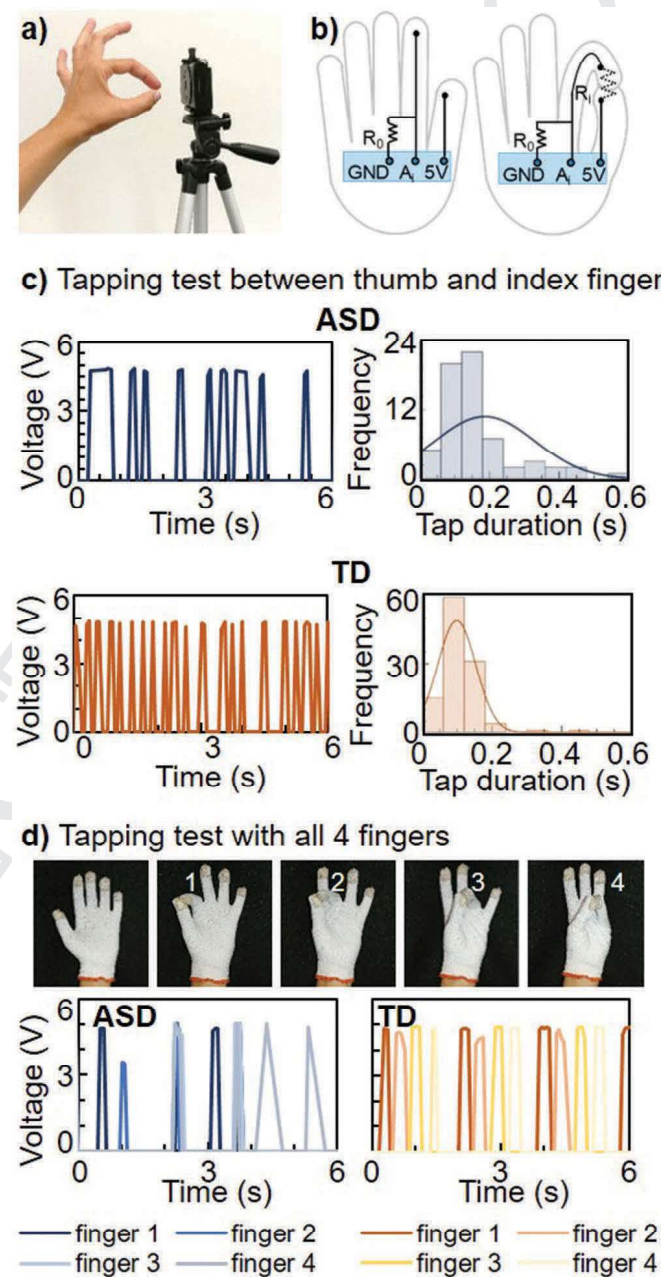


Figure 4. Finger tapping test for evaluating fine motor skills. a) Test recorded by video camera. b) Circuit schematics of the resistive touch sensor glove, with fingers not in contact (left) or in contact (right). c) Representative data of a tapping test between the thumb and the index finger. When fingers are in contact, the signal is high around 5 V; if not in contact, the signal is low at 0 V (left). Corresponding histograms of the contact duration (right). d) Photograph of the contact sequence in a tapping test using all fingers. Representative data showing contact frequency and duration for each finger. Tests here done by children with autism spectrum disorder (ASD) or in typical development (TD).

1 that evaluate motor skills and cognition.^[144] This glove is used
2 for research in ASD, a common neurodevelopmental disorder
3 that occurs in 1 out of every 68 children and affects ≈1% of the
4 general population. ASD is characterized by deficits in social
5 communications and interactions, as well more commonly over-
6 looked but also important, motor skill deficits, and repetitive
7 behaviors.^[11,145–147] Early, accurate identification of motor skill
8 deficits is critical for providing timely intervention treatments
9 that help both the physical and the psychological development
10 of children with ASD. Often their motor skill assessment is
11 done only by observation and questionnaires, which are limited
12 in scope and subjective. Incorporating neuropsychological
13 tests like FTT has helped to objectively characterize movement
14 issues in ASD children and FTT would provide outcome meas-
15 ures to evaluate motor-based interventions.

18 4.4.2. Measurement Methods and Results

20 Our resistive touch sensing glove for FTT was made by modi-
21 fying a fabric glove, on which we patterned conductive elec-
22 trodes using silver paste^[130] or conductive fabric (Sparkfun).
23 The electrodes on finger tips were connected by sewing conduc-
24 tive threads to a microcontroller board (Arduino Nano) sewn
25 at the base of the glove and then packaged up for better user
26 experience.^[148] If needed, the resistive touch sensor can be
27 easily modified to integrate pressure sensing function, but for
28 this demonstration the device was made as simple as possible.
29 The electrodes and interconnects could be stretched to about
30 150% of their original value; this stretchability was sufficient to
31 allow putting the glove on and off without damaging it, and it
32 allowed different test subjects with different hand sizes to wear
33 the same glove. The touch readout used a voltage divider circuit
34 between the thumb and each of the four fingers (index, middle,
35 ring, and pinky), with $R_0 = 510 \Omega$ and $R_1 = 10 \Omega$ (Figure 4b).
36 When fingers were in contact, the signal was high around 5 V;
37 if not in contact, the signal was low at 0 V. Our user interface
38 code provided real-time data analysis.

39 Two types of FTTs were carried out to compare the tapping
40 patterns of ASD and typically developing (TD) children: index-
41 finger tapping test and all-finger tapping test. Three children
42 aged 6–13 in each category was recruited for the first and two
43 children in each category for the latter test. All participants
44 signed an informed consent sheet verified by the University
45 of California San Diego (UCSD) Human Research Protec-
46 tions Program under Institutional Review Board #171587.
47 In the index-finger tapping test, subjects were asked to tap
48 their index finger and the thumb together as many times as
49 possible in a given time interval of 30 s. ASD children per-
50 formed fewer tap counts than TD children, as seen from
51 the frequency axes of the histograms in Figure 4c. In addi-
52 tion to counts per second, the contact duration was longer
53 and showed more variance for ASD children in comparison
54 to TD children. There was no statistically significant different
55 between measurements with and without the glove (analyzed
56 from video recording), attesting that wearing the glove did not
57 impair finger tapping motion.

58 In the all-finger tapping test, the subjects were asked to
59 touch their thumb with one finger at a time in sequence as

shown in Figure 4d, performing as fast as possible in a given
time frame of 30 s. The results of the all-finger tapping test
showed irregular tapping pattern by ASD children, as well as
events when multiple fingers touching the thumb together at
the same time. In contrast, TD children were able to regularly
repeat the intended tapping pattern.

The motor deficits in ASD population are being addressed
by game-based trainings,^[149] and FTTs were part of the toolset
to monitor individual progress due to medication or training,
and not only for comparison between groups of subjects. Our
simple, low-cost touch sensing glove have been incorporated
into multitask experiments, for example simultaneously quanti-
fying FTT, balance, and working memory in ASD.^[144] The touch
sensor is modular and can be adapted to different research tests
such as measuring reaction time and touch sequence, as the
subject performs tasks while wearing the unintrusive sensor
glove. As the fabrication cost is very cheap, it could be widely
distributed for clinical studies of motor disorders and moni-
toring fine motor skills.

4.5. Objective Assessment of Spasticity

4.5.1. Motivation

Our second application example focuses on the assessment of
spasticity. Spasticity is a neuromuscular disorder due to brain
or nerve damage in patients suffering from cerebral palsy,
stroke, etc.^[150–152] Spasticity results in muscle stiffness, painful
contractures, and jerky limb movements, affecting a person's
motor control in balance, gait, eating, hygiene situation, and
more. It is estimated that over 12 million people worldwide
suffer from some level of spasticity.^[151] Timely evaluation and
treatment are important to relieve pain and muscle deformi-
ties (Figure 5a, top), but there is a key problem of imprecise
assessment.

An objective, consistent metric is critical for inspecting
whether a therapy effectively relieves symptoms. Yet current
clinical practice relies on perception. Clinicians would assess
patient severity by performing standardized maneuvers,^[13,150]
in which the clinician extend and flex the patient's affected
muscles to gauge the muscular resistance to movement. Based
on subjective perception, the clinician scores the muscle con-
dition against benchmark scales, the most common being the
Modified Ashworth Scale with only six rating levels (Figure 5a,
bottom).^[153] This subjective score is not sensitive and known to
be inconsistent between raters and even for the same rater.^[12,13]
Besides inaccuracy, such evaluation requires appointments and
are done weeks apart, and the treatment decisions are often
lagging, especially affecting development in rapidly growing
pediatric patients.

There is ongoing research aiming to better quantify spas-
ticity. Surface electromyography (EMG) was demonstrated
to capture the involuntary muscle activations in patients'
limbs.^[154–156] While the neural signals objectively charac-
terize reflex thresholds, there is no clear relationship that
correlates the bursts of neural activities to severity levels of
spasticity.^[154,157] EMG suffers from low signal reproducibility
due to variations in electrode locations, patients' sweating, and

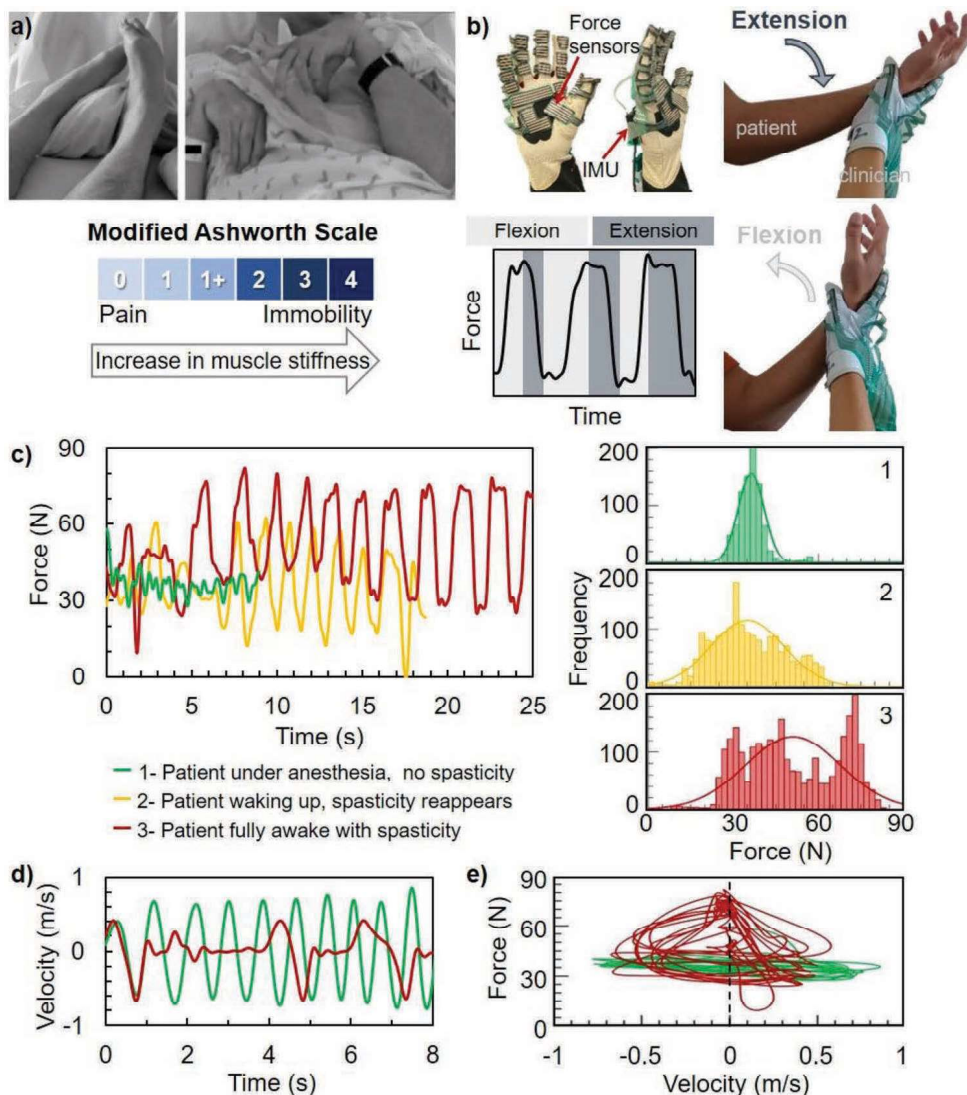


Figure 5. Objective assessment of spasticity. a) Photographs of affected limbs due to spasticity (Reproduced with permission.^[168] Copyright 2013, Taylor and Francis.). Benchmark scale commonly used in spasticity evaluation. b) Photographs of our instrumented glove with resistive force sensors and inertial motion unit (IMU). A clinician wears the glove to move a patient's limb into an extension or flexion position. The inset shows a typical force signal measured for multiple extension and flexion cycles. c) Force versus time (left), for three different levels of spasticity in a patient transitioning from being under anesthesia to the fully awake state. Corresponding histograms of force in each state (right). Legend applies to parts (c)–(e). d) Velocity versus time, measured simultaneously with the force data in part (c). e) Force versus velocity.

environmental noise interference. In a different approach, biomechanical devices were developed to measure the muscle stiffness and the peak torque of spastic limbs.^[158,159] A six-axis robotic arm was used to manipulate a person's limb and precisely record the position and force to extract resistance of the muscle.^[160] However, large motorized structures^[160,161] are expensive, complex to operate, and pose safety and clinical adoption barriers. Recently, a compact device was demonstrated to concurrently record muscle torque and EMG.^[162,163] Most modules place instrumentation on the patient. In contrast, we design our equipment to be worn by clinicians or caregivers, because it is more reliable to cater the device size to a rater than to various patients, and there is no hardware on patients that may impede movement.

4.5.2. Measurement Method and Results

To address the need for objective assessment of spasticity, we integrate an instrumented glove,^[20] equipped with resistive pressure sensors from Tekscan^[164] and an inertial motion unit (IMU) from MotionNode. The sensor glove is intended to be worn by the rater, to record the applied force and motion as the rater moves a patient's limb in extension and flexion cycles (Figure 5b). Signals from the pressure sensors are multiplied by active areas and summed together to extract the applied force. The IMU signals, including acceleration, angular velocity, and magnetic field orientation, provide information to extract movement trajectories. Simultaneous force and motion measurements are essential to fully capture

1 the dynamic stiffness/resistance of spastic muscles, since
2 spasticity is perceived to depend on movement velocity and
3 shows catch-and-release symptoms.^[165] Below we show data
4 measured by our sensor glove on a patient's spastic knee
5 joint muscles. All measurements were undertaken with
6 patient consent and approved by UCSD Institutional Review
7 Board #180115.

8 A patient with spasticity underwent a surgical procedure
9 that required general anesthesia. After surgery, the spastic
10 muscle resistance was measured three times: 1) when the
11 patient was under anesthesia, 2) when the anesthesia wore
12 off and patient was waking up, and 3) when the patient was
13 fully awake. As a neuromuscular disorder, spasticity is corre-
14 lated to one's neural state; it is known that anesthesia would
15 reduce muscle resistance because it decreases neural misfir-
16 ings. Figure 5c shows the cyclical force exerted to maneuver
17 the patient's joint muscles for several extension and flexion
18 cycles. With the patient under anesthesia, the average force
19 was the lowest at $F_{\text{average}} = 37 \pm 5$ N, and the force histo-
20 gram distribution was the narrowest among the three meas-
21 urement states. As anesthesia wore off, force distribution
22 became wider with $F_{\text{average}} = 35 \pm 13$ N. The peak-to-peak
23 force increased, indicating higher muscle resistance. When
24 the patient was fully awake, the average force increased sig-
25 nificantly at $F_{\text{average}} = 51 \pm 17$ N, accompanied by a change
26 in force distribution, which will be clarified by the velocity
27 discussion below.

28 Figure 5d shows large difference in movement velocities
29 between states 1 and 3. The linear velocity was converted from
30 the angular velocity acquired from IMU gyroscope along the
31 direction of movement, multiplied with the distance between
32 the rater's grip and the patient's knee joint. Under anesthesia,
33 the patient's muscle was easy to flex and extend, and the velocity
34 was highly cyclical as shown by the green data. When spasticity
35 became more prominent after anesthesia wore off, the recorded
36 velocity was no longer cyclical, because the patient's limb could
37 not be moved fluidly as shown by the red data.

38 Analysis of the combined signals, i.e., force as func-
39 tion of velocity in Figure 5e, attests that spasticity is motion
40 dependent. The maneuver force was independent of movement
41 velocity for state 1 (green data). In state 3, the force required
42 to move spastic muscle increased, particularly as velocity was
43 near zero when the rater switched between flexion and exten-
44 sion (red data). It should be noted that for the patient exam-
45 ined here, the force-velocity relationship was not symmetric
46 between flexion (i.e., positive velocity) and extension (i.e., nega-
47 tive velocity). Indeed, for some patients their muscle resistance
48 is more noticeable in one segment of the maneuver cycle and
49 less in the other. As the next step, we will explore additional
50 parameters to describe spasticity, such as the range of motion
51 and the expended power.

52 Overall, the significance of the above measurements is that
53 our multimodal sensor glove can clearly distinguish changes
54 in muscle spasticity and is promising for quantifying spas-
55 ticity. It will enable a point-of-care device that allows frequent
56 evaluation by caregivers. The shift from subjective scores to
57 objective measurements will promote better diagnosis and
58 dramatically improve the accuracy in tracking patient response
59 to therapy.

5. Summary and Future Outlook

Flexible pressure sensors facilitate objective assessment of
motor disorders. In particular, we have demonstrated their use
in two cases. In the first example we used resistive touch sen-
sors to quantify finger tapping patterns of autistic children. In
an index tapping test ASD children performed less counts per
minute, with larger tap duration and larger variation in the tap
duration, compared to TD children. In all-fingers tapping test
the ASD children had irregular patterns compared to TD chil-
dren and tended to skip fingers more often. This touch-sensing
glove could be used in combination with other tests and for
other conditions that require neuropsychological assessment of
fine motor skills via finger tapping tests, such as Parkinson's
disease, stroke, and epilepsy. Its computer interface could be
used in the future for gamification purposes, to use it for fine
motor skills training.

In the second case we demonstrated a multimodal sensor
glove, to be worn by clinicians for measuring spasticity. The
average applied force and its distribution increased, indicating
higher muscle resistance, as anesthesia wore off a patient
with spasticity. In addition, the limb movement was tracked
and shown to be less fluid with spasticity. The analysis of
force versus velocity clearly showed a difference in the level
of spasticity between flexion and extension maneuvers. This
novel wearable system will enable accurate diagnosis and effec-
tive evaluation of intervention outcomes, toward improving
care and quality of life for patients with spasticity. Moreover,
the instrumented glove could be used in a wide range of
clinical procedures that are currently based on perception of
motion and force, and in other studies on musculoskeletal
rehabilitation.

Objective assessment of motor disorders is an interdisci-
plinary effort. The basic pressure sensing principles open the
door to many possible applications that target various health-
care problems and provide physiological information. The
next steps for objective motion characterization are develop-
ments in signal processing to extract key information out of
noisy real-life data, in addition to improvements in sensor
design and data management. We anticipate that in the future,
additional sensor combinations will enable more accurate
characterizations of health status, and consequently, treat-
ments could be better tailored to improve the quality of life in
patients.

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Conflict of Interest

The authors declare no conflict of interest.

Keywords

autism, finger tapping test, flexible electronics, pressure sensors, spasticity

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