

Review

Smart Materials for Artificial Tendons in Prosthetic Hands: A Review of Materials, Functions, and Applications

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Abstract

The introduction of smart materials is driving a paradigm shift in artificial tendon design for prosthetic hands—transforming them from passive force-transmission elements into active units capable of actuation, sensing, stiffness modulation, and self-repair. This addresses long-standing limitations of conventional tendons, such as creep, nonlinearity, and absence of intrinsic perception. Here we review recent progress in smart-material-based artificial tendons for tendon-driven prosthetic hands, focusing on shape memory alloys (SMAs), smart fibers, electroactive polymers (EAPs), magnetorheological/electrorheological (MR/ER) materials, self-healing polymers, and multi-material composites. Among these, SMAs represent the most mature actuation choice, while piezoresistive and piezoelectric fibers enable static tension sensing and dynamic slip detection, respectively; EAPs offer compliant actuation but remain limited by engineering readiness; MR/ER materials enable tunable stiffness and self-healing polymers offer pathways to mitigate fatigue damage. Multi-material composites are increasingly enabling functional integration toward system-level synergy. Nevertheless, key challenges remain in actuation-sensing compatibility, fatigue life under service conditions, miniaturized packaging, and system integration. Critically, no single material simultaneously fulfills actuation, sensing, and stiffness modulation, making multi-material synergy emerge as the most viable pathway toward next-generation smart tendons. Future efforts should prioritize actuation-sensing integration, multi-material composites with adaptive stiffness, and high-durability designs—aiming for synergistic coupling of material functionality with prosthetic actuation, sensing, and control architectures.

Keywords

Smart tendons, Tendon-driven prosthetic hands, Smart materials, Multi-material composites

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1. Introduction

1.1 Background and Motivation

Prosthetic hands are core devices for fine manipulation, human-robot collaboration, and rehabilitative compensation. In scenarios such as humanoid robot operations, home services, and limb function reconstruction, there is a growing demand for lightweight end-effectors, human-like grasping dexterity, and safe, compliant human-robot interaction [1]. Among the various actuation architectures for multi-fingered dexterous hands, tendon-driven actuation-inspired by the biological tendon-muscle system places actuators proximally in the forearm or the robot trunk, substantially reducing distal inertia, freeing up internal space within the palm, and enabling underactuated or fully actuated designs with passive compliance. This approach has been widely adopted across research prototypes, commercial prostheses, and humanoid robot hands, from the Shadow Dexterous Hand and the commercially available i-Limb myoelectric prosthesis to the DexHand 021 [2] and the ORCA anthropomorphic hand [3]. As the core medium of force transmission, the material properties of artificial tendons exert critical influences on transmission accuracy, grasp stability and operational reliability [4].

Conventional prosthetic hands typically rely on passive tendon materials such as nylon, Ultra-High-Molecular-Weight Polyethylene (UHMWPE), aramid fibers, and stainless steel cables, which mainly function as force-transmission elements. Although these materials satisfy basic grasping requirements, they exhibit significant limitations in high-precision, highly adaptive, and safe human-robot interaction scenarios: they lack intrinsic actuation capability and rely heavily on external motors; they lack built-in sensing capabilities requiring additional sensors for tension detection; their fixed stiffness offers limited adaptability to varying grasping tasks [5]; and they are susceptible to creep relaxation [6] and long-term fatigue damage [7], severely compromising closed-loop control accuracy and adaptive manipulation capability. Accordingly, passively transmitted tendons alone can hardly meet the performance requirements of next-generation intelligent, adaptive and safe prosthetic hands.

Smart materials including Shape Memory Alloys (SMAs), piezoresistive/conductive fibers, electroactive polymers (EAPs), and magnetorheological (MR)/electrorheological (ER) materials, offer new pathways to overcome the above bottlenecks. Through multi-field coupling mechanisms (thermo-mechanical, electro-mechanical, and magneto-mechanical), conventional artificial tendons are gradually evolving from predominantly passive force-transmission elements into intelligent functional units capable of active actuation, proprioceptive sensing, stiffness self-adaptation, and self-healing [8].

1.2 Functional Evolution of Artificial Tendons

The functional role of artificial tendons in tendon-driven prosthetic hands has progressively evolved from passive force transmission toward multifunctional and adaptive operation. This evolution can be broadly characterized by four stages: passive force transmission, function addition, function integration, and multifunctional system integration.

In the first stage, artificial tendons primarily transmit force and displacement between the actuator and the finger mechanism. Conventional materials, including nylon, UHMWPE, aramid fibers, and metallic cables, provide high tensile strength and relatively simple mechanical integration. However, long-term cyclic operation can be affected by frictional losses, creep, termination failure, and fatigue wear [9,10]. These limitations have motivated the development of tendon materials capable of providing functions beyond mechanical transmission.

The second stage introduces additional functions, particularly intrinsic sensing and active actuation. Conductive and piezoresistive fibers can enable tendon-level monitoring of strain or tension through changes in electrical resistance, allowing force transmission and state sensing to be integrated within the same structural element [11]. In parallel, SMAs can generate contractile force through thermally induced phase transformation, enabling the tendon itself to participate in force generation rather than acting solely as a passive transmission element [12]. These developments mark the transition from conventional force transmission toward functionally enhanced artificial tendons.

The third stage concerns functional integration, in which multiple smart-material functions are combined within a tendon or tendon-based transmission architecture. SMAs and EAPs enable the tendon to actively generate contractile force or deformation under external stimuli without the need for additional motors or gear trains [13,14]. Piezoresistive, conductive, and piezoelectric fibers enable the tendon to sense its own strain, tension, and dynamic contact forces, providing real-time feedback for closed-loop control [11,15]. MR and ER materials render the tendon's equivalent stiffness a tunable variable, enabling "rigid-soft switching" when grasping different objects [16].

The fourth stage represents the emerging concept of multifunctional smart artificial tendons, in which actuation, force transmission, sensing, adaptive mechanical behavior, and durability-related functions are integrated within a single tendon architecture. Self-healing polymers and multifunctional material composites have been proposed as potential approaches to improving resistance to damage accumulation and extending service life [17]. However, unlike SMA-based actuation and fiber-based sensing, many of these functions remain at the material-level or proof-of-concept stage and have not yet been demonstrated as fully integrated tendon structures in prosthetic hands.

This functional evolution provides a framework for distinguishing the maturity of different smart-material approaches. Accordingly, this review considers not only the intrinsic properties of smart materials but also their specific tendon-level

functions, evidence of application in prosthetic hands, quantitative performance, and system-level engineering requirements.

1.3 Definition and Scope

In this review, an artificial tendon is defined as a tensile force-transmission element that connects a proximal actuator or driving source to a distal joint or grasping mechanism in a tendon-driven prosthetic hand. Its primary role is to transmit force and displacement to generate or control finger motion. This function distinguishes artificial tendons from conventional cables or wires, which primarily provide passive tensile transmission, and from compliant transmission elements, which additionally accommodate deformation or misalignment. Artificial muscles, in contrast, primarily generate force or deformation rather than serving as the distal force-transmission pathway.

A smart artificial tendon is defined as an artificial tendon that combines tendon-level force transmission with additional functions enabled by smart materials, such as active force generation, intrinsic sensing, variable stiffness, adaptive mechanical behavior, or damage adaptation. The classification adopted in this review is therefore function-based rather than material-based. A smart material is considered within the scope of this review when its responsive properties directly contribute to tendon-level transmission or to an integrated tendon function in a prosthetic hand.

Accordingly, this review focuses on smart materials used or proposed for artificial tendons in tendon-driven anthropomorphic hands and wearable prostheses. Material-level demonstrations are distinguished from tendon-level prototypes, robotic-hand implementations, and integrated prosthetic applications to clarify the technological maturity and practical relevance of different material systems.

1.4 Review Scope and Methodology

Existing reviews have examined smart materials, artificial muscles, tendon-driven prostheses, and prosthetic-hand actuation from material, actuator, mechanical design, or control perspectives [1,18-22]. However, few have taken artificial tendon materials as the central organizing principle to connect material properties with tendon-level functions, prosthetic-hand applications, and engineering requirements. Therefore, this review adopts an artificial-tendon-centered perspective and establishes a material-function-tendon-prosthesis framework for analyzing smart artificial tendons. The review considers material mechanisms, functional capabilities, application levels, quantitative performance, and system-level engineering trade-offs, including force transmission, actuation, sensing, stiffness modulation, durability, integration, and power requirements.

A structured literature search was conducted using Web of Science, ScienceDirect, IEEE Xplore, Scopus, and PubMed, with Google Scholar and reference-list screening used as supplementary sources. The primary search covered publications from 2016 to 2026, with particular emphasis on studies published in the last three years. A small number of earlier publications were also included when they represent foundational work or provide essential background for the development of the field. Keywords included artificial tendons, tendon-driven prosthetic hands, smart materials, SMAs, smart fibers, EAPs, MR/ER materials, self-healing materials, sensing, variable stiffness, and multifunctional tendon systems. Studies were selected when they reported smart materials directly used in tendon-like transmission elements or demonstrated functions with clear potential for tendon-level integration. Reported actuation strain, stress/force output, response speed, operating conditions, fatigue durability, and energy or power requirements were extracted where available, while experimental, theoretical, and manufacturer-reported values were treated separately.

2. Smart Tendon Materials and Their Functional Mechanisms

Based on the transmission characteristics of tendon-driven prosthetic hands and recent advances in smart artificial tendon research, this review identifies five core functional requirements and their corresponding material categories, as summarized in Table 1.

Table 1. Functional requirements and corresponding smart materials for prosthetic hand tendons.

Functional requirements	Key material characteristics	Primary material categories
Active actuation	Large deformation, high actuation force, fast response	SMA, EAP
State sensing	High sensitivity, stability, repeatability	Piezoresistive/conductive fibers, piezoelectric materials
Stiffness modulation	Reversible stimulus response, wide tuning range	MR/ER
Long-term durability and reliability	Self-healing, fatigue resistance	Self-healing polymers
Multi-functional integration	Material compatibility, functional synergy	Multi-material composite systems

2.1 Shape Memory Alloys (SMAs)

SMAs, particularly Ni-Ti alloys, are widely investigated smart materials for artificial-tendon applications. Their actuation is driven by reversible thermoelastic phase transformation between martensite and austenite [23]. A pre-strained SMA

wire, when heated via Joule heating to above the austenite transformation temperature, recovers its trained shape and generates contractile force. Upon cooling, an external bias mechanism is required for restoration. The functional performance of Ni-Ti-based SMAs also depends on alloy composition, microstructure, and processing history. Ni-Ti-Cu alloys can be fabricated by vacuum induction melting, while electric discharge drilling has been investigated for producing small holes and complex geometries [24,25]. This electro-thermo-mechanical conversion endows SMA wires with self-actuation features, enabling tension generation independent of external motors or gearboxes. The phase-transformation mechanism and representative SMA tendon configurations are illustrated in Figure 1.

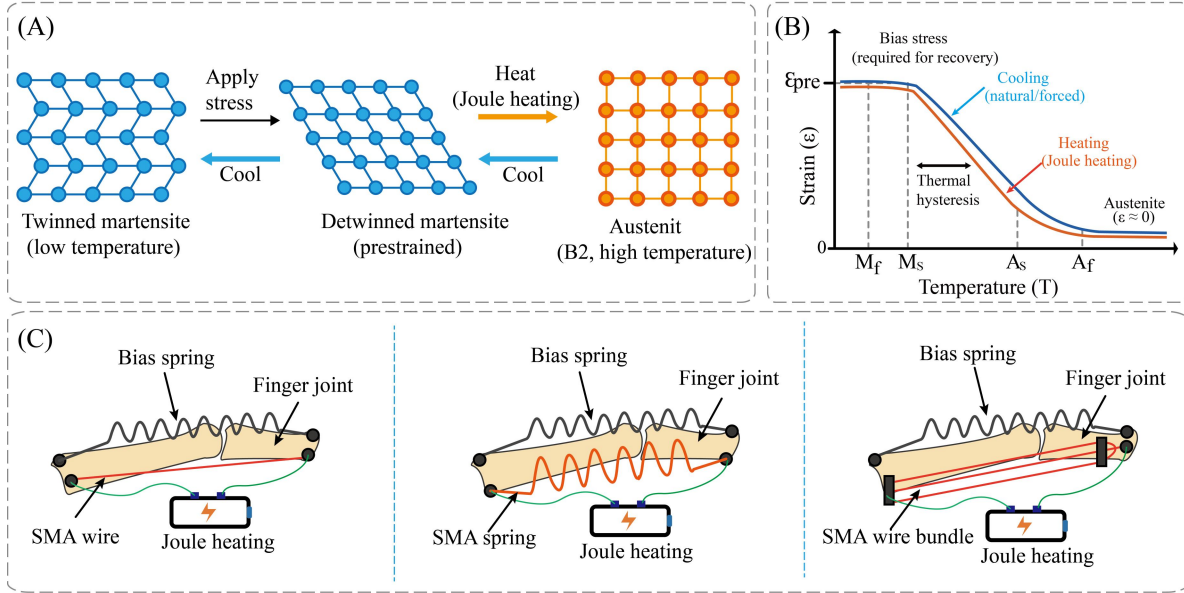


Figure 1. Shape memory mechanism and representative SMA tendon configurations. (A) Thermoelastic martensitic transformation between twinned martensite, detwinned martensite (pre-strained), and austenite (B2, high T). (B) Temperature-strain path during Joule heating and cooling, showing thermal hysteresis. A bias stress is required to restore pre-strain after cooling. (C) Three representative SMA tendon configurations in prosthetic fingers: straight wire, coiled spring, and wire bundle, all with a bias spring for restoration and Joule heating for actuation.

Several approaches have been explored to integrate SMA wires into tendon-driven systems. A relatively straightforward configuration uses pre-strained SMA wires directly as tendons connected to finger joints. Contraction upon energization drives finger flexion, while the bias spring or elastic joint restores extension upon de-energization. Liu et al. validated the feasibility of this approach in robotic fingers [26]. However, the limited stroke of single SMA wires has motivated two improvement strategies. One strategy involves coiling the wire into a spring configuration, which converts small material strain into larger axial displacement. Yang et al. designed a 19 degrees of freedom (19-DOF) prosthetic hand using dual SMA spring actuators for each finger, integrating 38 SMA actuators with a total weight of only 370 g, achieving dexterous manipulation tasks such as combing hair, writing, and playing chess through a biomimetic tendon transmission system [12]. The other strategy employs wire bundles, connecting multiple thin SMA wires in parallel to increase total contractile force while maintaining small wire diameters for rapid cooling. Bishay et al. adopted this approach in the MataPro-1 biomimetic prosthetic arm, routing multiple SMA wires in parallel within the forearm with fan-assisted cooling to achieve effective grasping of various everyday objects [27].

Cooling rate stands as a major technical bottleneck restricting SMA actuator performance. Heating can be accomplished within a very short time by increasing driving current, but cooling relies on natural convection or heat dissipation, which is considerably slower than heating. To address this issue, Oh et al. [28] developed a copper nanowire-Ni-Ti hybrid artificial muscle by growing a copper nanowire forest directly on the surface of SMA coils, increasing the effective heat dissipation area by 15-fold, doubling actuation speed under natural cooling and improving it by 170% under forced-air cooling. The prosthetic hand based on this technology could perform a series of sign language gestures and play the piano within 4 seconds. The 19-DOF prosthetic hand from the University of Science and Technology of China (USTC) adopted a system-level approach with an integrated cooling fan in the forearm to enhance interlayer air circulation and accelerate cooling of the SMA spring array [12].

Furthermore, SMA wires undergo functional fatigue during repeated phase transformation cycles, manifested as reduced contraction, incomplete recovery, and even fracture. Studies have shown that under optimized actuation parameters (e.g., 0.7 W power and 4% pre-strain), the recovery stress of Ni-Ti SMA wires remains stable over 10,000 cycles [29]; Ti-Zr-Hf-Ni-Co-Cu high-entropy SMAs retain 3.1% recoverable strain after 10,000 loading-unloading cycles [30]. However, under test conditions more closely resembling actual prosthetic service environments, SMA-driven prosthetic fingers showed a decline in bending angle from 60° to 40°-45° and a reduction in stress resistance from 50 MPa to 40 MPa after only 100 cycles [31].

2.2 Smart Fibers

Smart fibers are a core material category for achieving "self-sensing" functionality in smart tendon systems, capable of sensing their own mechanical state through changes in their electrical signals [32]. Based on their sensing mechanisms, smart fibers can be divided into four types: conductive/piezoresistive fibers, capacitive fibers, triboelectric fibers, and piezoelectric fibers. The representative sensing mechanisms of these smart fibers are illustrated in Figure 2.

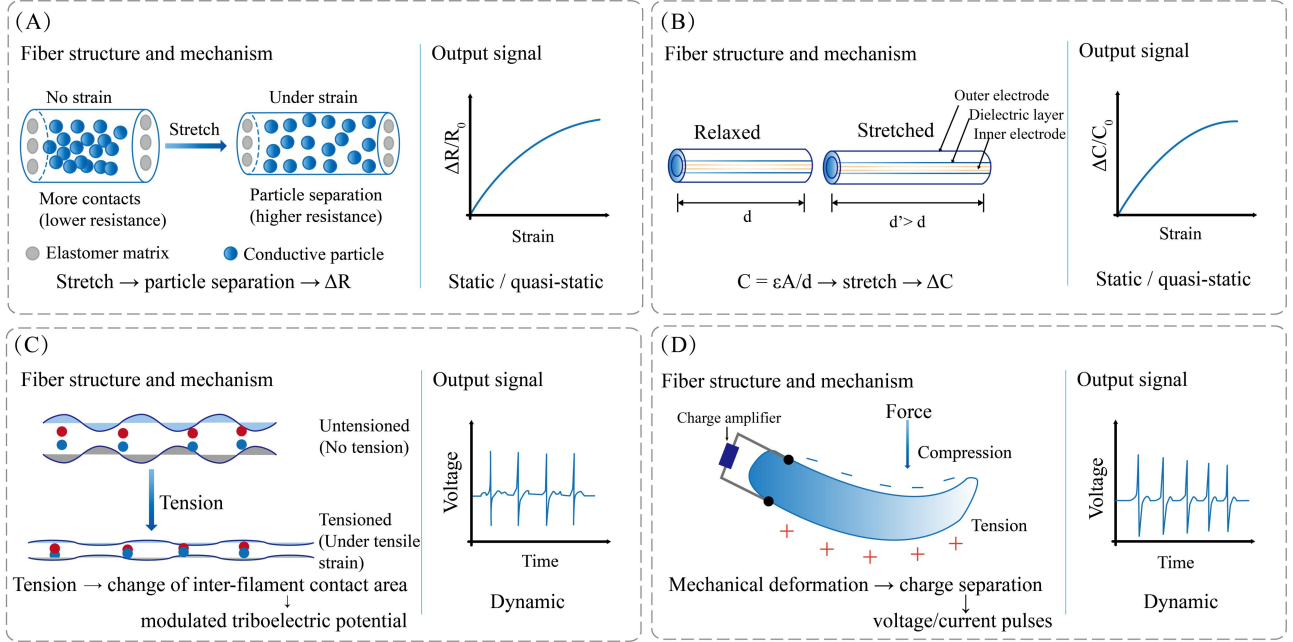


Figure 2. Sensing mechanisms of four types of smart fibers. (A) Piezoresistive/conductive and (B) capacitive fibers for static/quasi-static tension sensing. (C) Triboelectric and (D) piezoelectric fibers for dynamic tension, slip, and contact sensing.

Currently, the application of smart fibers in prosthetic hands is primarily focused on surface strain, tactile, and slip sensing, while studies directly using them as the tendon body for tension self-sensing remain limited. Therefore, this section includes both fiber sensing applications in prosthetic hands and direct sensing studies in other artificial tendon or fiber carriers, to distinguish existing application evidence from potential migration directions.

For conductive/piezoresistive fibers, the sensing principle is based on the piezoresistive effect. Georgopoulou et al. integrated thermoplastic polystyrene-based piezoresistive sensing fibers onto the dorsal side of a flexible prosthetic finger, demonstrating the application of piezoresistive fibers in finger bending detection. The fiber is indeed in "fiber" form and is integrated into the prosthetic hand structure, but its function is as a sensing layer in the prosthetic structure, not as the tendon body [15].

For capacitive strain sensors, Yang et al. integrated a poly(3,4-ethylenedioxythiophene):poly(styrenesulfonate) (PEDOT:PSS)-based strain sensor into a surgical suture. The suture or fiber form used in this study closely resembles tendon morphology and enables in situ monitoring of tendon and ligament strain, but it targets post-operative monitoring scenarios rather than prosthetic hand drive systems [33].

For triboelectric tension-sensing fibers, the Chinese University of Hong Kong team proposed a bio-inspired "Exo Tendon" that senses tension through triboelectric effects to achieve self-force-regulated actuation [34]. The name "Exo Tendon" clearly points to the "tendon" form, and it has been integrated into an artificial muscle-driven wearable robotic system, with potential for migration to prosthetic hand tendons.

For piezoelectric fibers, the sensing principle of piezoelectric fibers is based on the positive piezoelectric effect. Unlike the piezoresistive effect, piezoelectric effect exhibits high sensitivity toward dynamic forces yet shows poor responsiveness to static or quasi-static deformation. This characteristic makes it highly suitable for detecting transient events such as contact initiation, object slip, and vibration—information that is crucial for prosthetic hands to prevent objects from slipping and to adjust grip force during grasping. Polyvinylidene fluoride (PVDF) and its copolymer poly(vinylidene fluoride-trifluoroethylene) (P(VDF-TrFE)) are currently the most commonly used piezoelectric fiber materials, which can be fabricated into nanofiber membranes or fiber forms via electrospinning, offering excellent flexibility and biocompatibility. Lee et al. developed a cladding-free thermal drawing process for scalable reel-to-reel manufacturing of piezoelectric coaxial PVDF fibers, demonstrating that the fiber possesses complete "fiber" morphology and high flexibility [35]. In terms of prosthetic hand system integration, Khalifeh et al. installed a PVDF-based flexible piezoelectric tactile sensing system on the fingertips of the Hannes bionic prosthetic hand for object hardness and texture recognition, demonstrating that piezoelectric materials have been applied in prosthetic hands, but their function is tactile sensing, not tendon tension sensing [36].

Conductive/piezoresistive fibers, capacitive fibers, triboelectric fibers, and piezoelectric fibers form a clear functional complementarity: piezoresistive and capacitive fibers provide static or quasi-static information about "how much force/deformation is currently present," while triboelectric and piezoelectric fibers provide dynamic information about "how the force/deformation is changing." Their combination provides a material foundation for constructing smart tendons capable of both static state monitoring and dynamic event recognition.

2.3 Electroactive Polymers (EAPs)

EAPs represent another class of smart actuation materials with tendon-compatible potential, capable of generating reversible deformation through electro-mechanical coupling mechanisms including electric fields, electrothermal effects, or ionic migration. Based on their actuation mechanisms, EAPs are broadly classified into electronic-type and ionic-type EAPs. Electronic-type EAPs, including dielectric elastomers (DEs), liquid crystal elastomers (LCEs), and ferroelectric polymers, offer fast response and high energy conversion efficiency but typically require kilovolt-level driving voltages. Ionic-type EAPs, including ionic polymer-metal composites (IPMCs), conductive polymers, and ionic gels, can be actuated at low voltages but exhibit slower response and suffer from long-term stability issues due to electrolyte evaporation. The principal actuation mechanisms of representative EAPs are shown in Figure 3.

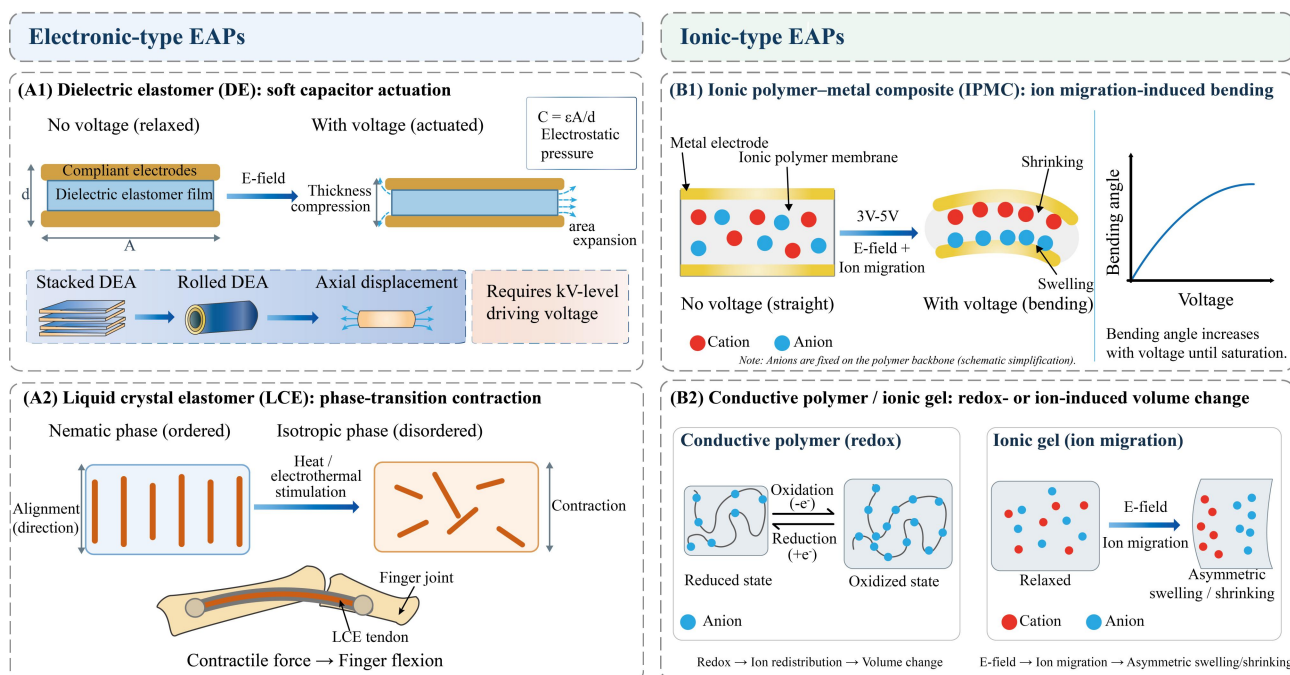


Figure 3. Actuation mechanisms of EAPs. (A1) DE: voltage-induced thickness compression and area expansion. (A2) LCE: phase-transition contraction; tendon-form demonstrated. (B1) IPMC: cation migration causes asymmetric swelling and bending. (B2) Conductive polymer/ionic gel: redox or ion migration induces volume change.

Among electronic-type EAPs, LCEs offer some of the most compelling reported evidence for the artificial-tendon concept. Lu et al. reported an LCE artificial tendon ("LCE tendon") integrated into a biomimetic prosthetic hand, where the LCE generates contractile force through reversible phase transition under thermal or electrothermal stimulation, enabling finger flexion grasping and object manipulation [13]. This study directly demonstrates that LCEs can be integrated into prosthetic hands in "tendon" form for actuation, making them the second smart material, after SMAs, confirmed to be directly usable in the "tendon" morphology for prosthetic applications. Liu et al. further developed liquid crystal elastomer optical fiber artificial muscles, achieving multimodal actuation and precise control [37].

DEs are another important class of electronic-type EAPs. Their working principle resembles that of a soft capacitor: a dielectric elastomer film sandwiched between two compliant electrodes undergoes thickness compression and area expansion under electrostatic forces upon voltage application, achieving in-plane strain that can exceed 200% [38]. Mao et al. designed a self-healing dielectric elastomer with a bimodal network structure, achieving 125% stable actuation strain at 25 MV/m [39]. Shi et al. developed a dielectric elastomer with a semiseparated biphasic bicontinuous structure, achieving 90% in-plane strain at a low electric field of 35 V/ μm without pre-stretching, with an electromechanical sensitivity of 360 MPa $^{-1}$, combined with high energy density (375 J/kg), high power density, and ultra-long lifetime [40]. However, the inherent nature of DEs is area expansion rather than linear contraction, requiring structural designs such as stacking or rolling to convert into the axial displacement needed for tendons, which increases system complexity. At the prosthetic application level, stacked dielectric elastomer actuators (DEAs) have been integrated into the 3D-printed pediatric prosthetic hand Cyborg Beast [41], and fiber-constrained DEAs have been designed as antagonistic artificial muscles for finger prostheses [42], but these applications remain primarily at the proof-of-concept stage.

Among ionic-type EAPs, IPMCs can generate bending deformation under low voltages (typically 3-5 V). Kim et al.

constructed fibrillar ion-transport networks that enabled IPMCs to achieve stable actuation frequencies up to 50 Hz in air with over 30,000 cycles [43]. Conductive polymers (e.g., polypyrrole, PEDOT:PSS) generate actuation through volume changes induced by electrochemical redox reactions, with self-sensing potential [44]. Ionic gels represent a rapidly emerging direction, with 2,2,6,6-tetramethylpiperidine-1-oxyl (TEMPO)-oxidized cellulose nanofiber-based ionic gels achieving 1.11% strain at 2.8 V and exceeding 10,000 cycles of lifetime [45]. However, ionic-type EAPs currently remain primarily at the material-level performance characterization stage, and systematic validations within prosthetic-hand tendon systems are still scarce.

The current limitations of EAPs can be summarized in three aspects. First, electronic-type EAPs require kilovolt-level high voltages, posing challenges for portable power supplies in prosthetic hands, while ionic-type EAPs, though operable at low voltages, offer limited force output. Second, ionic-type EAPs are susceptible to performance degradation due to solvent evaporation in open environments, while electronic-type EAPs face mechanical fatigue and electrical breakdown issues. Third, the inherent area-expansion nature of DEAs requires stacking, rolling, or other structural designs to convert into the axial displacement required for tendons, adding system complexity.

2.4 Magnetorheological/Electrorheological (MR/ER) Materials and Self-Healing Materials

When grasping different objects, prosthetic hands must maintain sufficient stiffness to effectively transmit force at the moment of grasping, while also providing compliance to avoid damage when contacting fragile objects or human body parts. Meanwhile, during long-term repetitive use, microcrack accumulation and fatigue fracture of tendons are practical engineering problems affecting prosthetic hand reliability. MR/ER materials and self-healing materials offer unique solutions to these two types of requirements, respectively. The mechanisms of tunable stiffness and self-healing are illustrated in Figure 4.

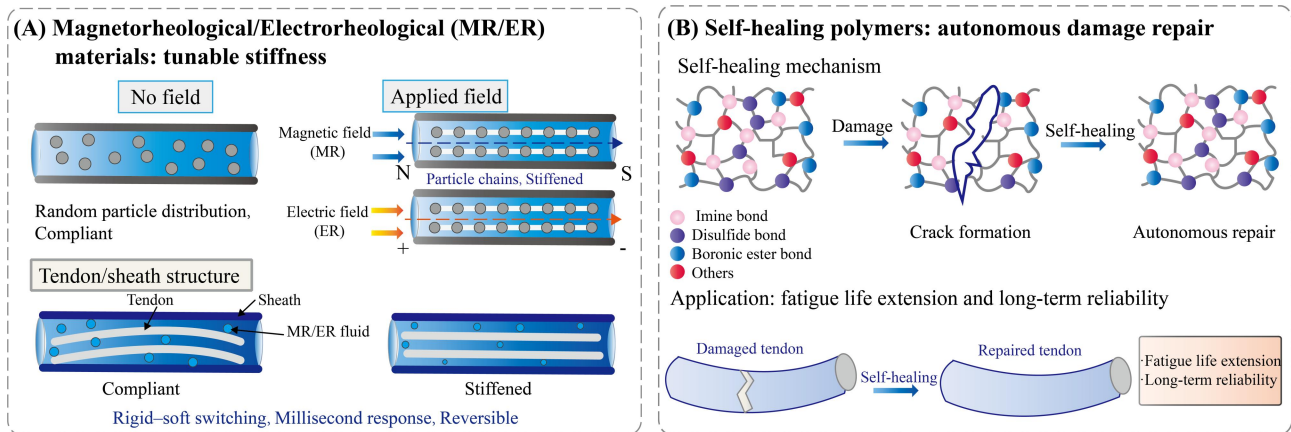


Figure 4. Tunable stiffness and self-healing mechanisms. (A) MR/ER materials: field-induced particle chaining enables reversible stiffness switching. (B) Self-healing polymers: reversible bonds restore mechanical integrity after damage.

MR and ER materials are core material categories for achieving tunable tendon stiffness [46,47]. Their working principle is based on the controllable rheological effect: when a magnetic or electric field is applied to MR or ER fluids, the magnetic or polar particles within align along the field direction to form chain-like structures, causing reversible, millisecond-scale changes in viscosity and yield stress [46]. When this effect is integrated into tendon or tendon-sheath structures, the equivalent stiffness of the tendon can be dynamically modulated by external field strength. Specifically, the tendon remains compliant without the field, and "solidifies" to become stiff upon field application. Current research predominantly integrates MR/ER materials into clutches, actuators, or sheath structures, rather than directly fabricating them into "rope"-like independent force-transmission elements. Lu et al. employed a magnetorheological cable-driven mechanism, in which MR materials are integrated into the drive mechanism rather than the tendon body itself, to achieve an adjustable rigid-soft coupled wearable hand robot [48]. Although this study targets stroke rehabilitation training scenarios rather than prosthetic hands, its design concept has potential for migration to prosthetic hand tendons. A non-motorized hand exoskeleton based on MR actuators achieved a peak holding force of 1046 N at only 5 W power input, demonstrating the potential of MR technology for efficient grip holding [49], also belonging to the exoskeleton category. In ER materials, Xu et al. developed an "electro-stiffened ribbon actuator" achieving rapid stiffness modulation in less than 10 ms [50]; an ER-based dual-mode variable-stiffness soft gripper achieved fast, wide-range, high-precision stiffness modulation [51].

Self-healing polymers represent another class of smart materials targeting long-term tendon reliability. By introducing reversible covalent bonds (such as imine, disulfide, or boronic ester bonds) or supramolecular interactions into polymer networks, these materials can autonomously repair cracks and restore mechanical integrity after damage. Huang et al. developed a self-healing tough actuator material with muscle-like relaxation and contraction via multilevel relaxations, achieving a tensile strength of 33.5 MPa and a mechanical self-healing efficiency of 89.7% [52]. An integrated self-healing dielectric elastomer (ISDEA) was reported, in which both the dielectric film and compliant electrodes possess self-healing capabilities, with an interfacial bonding force of 4.12 N and an overall self-healing efficiency exceeding

92.1%; a finger rehabilitation trainer based on this technology has been used as a motion-assistive soft robot for patients with spinal muscular atrophy [53]. A poly(ionic liquid) electrode was developed for dielectric elastomer actuators, with an elastic modulus of only 3.4 kPa, capable of rapid self-healing within 10 seconds at room temperature and underwater, enabling DEAs to maintain functionality after damage [54]. However, self-healing polymer research currently remains largely at the materials science level. Although excellent repair performance has been demonstrated in film, bulk, and hydrogel forms, reports simultaneously addressing mechanical strength, repair efficiency under cyclic actuation, and long-term cyclic stability in fiber or tendon form remain limited. Direct application evidence of self-healing fibers as artificial tendons is extremely scarce, and their more accurate positioning is as a forward-looking material direction for next-generation smart tendons.

2.5 Multi-Material Composite Smart Tendons

The preceding sections have covered the actuation functions of SMAs and EAPs, the sensing functions of smart fibers, and the performance modulation of variable-stiffness and self-healing materials. However, a practical prosthetic hand tendon system demands synergy across multiple functions: actuation, sensing, stiffness modulation, and beyond. Multi-material composites at the material level, integrating different smart materials into a single tendon structure to carry multiple functions, offer an optimal solution. This direction remains in its early exploratory stage, with its core value lying in providing transferable design paradigms.

In the direction of actuation-sensing integration, Kwak et al. developed a self-sensing SMA-textile composite soft actuator that integrates SMA wires and conductive fabric into a polydimethylsiloxane matrix, achieving simultaneous actuation and sensing in bending, torsion, and combined deformation modes [55]. Experiments confirmed consistent correlation between resistance changes and bending curvature/torsion angle, and the actuator was applied to soft grippers. The Seoul National University team developed a "smart artificial muscle" [56], integrating LCEs in series with two embedded liquid metal channels, with one channel serving as the active actuator and the other as the sensor, thereby achieving actuation-sensing integration within a single fiber-like form. The robot finger equipped with this muscle can finely grasp objects and autonomously distinguish their stiffness and size. This work provides a transferable composite material design paradigm for actuation-sensing integrated smart tendons.

In the area of SMA-based composite tendons, Li et al. performed surface functionalization modification of SMA fibers via polydopamine-integrated silver nanowire decoration, achieving approximately 3.2-fold improvement in near-infrared actuation speed and 35% improvement in electrothermal response speed [57]. This surface functionalization strategy essentially integrates "photothermal response" functionality onto the SMA's "actuation" function, offering a new technical pathway for multi-material composites. A highly stretchable fibrous actuator was reported [58], combining SMA's temperature-induced phase transformation with elastomeric polymer chain rearrangement to overcome the limitations of SMA's inherent rigidity for soft actuator applications. In the actuation-stiffness modulation direction, superelastic SMAs were used to develop tendon-like Bowden cable systems [4], validating their performance advantages in hand-assistive exoskeletons.

3. Application and Performance Comparison of Smart Tendons in Prosthetic Hands

3.1 Prosthetic Hand Actuation

Finger flexion, thumb opposition, and underactuated grasping represent three fundamental actuation requirements of prosthetic hands, each imposing distinct demands on tendon materials.

Finger flexion requires tendons to generate sufficient tensile force within compact spaces while achieving adequate stroke, with consideration of fatigue life under repeated flexion-extension cycles. For this function, SMAs currently offer the strongest evidence. Their high power density enabling sufficient tensile force at small diameters [12,59], and their multiple configurations—straight wire, spring, and bundle—accommodate different stroke and force requirements [60-62]. EAPs offer advantages in response speed and compliance [43,63], but their lower force density makes them more suitable for scenarios with modest force requirements [42]. For high-frequency continuous grasping tasks, SMAs are better suited for intermittent or assistive actuation, whereas conventional actuators such as motors can handle high-frequency motions. A hybrid use of both may represent an engineering-feasible solution.

Thumb opposition prioritizes positioning accuracy and posture maintenance over force magnitude. SMAs require closed-loop control or mechanical locking to compensate for position errors caused by thermal hysteresis in this scenario [12], while superelastic SMAs or MR/ER-assisted locking mechanisms may be more appropriate alternatives.

Underactuated grasping emphasizes the passive compliance of tendons: the ability to adaptively distribute forces upon contact with objects. The inherent compliance of SMAs provides a natural advantage in this scenario [59]. For scenarios requiring greater compliance, EAPs and superelastic SMAs can serve as complementary options.

3.2 Prosthetic Hand Sensing and Closed-Loop Control

Prosthetic hand sensing functions primarily include tendon tension sensing, grip force estimation, and dynamic

contact/slip detection, with different sensing requirements demanding distinct material properties.

For tendon tension sensing, piezoresistive/conductive fibers offer the direct advantage of "force transmission with intrinsic sensing." As previously described integration of piezoresistive fibers on the dorsal side of prosthetic fingers has demonstrated that object size differences can be distinguished through resistance changes [15]. This approach eliminates the need for additional sensors, consumes no internal space within the prosthetic hand, and is particularly suitable for spatially constrained finger segments. Piezoelectric fibers, due to their sensitivity only to dynamic forces, primarily provide information on contact initiation and slip events [35,36], complementing piezoresistive fibers.

For grip force estimation, the static strain-sensing capability of piezoresistive fibers can provide continuous force information, though the nonlinearity and temperature drift of the resistance-strain relationship remain engineering concerns [15]. Piezoelectric fibers, while unable to provide static force information, are highly suitable for detecting the onset of slip due to their high-frequency response characteristics.

For closed-loop control, sensory information must be incorporated into the control loop to realize its value. In the cases described above, the SMA prosthetic hand achieves joint-angle closed-loop control through Hall sensors [12], while piezoresistive fibers enable tension feedback through resistance changes [64], both representing concrete implementations of using sensory information to modulate actuation inputs. The establishment of the complete "sensing-control-grasping feedback" chain represents a critical step in the evolution of smart tendons from "functional materials" to "intelligent systems."

3.3 Compliant Interaction and Adaptive Grasping

Prosthetic hands must accommodate a wide range of stiffness requirements when grasping different objects—from fragile items (e.g., eggs) to rigid tools (e.g., metal implements). Using fixed-stiffness tendons to address all these scenarios represents an inherent compromise, sacrificing either the safety of fragile objects or the grip strength for heavy objects.

MR/ER materials address this challenge by enabling the equivalent stiffness of tendons to become a "tunable variable" through external-field control of viscosity and yield stress [51,65]. The MR cable-driven mechanism [48] and ER dual-mode variable-stiffness soft gripper [50] described above have both validated the feasibility of this concept at the prototype level. However, integrating the MR/ER effect into "tendon" form rather than "actuator" or "joint" form remains an open challenge [65].

Superelastic SMAs and variable-stiffness polymers can also provide a degree of stiffness modulation [4], though their modulation range and response speed are generally inferior to those of MR/ER materials [65]. From a system integration perspective, integrating stiffness modulation and actuation functions into the same tendon (e.g., SMA + MR composite tendons) represents a promising future direction.

3.4 Comprehensive Comparison and Material Selection Considerations

Table 2 provides a quantitative comparison of representative smart materials and tendon-related systems in terms of actuation or sensing strain, stress/force or sensitivity, response characteristics, operating conditions, durability, energy or power requirements, and demonstrated application level. The comparison indicates that different material classes offer distinct functional advantages, but these advantages must be weighed against system-level engineering trade-offs. SMA-based systems provide relatively high actuation force and are relatively mature for integration into prosthetic hands; however, they operate at low voltage but require high current (e.g., 0.2 A per wire, 1.8 A per bundle, or 4 A pulses), and their Joule heating consumes up to 27 W per finger and 1100 W/kg, necessitating forced-air cooling, additional fans, and larger battery capacity, which increase weight, volume, and thermal management complexity. Moreover, their fatigue life drops to only 100 cycles under realistic prosthetic conditions. Smart fibers are attractive for lightweight force transmission and intrinsic tension sensing, but their integration as tendon bodies remains limited, and multi-channel signal acquisition and processing add control complexity. EAPs offer large deformation and compliant actuation, but DEs require kilovolt-level electric fields (e.g., 35-60 V/ μm), necessitating bulky high-voltage circuitry, while ionic-type EAPs suffer from electrolyte evaporation and limited force output, affecting long-term reliability and maintenance. MR/ER materials provide tunable mechanical stiffness for adaptive interaction, but they require electromagnetic coils (1-3 A) or high-voltage sources (0-12 kV) for field generation, which introduces additional power supplies and thermal loads; moreover, their integration into tendon-like structures remains less mature. Self-healing materials mainly contribute to damage tolerance and service-life enhancement, yet they are currently demonstrated predominantly at the material or actuator level, with self-healing efficiencies of 89.7-92.1%. Their cost, specialized processing, and encapsulation requirements remain challenging for fiber/tendon-form deployment.

Therefore, material selection should be based on the required tendon function and system constraints rather than a single performance metric. Overall, the current literature suggests that multifunctional tendon architectures combining complementary smart materials may provide a more practical route toward simultaneous force transmission, sensing, adaptive stiffness, and durability in prosthetic hands.

Table 2. Quantitative performance comparison of representative smart materials for artificial-tendon applications in prosthetic hands.

Category	Material	Main function	Actuation/ Sensing strain	Stress/Force/ Sensitivity	Response characteristics	Operating conditions	Durability	Energy/ Power	Application
SMA	SMA springs [12]	Actuation	-	2.5 kg load; 34 MPa at 7 N	Voice-actuation ≈ 1.3 s; $63^\circ/\text{s}$; error $<0.5^\circ$	Joule heating; 90°C ; forced air cooling	10,000 cycles at 7 N	640 grasps	Prosthetic hand
	Ti-Ni-Cu SMA wire [14]	Actuation	Contraction strain 5%	-	-	15 V; PWM 1526 Hz	-	-	Robotic finger
	Ni-Ti-Cu SMA wire [26]	Actuation	Contraction strain 4.0%	144 gf (80 MPa); breaking 1.8 kgf	Activation 0.3 s; stabilization 1.5 s; relaxation 2 s	20.7 V/m; 7.05 W/m; $>105^\circ\text{C}$	10^6 cycles	7.05 W/m	Robotic finger
	Flexinol SMA wire [27]	Actuation	Max strain 4%	Fingertip force 7.5-23.7 N; total force 71.8 N; single wire ≈ 35 N	Actuation <1 s; retraction 4-5 s	4 A pulse, 2 s, 50% duty; fan 12 V	-	-	Prosthetic hand
	SMA + Cu nanowire [28]	Actuation/fast cooling	Max contraction strain 55.95%	≈ 2.45 N	Natural cooling 1.7 mm/s; forced 6.6 mm/s; cooling acceleration 100%-170%	1.0-2.4 V; natural/forced air 0.05-0.2 m/s	200 cycles	1100 W/kg	Prosthetic hand
	PDA-AgNWs coated SMA coil [57]	Actuation/thermal response	-	0.77 N at 40%; 1.72 N at 200%; grasping force 4.6 N	NIR speed $3.2\times$; electrothermal response $+35\%$	808 nm NIR, 13.75 mW/mm ²	100 cycles	-	Robotic hand
	Arc-shaped SMA wire [59]	Joint actuation	Max recoverable strain 4.1%; rotation $\approx 100^\circ$	Fingertip force 39.465 N; thumb-side 17 N	Fast heating; $\approx 100^\circ\text{C}$; control $65-85^\circ\text{C}$	>1 A; PWM	-	11.7-36 W	Dexterous hand
	Coiled SMA [58]	Actuation	Contraction displacement ≈ 50 mm	Single muscle 1.11 N; grasp 133 g; axial 40 N	Max frequency 0.125 Hz; heating 2 s; cooling 8 s	0.3-0.7 A; 3.6-4.4 V	-	Avg. 7.82 W	Soft robotic hand
	SMA wire [60]	Actuation	3.5% (safe)	Single finger ≈ 9 N; whole hand ≈ 45 N	Pulse 0.2-0.3 s; regular 1.6-2 s	0.2 A/wire; 1.8 A/bundle	-	≈ 27 W/finger	Five-finger soft hand
Smart fiber	Spider-silk composite [11]	Force transmission + conduction + sensing	Sensing 0-15%; breaking $\approx 60\%$	Toughness 420 MJ/m ³ ; 1077 S/cm; lift 7.6 kg	Finger $8^\circ \rightarrow 73^\circ \approx 1.5$ s; force feedback <4 ms	Mechanical stretch/bend; 0-1 kPa	$>40,000$ cycles	-	Bionic robotic hand
	TPU/TPS + CB [15]	Strain/temperature/tactile sensing	Strain 0-30%	GF 0.4-47; tactile force ≤ 2.5 N	Temperature detection 10 ms	25-170°C; mechanical strain	-	-	Prosthetic hand
	Triboelectric fiber [34]	Tension sensing	Tension sensing: 10-34 N; resolution: 0.02 N	10-34 N; resolution 0.02 N; sensitivity 1.59 V/N·m; $R^2 = 0.9982$	8 Hz stable output	Mechanical tension; RH $\leq 90\%$	>6000 cycles no decay; 60,000 cycles decay $<5\%$	-	Wearable exoskeleton
	PVDF fiber [35]	Strain/pressure sensing	Sensing 0.25-1.5%	6.76-14.85 V; 9.42 V at 0.5% strain, 1 Hz; γ phase $>82\%$	0.5-2.5 Hz	Mechanical stretch/bend	10,000 cycles; output retention 87.3%	-	Wearable sensing
	PVDF [36]	Tactile sensing	-	Force 0.12-6 N; 1.5-31.96 pC/N; linearity error 2.3%	100 ms window; 101 ms delay	Mechanical contact; 500 Hz sampling	-	72.3 mW; 63.8 $\mu\text{J}/\text{inference}$	Prosthetic hand tactile recognition
EAPs	DE [38]	Actuation	Area strain 200% at 60 V/ μm	0.95 N; blocking stress ≈ 0.23 MPa	1-100 Hz	60 V/ μm	10,000 cycles	283 J/kg; 1701 W/kg	Soft robot; gripper
	DE [40]	Actuation	Area strain 90% at 35 V/ μm ; linear $>50\%$	electromechanical sensitivity: 360 MPa ⁻¹	Frequency up to 260 Hz; strain rate 400%/s	35 V/ μm	$>375,000$ cycles at 1 Hz; $>10^8$ cycles at 260 Hz	209-375 J/kg	Robotic arm; soft robot
	DEA [42]	Actuation	-	P = 2.94 N; $\mu = 31$ kPa	-	High voltage	-	-	Prosthetic finger
MR/ER	MR fibrous material [16]	Responsive/tunable properties	linear stroke 5 mm	Torque density ≈ 7 N m kg ⁻¹ ; bending stiffness $\approx 30\times$; shear force 110 mN; compression modulus 0.45-22.5 kPa; linear force 150 mN; tactile normal force 150 mN	0.07-0.23 s; 0.1-10 Hz	≤ 300 mT; 1-3 A	10,000 cycles	-	Soft robot; conformal grasping; tactile finger sleeve

Category	Material	Main function	Actuation/ Sensing strain	Stress/Force/ Sensitivity	Response characteristics	Operating conditions	Durability	Energy/ Power	Application
	MR + cable [46]	Variable stiffness + cable drive	Circumferential/lateral deformation 2-15 mm	MRC max torque 520 mN·m; robotic fingertip force 5.8 N	Fast adjustment	0.05-1 A; current/magnetic field	Stable continuous operation	-	Hand assist robot; wearable robotic hand; hand exoskeleton
	ER fluid [51]	Variable stiffness	compression displacement 0-10 mm	Normal force 0.38→3.55 N; normal stiffness 0.02→0.73 N/mm (36.5×)	VSM $\approx$ 0.1 s; recovery 3.29-15.68 s	0-12 kV	-	-	Soft gripper
Self-healing	Self-healing DE [39]	Actuation + self-healing	Max area strain 272% at 32.5 MV/m; stable strain 125% at 25 MV/m	lifts 320× its own weight; 115 g load	2.5-5 Hz	25-32.5 MV/m; self-healing 60°C	150,000 cycles; total lifetime >18 0,000 cycles	specific energy 152 J/kg; specific power 1520 W/kg	Self-healing
	Self-healing actuator [52]	Actuation + self-healing	Relaxation 30.3%; contraction 36.8%	Tensile strength 33.5 MPa; self-healing efficiency 89.7%	Cycle deviation 0.54%-2.09%	NIR 808 nm; $\leq$ 82°C	Self-healing efficiency 89.7%; 3 actuation cycles stable	0.56/3.68 W	
	Self-healing DE [53]	Actuation + self-healing + self-powered + flexible capacitive sensing	18.46% at 27.01 kV/mm	Interfacial adhesion 4.12 N; self-healing efficiency > 92.1%	Sensor 0.34-0.79 s	DC 27.88 kV/mm; TENG 27.01 kV/mm; 5.2 kV; 14.7 kV/mm; 60 °C self-healing	TENG 10,000 cycles; self-healing 5×	<10 μ A (TENG)	
	Self-healing ionic polymer [54]	Electrode + self-healing	DEA area strain 63.2%; bending displacement 1.74 mm	blocking force 1.27 mN $\rightarrow$ 1.13 mN	Self-healing 10 s; resistance recovery $\approx$ 7 s; bending equilibrium $\approx$ 2 s	11-32 V/ μ m; -20-80 °C	50,000 cycles; 5× healing	low and short-term energy	
Multi-material composite	SMA + conductive fabric [55]	Actuation + self-sensing	Bending curvature $\leq$ 0.025 mm ⁻¹ ; torsion $\leq$ 90°	GF 32.69-81.48; hysteresis 17.6-57.9%	Response 1.81-6.37 s; cooling 30-40 s	0.9 A; SMA $\approx$ 60 °C	100 cycles	-	Soft gripper

Note: The values reported in the table are representative experimental results from the cited studies and should not be interpreted as intrinsic material limits. Differences in material composition, specimen geometry, pre-strain, loading conditions, and testing methods should be considered when comparing values across studies. “-” indicates that the corresponding parameter was not reported in the cited study. The application column indicates the highest demonstrated application level reported in the corresponding study rather than a prospective or target application.

Abbreviations: SMA, shape memory alloy; EAP, electroactive polymer; DE, dielectric elastomer; DEA, dielectric elastomer actuator; MR, magnetorheological; ER, electrorheological; MRC, magnetorheological clutch; PDA, polydopamine; AgNWs, silver nanowires; NIR, near-infrared; PWM, pulse width modulation; TPU, thermoplastic polyurethane; TPS, thermoplastic starch; CB, carbon black; GF, gauge factor; PVDF, polyvinylidene fluoride; RH, relative humidity; VSM, variable stiffness modulation; TENG, triboelectric nanogenerator; DC, direct current.

4. Conclusion

The introduction of smart materials is driving the evolution of artificial tendons from passive force-transmission elements toward active multifunctional units capable of actuation, sensing, and modulation. This review has surveyed SMAs, smart fibers, EAPs, variable-stiffness materials, self-healing polymers, and multi-material composites in tendon-driven prosthetic hands. These material systems exhibit different levels of maturity and functional emphasis. SMAs are the most mature actuation option for prosthetic hand tendon-driven systems, attributed to their high power density and multiple tendon configurations. EAPs show promise in compliant actuation with limited engineering readiness. MR/ER materials and self-healing polymers extend smart tendon functionalities from environmental adaptability and long-term reliability perspectives. Multi-material composites are advancing these functions toward system-level integration.

Despite these advances, several challenges remain for engineering deployment: the inherent conflict between actuation and sensing within the same tendon; the lower cycle life of SMAs under real service conditions; the increased system complexity from high-voltage EAP driving and MR/ER field-generation devices; and higher demands on control architecture and energy management. The core issue is no longer "whether a material can achieve a function," but "how multiple material functions can reliably work together within the constrained space of a prosthetic hand."

Addressing these challenges requires coordinated efforts covering material development and structural design. New intrinsically self-sensing actuation materials, as well as fatigue-resistant self-healing fibers with condition monitoring and lifetime prediction, may resolve material-level conflicts between actuation and sensing and enable controllable maintenance cycles. Structural designs via functional segmentation and interface optimization could realize powerful actuation, adaptive grasping, and safe contact.

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Ethics Statement

Not applicable. This study is a review of previously published literature and did not involve human participants, animals, biological materials, or personally identifiable data.

Informed Consent

Not applicable.

Data Availability Statement

This review is based entirely on previously published studies cited in the manuscript. No new datasets were generated during the present study.

Author Contributions

Linghui Cheng conducted the literature search, screening, and analysis and prepared the original draft of the manuscript. Tian Yang reviewed the manuscript and contributed to language editing. Zheng Deng contributed to the language refinement and detailed revision of the manuscript. Qiaoling Meng contributed to the development of the review methodology and the critical revision of the manuscript. Sujiao Li and Hongliu Yu supervised the work and coordinated the overall manuscript review and revision process. All authors read and approved the final version of the manuscript.

Conflict of Interest

The authors declare no conflict of interest.

Generative AI Statement

GPT-5.6 Luna (OpenAI) and DeepSeek (DeepSeek-V3, 2025) were used during revision to assist with English-language editing, literature organization, and structural refinement of the review manuscript. These tools were not used to generate core scientific insights, experimental data, or to perform any form of analysis. All technical content, including material classifications, functional mechanism descriptions, application comparisons, cited references, and scientific interpretations, was independently verified and finalized by the authors, who assumes full responsibility for the originality, accuracy, and integrity of the manuscript.

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