

Review

Pneumatic Soft Actuation in Elbow Rehabilitation Devices: Actuator Architectures, Sensing Modalities, and Control Strategies—Scoping Review

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Abstract

Soft and compliant rehabilitation devices may provide improved anatomical adaptability, reduced distal mass, and more flexible human–robot interaction compared with conventional rigid exoskeletons. This scoping review examines elbow rehabilitation technologies across three interconnected domains: (1) the broader landscape of soft and compliant actuation, (2) the structural, material, and operating architectures of pneumatic soft actuators, and (3) the sensing, intention-detection, and closed-loop control methods used in pneumatic systems. The review included 109 peer-reviewed reports. Four main actuation families were identified: pneumatic soft actuators, motor-driven cable and tendon systems, series-elastic or variable-stiffness actuators, and shape-memory-alloy-based devices. Pneumatic architectures were primarily organized around linear artificial muscles and chamber-based bending or rotary actuators, complemented by rigid–soft integrated, cable-transmitted, modular, antagonistic, self-sensing, and variable-stiffness configurations. Feedback most relied on pressure, kinematic, force, and electromyographic signals. Control architectures combined position, force, torque, impedance, and pressure regulation, while nonlinearities and uncertainties were addressed using model-based, adaptive, sliding-mode, fuzzy, neural, and hybrid methods.

Keywords: elbow rehabilitation; pneumatic soft actuator; pneumatic artificial muscle; soft exosuit; sensing; intention detection; control architecture; scoping review

1. Introduction

As rehabilitation robotics continues to develop, soft and mechanically compliant wearable devices are receiving increasing attention alongside rigid exoskeletons. Conventional rigid robots can provide well defined kinematics, high structural stability, and accurate joint positioning. However, their weight, inertia, rigid human–robot interfaces, and the need for precise alignment with the anatomical joint axis may limit their wearability. In contrast, systems that use deformable actuators, flexible force transmission, and textile-based interfaces may accommodate variations in body geometry and joint alignment while allowing part of the actuation hardware to be positioned away from the distal limb.

However, there is no single, widely accepted definition of soft rehabilitation robots. The term “soft” may refer to the energy converting actuator, the force transmission structure, the wearable interface, or mechanical compliance intentionally incorporated into the system. For example, a textile based exosuit may use a rigid electric motor together with a flexible



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Bowden cable, while a mechanically more rigid exoskeleton may employ intrinsically compliant pneumatic artificial muscles. Soft devices should therefore be classified by treating the actuator, transmission, and human–robot interface as separate functional subsystems [1,2]. This distinction is especially important in elbow rehabilitation, where the actuator force must be converted into joint torque through an appropriate moment arm, while the device must also accommodate changes in arm shape and length.

This distinction is particularly relevant for elbow rehabilitation. Assistance generated by an actuator must be transferred to the arm through suitable attachment points and converted into joint torque through an effective moment arm. At the same time, the wearable structure must accommodate changes in limb shape, muscle volume, and segment length during movement. As a result, actuator performance alone does not determine the assistance delivered to the joint, since transmission geometry, interface deformation, and attachment stability also influence the mechanical output [3–5].

Several actuation technologies have been investigated for active elbow support, including motor-driven cable and tendon systems, series-elastic and variable-stiffness actuators, shape memory alloys, and pneumatic soft actuators. These approaches differ in how actuation, compliance, and force transmission are distributed across the device. Cable-driven systems allow the motor to be placed away from the limb, although friction, backlash, and path-dependent transmission losses may affect their performance [6,7]. Series-elastic actuators provide mechanical compliance and support indirect torque estimation, but often require a more structured mechanical frame [8,9]. Shape memory alloys enable compact and quiet actuation, although their response is limited by thermal dynamics, cooling time, and hysteresis [10,11].

Pneumatic actuation is especially relevant to soft rehabilitation devices because force generation and mechanical compliance can be combined within the same deformable structure. Pneumatic artificial muscles generally produce axial contraction, while bending, bellows, pouch, and textile-chamber actuators generate directed deformation or rotational motion. Their behavior is shaped by chamber geometry, reinforcement patterns, textile arrangement, actuator loading, and the configuration of the attachment points [12–15]. This design flexibility allows pneumatic actuators to be lightweight and to conform to the user's anatomy without necessarily relying on a rigid joint axis. However, actuator routing, anchoring geometry, and limb configuration must therefore be considered together when designing a pneumatic elbow rehabilitation device [16].

Previous reviews have addressed soft rehabilitation robots, wearable exosuits, and individual soft-actuator technologies from broader perspectives. In particular, Kalita et al. [17] reviewed the development of pneumatic artificial muscles, including their operating principles, manufacturing methods, force models, and applications in robotics, medicine, and rehabilitation. However, that actuator-centered perspective does not specifically examine how different pneumatic actuator architectures interact with force-transmission topology, the body-worn interface, sensing modalities, movement-intention detection, and multilevel closed-loop control in elbow rehabilitation. The present review therefore adopts a system-level perspective in which actuator architecture, mechanical transmission, sensing, and control are analyzed as interconnected subsystems.

The paper is structured as follows. Section 3 places pneumatic soft actuation within the broader landscape of soft and compliant elbow rehabilitation devices. Section 4 reviews the structural, material, and operating architectures of pneumatic soft elbow actuators. Section 5 examines the sensing modalities, intention-detection methods, and closed-loop control architectures used in these systems. Together, these sections provide an integrated view of how actuator design, force transmission, sensing, and control interact in pneumatic soft elbow rehabilitation.

2. Materials and Methods

This study is a scoping review aimed at mapping the scientific literature on actuation, structural design, sensing, intention detection, and control in soft and compliant elbow rehabilitation devices. Particular attention was given to pneumatic soft actuators and to the functional relationships between actuator design, human–robot interfacing, sensing, and closed-loop control. The methodology followed the Preferred Reporting Items for Systematic Reviews and Meta-Analyses Extension for Scoping Reviews (PRISMA-ScR) guidelines [18].

2.1. Search Strategy

The systematic literature search was conducted in three major scientific databases: Web of Science, Scopus, and IEEE Xplore. These databases provide broad coverage of engineering, robotics, rehabilitation technology, biomedical engineering, control systems, and materials science. Separate but complementary searches were conducted for the three research questions so that the broader actuation landscape, the specific architectures of pneumatic soft actuators, and their sensing and control methods could be investigated at an appropriate level of detail.

2.2. Study Selection Process and Eligibility Criteria

The selection process was carried out in multiple stages. Records retrieved from Web of Science, Scopus, and IEEE Xplore were imported into Zotero (version 9.0.6.), where duplicate records were identified and removed before screening. The remaining records were screened sequentially at the title, abstract, and full-text levels using predefined eligibility criteria. Peer-reviewed journal articles and full-length conference papers published in English between 2015 and 2026 were included in the screened corpus; standards and pre-2015 seminal works were treated as contextual sources and were not screened. The final search update was conducted on 31 July 2026. No formal review protocol was prospectively registered. The review methodology and reporting were structured according to the PRISMA Extension for Scoping Reviews (PRISMA-ScR).

The title, abstract, and full-text screening was performed by one reviewer. To reduce the risk of premature exclusion, records with insufficient information or uncertain eligibility were retained for the subsequent screening stage. Because independent duplicate screening was not performed, inter-rater agreement was not calculated.

Studies were excluded if they were non-peer-reviewed (such as conference abstracts or book reviews), were outside the defined publication timeframe, or did not have clear relevance to soft or compliant actuation in elbow rehabilitation. Studies without experimental validation were also excluded because the review aimed to map actuator, sensing, and control architectures that had been physically implemented and evaluated. Purely conceptual, analytical, or simulation-only studies were therefore not included in the evidence corpus. After applying these criteria, studies with clear and explicit relevance to the corresponding research questions were retained for full-text analysis and data charting.

This study is structured around three core research questions (RQ), which are illustrated in Figure 1. These domains, which also define the main challenges, are addressed sequentially in this paper:

- RQ1: What are the main actuation technologies used in soft and compliant elbow rehabilitation devices, and what position do pneumatic soft actuators occupy among them?
- RQ2: What structural, material, and operating architectures are used in pneumatic soft actuators for elbow rehabilitation?

- RQ3: What sensing, movement-intention identification, and closed-loop control architectures are used in pneumatic soft elbow rehabilitation systems, and how do these architectures address nonlinearities and uncertainties arising from the actuator?

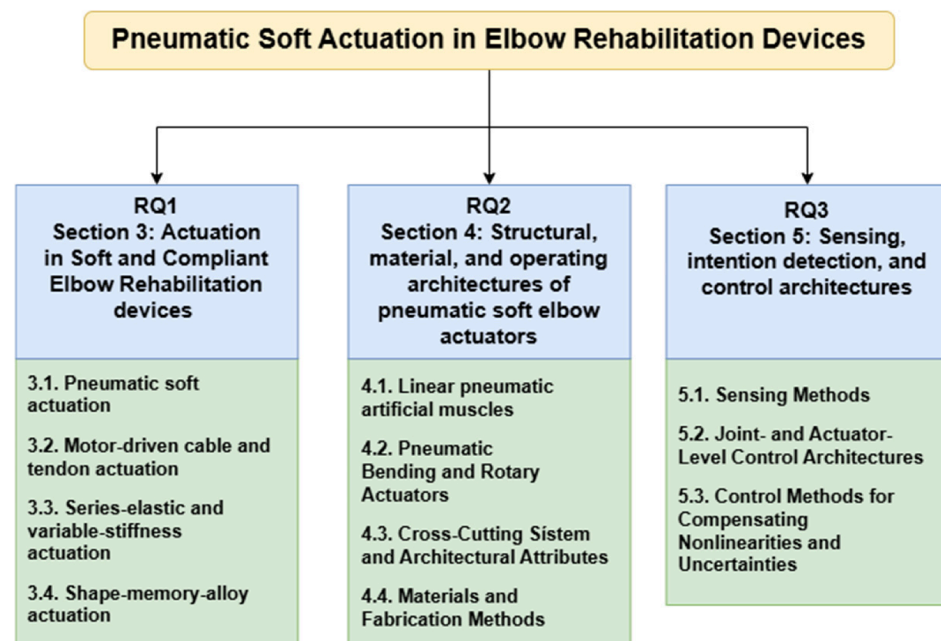


Figure 1. Paper roadmap: overview of research questions and thematic sections.

RQ1 intentionally adopts a broader actuation scope to establish the technological context in which pneumatic soft actuation is positioned. RQ2 and RQ3 then narrow the analysis specifically to pneumatic soft elbow rehabilitation systems.

The search strategy was based on Boolean combinations of keywords, as shown in Table 1.

Table 1. Search queries and databases are linked to each research question.

Research Question (RQ)	Keywords/Queries	Database
RQ1	(elbow OR “elbow joint”) AND (rehabilitation OR therapy OR assistive OR assistance) AND (exoskeleton OR exosuit OR orthosis OR “wearable robot” OR “robotic sleeve”) AND (actuator OR actuation OR pneumatic OR McKibben OR cable OR tendon OR bowden OR compliant OR “series elastic” OR “shape memory alloy” OR hydraulic OR vacuum)	Web of Science Scopus IEEE Xplore
RQ2	(elbow OR “elbow joint”) AND (“soft pneumatic actuator” OR “pneumatic soft actuator” OR “pneumatic artificial muscle” OR McKibben OR “fabric inflatable actuator” OR “textile pneumatic actuator” OR “pneumatic bending actuator” OR “pneumatic rotary actuator” OR “pneumatic bellows” OR “corrugated diaphragm” OR “vacuum actuator”) AND (design OR architecture OR geometry OR material OR fabrication OR manufacturing OR characterization OR performance)	Web of Science Scopus IEEE Xplore

Table 1. Cont.

Research Question (RQ)	Keywords/Queries	Database
RQ3	(elbow OR “elbow joint”) AND (“soft pneumatic actuator” OR “pneumatic soft actuator” OR “pneumatic artificial muscle” OR McKibben OR “fabric inflatable actuator” OR “textile pneumatic actuator” OR “pneumatic bending actuator” OR “pneumatic rotary actuator” OR “pneumatic bellows” OR “corrugated diaphragm” OR “vacuum actuator”) AND (“pressure control” OR “position control” OR “angle control” OR “force control” OR “torque control” OR impedance OR admittance OR “assist as needed” OR “adaptive control” OR “sliding mode” OR EMG OR sEMG OR IMU OR “movement intention” OR “intention detection”)	Web of Science Scopus IEEE Xplore

The database searches identified 1403 records. Following the removal of duplicates and records excluded on the basis of publication year, language, or peer-review status, the remaining records underwent title and abstract screening. Full-text reports were then assessed against the predefined eligibility criteria. The principal reasons for full-text exclusion were an actuation technology or device outside the scope of the review, the absence of experimental validation, and insufficient technical or performance-related information. The final corpus comprised 109 included reports. The complete selection process is presented in Figure 2.

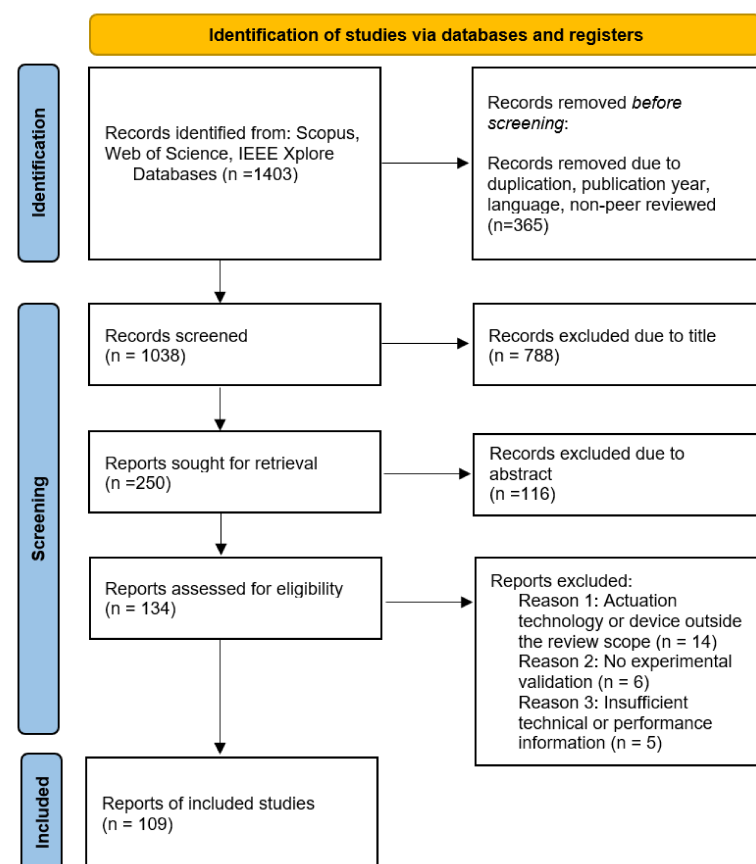


Figure 2. Identification of studies via databases.

Study screening was performed sequentially at the title, abstract, and full-text levels using predefined eligibility criteria. At each stage, records were evaluated for relevance to the corresponding research question. For the included reports, data were charted using a structured extraction framework developed according to the three research questions. For RQ1, the extracted variables included actuation technology, source of mechanical compliance, and force-transmission principle. For RQ2, actuator type, deformation mechanism, structural architecture, material, fabrication method, operating direction, and transmission configuration were recorded. For RQ3, the extracted variables included sensing modality, measured variable, intention-detection method, controlled variable, control architecture, and method used to address actuator nonlinearities or uncertainties. The extracted information was subsequently grouped into recurring technological and functional categories for qualitative synthesis. Figure 3 therefore illustrates the annual distribution of the literature across RQ1, RQ2, and RQ3 and highlights changes in the relative research emphasis over the review period.

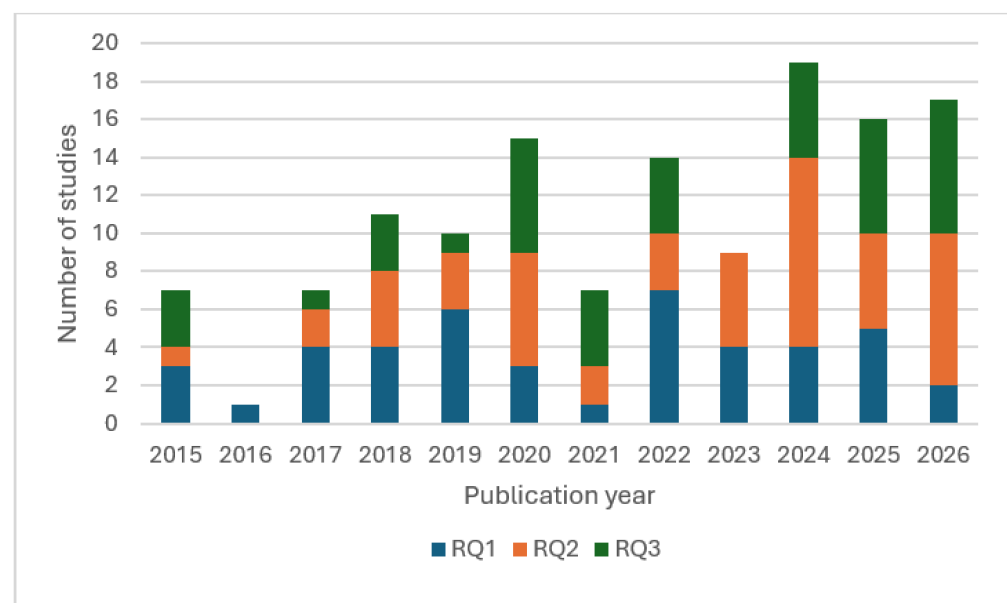


Figure 3. Temporal distribution of included studies across the research questions.

Studies without experimental validation were excluded because the review aimed to characterize actuator, sensing, and control architectures that had been physically implemented and evaluated. Purely conceptual designs, simulations, and analytical models without experimental verification were therefore considered insufficient for assessing the practical implementation of pneumatic soft actuation in elbow rehabilitation.

In this review, “rehabilitation device” is used as an umbrella term for systems developed for therapeutic exercise, physical assistance, or rehabilitation-oriented human–robot interaction. Assistive systems primarily augment task performance, whereas therapy-oriented robots are intended to support repeated rehabilitation exercises or motor training. Because several prototypes serve both purposes, the categories are not always mutually exclusive.

The review aimed to achieve broad and systematic coverage of the literature relevant to the predefined research questions, while acknowledging that database selection, indexing practices, and language restrictions may have resulted in the omission of some eligible studies. The final corpus of 109 peer-reviewed studies covered the broader actuation landscape, the structural and material architectures of pneumatic soft actuators, and the sensing, intention-detection, and control strategies used in these systems. Although some

relevant studies may not have been captured, the size and diversity of the corpus were considered adequate to identify the main design approaches, recurring system-level challenges, research gaps, and emerging trends in actuator, interface, sensing, and control integration.

The final evidence corpus comprised 109 included peer-reviewed reports. Review articles and methodological references used for contextualization, including Kalita et al. [17] and the PRISMA-ScR guideline [18], were not included in the screened evidence corpus.

3. Actuation in Soft and Compliant Elbow Rehabilitation Devices

The actuation architectures of active elbow rehabilitation devices cannot be classified solely according to whether the overall device is described as “soft” or “rigid.” In soft wearable systems, deformability and compliance may be provided by different components, including the energy-converting actuator, the force-transmission system, the body-worn interface, or an intentionally integrated elastic element. Therefore, a textile-based exosuit does not necessarily use a soft actuator. Mechanical power may instead be generated by a rigid electric motor and transmitted to the elbow through a Bowden cable or an artificial tendon.

Previous reviews of soft exosuits have accordingly distinguished between the actuation mechanism and the physical human–robot interface. In the taxonomy proposed by Xiloyannis et al. [1], the soft nature of a device mainly refers to the absence of a rigid kinematic chain and the use of deformable transmission and attachment structures. Similarly, Hussain and Ficuciello [2] identify the actuator and the body-connected interface as the two main functional subsystems of a soft exosuit.

The reviewed devices can be grouped into four main actuation technology families: pneumatic soft actuators, motor-driven cable and tendon systems, series-elastic or variable-stiffness electromechanical actuators, and shape-memory-alloy-based actuators. These categories are compared according to the mode of energy conversion, the source of mechanical compliance, and the force-transmission architecture.

3.1. Pneumatic Soft Actuation

Fluidic actuators convert the energy of pressurized gas into mechanical deformation. Within this technology family, pneumatic actuation is the most common approach in soft elbow rehabilitation devices: compressed air produces contraction, bending, or rotational motion in a deformable chamber or artificial muscle. A key system-level feature of pneumatic solutions is that energy conversion and mechanical compliance are usually provided by the same deformable component. This distinguishes them from motor-driven cable systems, for example, in which the soft characteristics are mainly associated with the transmission and the body-worn interface.

Two basic force-transmission approaches can be identified in the literature. Direct pneumatic bending actuators are positioned around the elbow and generate joint motion through their own deformation, eliminating the need for a separate rigid joint. This group includes bending joints formed from axis-free elastic chambers, pneumatic pads integrated around the elbow, and textile-based actuators [19–21]. These designs can passively adapt to the user’s anatomy, although their mechanical output depends strongly on chamber geometry, operating pressure, actuator deformation, and the current elbow angle.

The second main approach is based on contractile pneumatic artificial muscles. These actuators shorten under pressure, while attachment points, straps, cables, or external brackets convert the resulting tensile force into joint torque [22]. The reviewed systems included both conventional pneumatic artificial muscles and artificial muscles integrated into textile or garment-based structures [23–26]. The functional sequence from pneumatic pressurization to assisted elbow motion is presented in Figure 4.

Other variants within the pneumatic actuator family include bellows-, diaphragm-, origami-, and textile-based inflatable structures. Folded or corrugated geometries can enable directed deformation and a larger range of motion, while modular chambers allow actuator length and performance to be adapted to a specific application [27–30]. Textile-chamber systems can provide lightweight actuators that are suitable for integration into garments. Their performance, however, is determined not only by chamber design, but also by the surrounding textile layers, attachment points, and force-distribution elements [31–33].

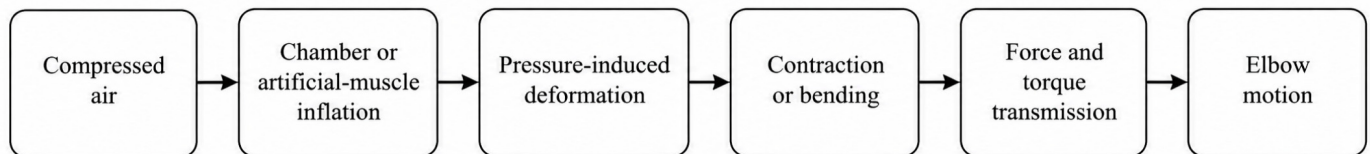


Figure 4. Operating principle of pneumatic soft actuation for elbow assistance.

Some designs incorporate local stiffening or guiding elements into the deformable pneumatic structure. Rigid plates placed between chambers, fibre or fabric reinforcement, and semi-rigid interfaces can reduce unwanted deformation, increase useful force output, and improve torque transmission without turning the entire device into a rigid exoskeleton [34,35]. More recent modular systems connect multiple pneumatic units to support the shoulder, elbow, and wrist, demonstrating the potential of this technology for multi-joint applications [34,36].

The main advantages of pneumatic actuation are low local actuator mass, intrinsic compliance, passive adaptation to body geometry, and the possibility of designing systems without a rigid joint axis. Its main limitations include air compressibility, hysteresis in elastomeric and textile materials, load-dependent deformation, and the need for a supporting system consisting of pumps, valves, tubing, and a power source. Since pneumatic actuators differ considerably in terms of materials, chamber geometry, fabrication, force transmission, and attachment, these aspects are examined in greater detail in Section 4.

3.2. Motor-Driven Cable and Tendon Actuation

In motor-driven cable and tendon systems, mechanical energy is typically generated by a DC motor or servomotor, meaning that the energy-converting element itself is not intrinsically soft. The soft or compliant nature of the system mainly arises from the flexible force-transmission path, the textile-based human–robot interface, deformable attachment structures, or the absence of a rigid kinematic chain. This represents an important difference from pneumatic actuators, in which energy conversion and mechanical compliance are usually provided by the same deformable component.

The mechanical performance of cable-driven systems is not determined solely by motor torque. The useful joint torque also depends on cable-routing geometry, the moment arm, sheath stiffness, and the stability of the attachment points connected to the body. Deformation and friction within the cable system may reduce the transmitted force, while displacement or stretching of the textile interface may further increase uncertainty in position and force transmission. The transmission and the human–robot interface therefore cannot be treated as independent subsystems. The operating principle of a motor-driven cable and tendon actuation is presented in Figure 5.

One of the main design advantages of cable-driven architectures is that the actuators can be positioned remotely. Motors may be placed on the torso, back, waist, or other body regions where their inertial effect is lower, while Bowden cables or artificial tendons transmit force to the elbow. This can reduce the mass carried on the upper arm and, in particular, on the forearm, thereby decreasing the distal inertia that resists limb movement. However,

remote actuation requires a longer and mechanically more complex force-transmission path, in which friction, backlash, cable elongation, sheath compression, and losses related to path curvature may occur [6,7,37].

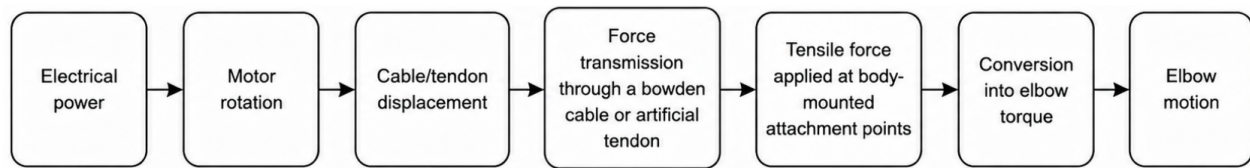


Figure 5. Operating sequence of motor-driven cable and tendon actuation for elbow assistance.

Several cable arrangements have been used to generate elbow flexion and extension. Unidirectional assistive systems use a single tendon or cable, while movement in the opposite direction is produced by the user, gravity, or a passive elastic element. Bidirectional systems use antagonistic cable pairs or two separate tendons, allowing active control of both flexion and extension [38–41].

The design of Bowden cables has a particularly strong influence on system efficiency. An appropriate configuration can improve force transmission and position tracking, although a stiffer cable may provide less geometric adaptability. Combining semi-rigid and textile components at the attachment points may reduce local contact loads while providing more stable force application than fully flexible textile attachments [42].

Twisted-string actuators represent a specialized form of this technology. In these systems, a motor twists parallel bundles of fibres, and their geometric shortening produces tendon tension. The mechanism can provide a high mechanical transmission ratio with a lightweight and compact motor, but the relationship between string rotation and contraction is strongly nonlinear. Cyclic twisting, friction between the fibres, and fibre elongation may also lead to wear, hysteresis, and repeatability problems [43].

Another development direction in cable-driven systems is to reduce the number of motors by operating multiple degrees of freedom from a shared power source. Mechanical or electromagnetic clutches can distribute the output of a single motor over time between different cable branches and joints. This may reduce the total motor mass and the complexity of the power supply, although it increases control complexity [44].

3.3. Series-Elastic and Variable-Stiffness Actuation

Series-elastic and variable-stiffness actuators are grouped together here because both introduce mechanical compliance through a deliberately designed element located between the motor and the joint output. However, SEA architectures primarily use fixed series elasticity for force transmission and torque estimation, whereas VSA architectures allow the effective mechanical stiffness to be actively adjusted.

In series-elastic actuators (SEAs), a deformable mechanical element is placed between the motor and the output of the device. Although energy conversion is typically performed by an electric motor, the series elasticity reduces the output mechanical impedance, partially decouples the user from the stiffness of the motor and transmission, and attenuates force peaks caused by sudden changes in loading. By measuring deformation of the elastic element, the output force or torque can be estimated indirectly. SEA architectures are therefore particularly suitable for force-, torque-, and impedance-controlled human–robot interaction.

The design of SEA systems can vary considerably depending on the type of transmission used and the position of the elastic element. When combined with cable or Bowden transmission, the motor can be placed away from the limb, while the series-elastic element provides compliance within the cable path or at the output connection. In other

systems, a compact rotary SEA generates the required torque directly near the joint. In multi-degree-of-freedom mechanisms, series-elastic actuators can be coupled so that elbow flexion extension and forearm pronation and supination can be controlled with partial independence [8,45,46].

In multi-joint rehabilitation exoskeletons, SEAs are generally combined with semi-rigid kinematic chains, self-aligning passive degrees of freedom, and mechanisms that compensate for misalignment with anatomical joint axes. In this architecture, the soft or compliant character of the device does not arise from deformation of the entire structure, but from elasticity intentionally incorporated into the drive system. This may provide more accurate joint kinematics and more controllable torque transmission than fully textile-based systems, although it may also involve greater structural mass, a more rigid human–robot interface, and more complex joint-axis alignment [9,47,48].

In variable-stiffness actuators (VSAs), the output position and effective mechanical stiffness can be adjusted partly independently. One important advantage of VSA systems is that their mechanical behavior is not modified solely through the parameters of the feedback controller, but can also be changed physically. Higher stiffness may increase resistance to external loads and improve position holding, while lower stiffness may reduce force peaks during human–robot interaction. However, this functionality may require additional mechanical components, actuators, or adjustment mechanisms, increasing system mass, size, and control complexity [49,50].

Variable stiffness is not limited to conventional spring-motor systems. In some soft robotic joints, the deformable structure itself acts as the elastic element, and its geometry or pressure state determines the effective stiffness. Origami-inspired soft joints therefore form a transition between classical mechanical VSAs and intrinsically compliant pneumatic actuators, as their stiffness can be actively adjusted [51].

3.4. Shape-Memory-Alloy Actuation

Shape-memory-alloy (SMA) actuators generate mechanical deformation through a thermally induced solid-state phase transformation. In practice, an SMA wire or coil spring contracts when heated by an electric current and returns to its initial state during cooling. Because operation does not require a motor, gearbox, pneumatic tubing, or compressor, SMA actuation can enable a compact and lightweight mechanical architecture. A frequently reported advantage of SMA actuation is its high force-to-volume potential. Thin wires can be integrated into lightweight, additively manufactured, or textile-connected structures, while their operation is quiet and requires few moving mechanical components. The operating principle of SMA actuation is presented in Figure 6.

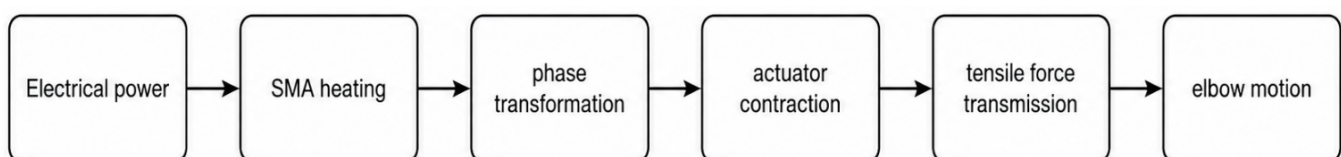


Figure 6. Operating sequence of SMA actuation for elbow assistance.

In elbow rehabilitation devices, SMA elements typically function as cable- or tendon-like tensile actuators. The linear contraction generated by wires or spring bundles is converted into joint rotation through attachment points and mechanical moment arms. By controlling multiple bundles separately, more complex upper-limb movements can be supported in addition to flexion and extension [10,52].

However, actuator dynamics are fundamentally determined by heat transfer. Heating can be accelerated through electrical control, whereas cooling may be substantially slower

under natural convection. Active cooling methods can therefore be used to improve heat dissipation. Integrating SMA coil springs into a flexible fluid reservoir allows the coolant to accelerate relaxation and reduce the thermal delay between successive cycles [11]. However, active cooling may require an additional pump, tubing, fluid reservoir, and control system, partly offsetting the advantages associated with the simple and compact architecture of SMA actuation.

Thermal limitations may also be reduced by operating multiple actuator bundles in a time-shifted sequence. During sequential activation, individual SMA bundles generate force alternately while the inactive elements are allowed to cool. This may reduce the thermal load on each actuator, improve the ability to operate continuously, and provide more accurate dynamic tracking than designs using a single continuously loaded bundle [53].

From a safety perspective, the high operating temperature of SMA actuators requires particular attention. Wires or springs must be placed within a thermally insulating enclosure, positioned at a suitable distance from the skin, or integrated into a liquid-cooled structure. Limiting the temperature may, however, reduce the available contraction and force. Thermal safety is therefore directly linked to the mechanical performance of the actuator.

3.5. Comparative Synthesis of Actuation Technologies

The technologies used in active soft and compliant elbow rehabilitation devices differ primarily in terms of the subsystem in which energy conversion, mechanical compliance, and force transmission are implemented. In pneumatic soft actuators, these functions are largely integrated within the same deformable chamber or artificial muscle. In motor-driven cable and tendon systems, energy conversion is performed by a rigid electric motor, while the soft characteristics arise from the flexible transmission and textile-based human–robot interface. In SEA and VSA systems, compliance is introduced through a separate, intentionally designed mechanical element, whereas SMA actuators rely on a thermally induced material transformation.

The main characteristics of the key technologies and the studies describing them are presented in Table 2.

Table 2. Comparison of the main actuation families used in soft elbow rehabilitation devices.

Actuation Family	Energy-Conversion Element	Main Limitation	Main Advantage	Representative Studies
Pneumatic soft actuation	Pressure-deformable chamber or artificial muscle	Nonlinear behavior and need for a pneumatic supply system	Low local mass and intrinsic compliance	[14,19,20,22,23]
Motor-driven cable/tendon actuation	Electric motor	Friction, backlash, and attachment-related losses	Remote motor placement and bidirectional actuation	[6,7,37,42–44]
Series-elastic actuation (SEA)	Electric motor and series elastic element	Greater mass and mechanical complexity	Indirect torque estimation and low output impedance	[8,9,45,46,48]
Variable-stiffness actuation (VSA)	Motor-driven stiffness-adjustment mechanism	Complex mechanics and control	Stiffness adaptable to the patient and task	[49–51]
Shape-memory-alloy actuation (SMA)	Thermally activated alloy	Slow cooling and hysteresis	Compact size, low mass, and quiet operation	[10,11,52,53]

Pneumatic soft actuators are distinguished most clearly from alternative technologies by their ability to combine mechanical power generation and intrinsic compliance within the same deformable structure. This integration can facilitate low-distal-mass, geometrically adaptable configurations and may enable designs that do not require a rigid joint axis. At the same time, it results in nonlinear actuator behavior, challenges related to pressure and flow regulation, and the need for a supporting system consisting of pumps, valves, and pneumatic tubing. This specific interaction among the actuator, wearable interface, and pneumatic supply system motivates the more detailed examination of pneumatic soft systems in the following sections.

4. Structural, Material, and Operating Architectures of Pneumatic Soft Elbow Actuators

Pneumatic soft actuators for elbow rehabilitation do not form a single homogeneous technological category. Although all of these systems convert a pressure difference into mechanical work, they differ considerably in their deformation mechanisms, actuator materials, chamber geometries, force-transmission methods, operating directions, and attachment to the body. The set diagram of soft actuators used in elbow rehabilitation is shown in Figure 7.

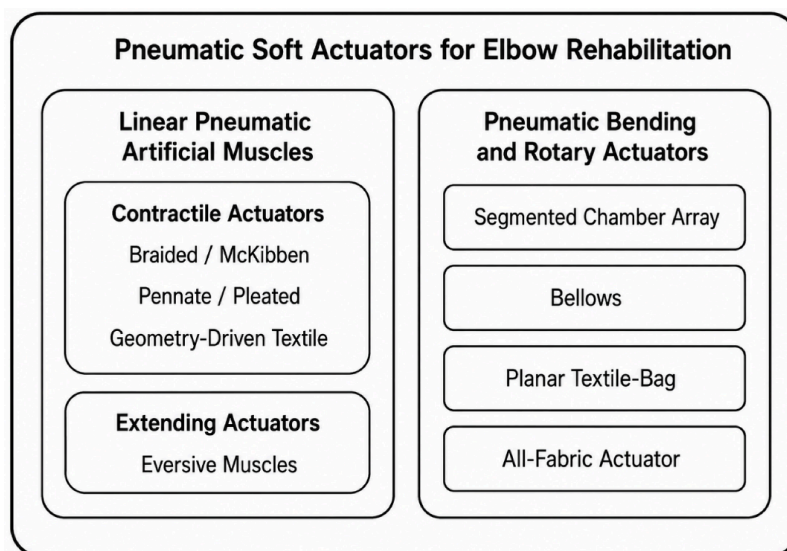


Figure 7. Classification of pneumatic soft actuators for elbow rehabilitation.

For the purposes of this review, the pneumatic architectures are described using two complementary levels of classification. The primary classification is based on the actuator's mechanical output mechanism and distinguishes linear pneumatic artificial muscles from pneumatic bending and rotary actuators. In addition, several cross-cutting architectural attributes are considered, including rigid–soft integration, cable-, tendon-, or flexure-mediated transmission, modularity and reconfigurability, antagonistic or bidirectional configuration, and variable-stiffness or multifunctional integration. These attributes are not mutually exclusive actuator classes and may occur in combination within the same device.

4.1. Linear Pneumatic Artificial Muscles

Linear pneumatic artificial muscles generate motion primarily through pressure-induced changes in actuator length. Their mechanical output is then converted into elbow torque through attachment geometry, external moment arms, or an additional transmission structure. Although these actuators share a common linear operating principle, their contraction mechanisms differ considerably. The following subsections distinguish braided

and McKibben-type muscles, pennate and pleated configurations, geometry-driven textile muscles, and eversive actuators according to how internal pressure is transformed into axial force and displacement.

4.1.1. Braided and McKibben-Type Muscles

Conventional pneumatic artificial muscles (PAMs) consist of a deformable inner bladder surrounded by a braided or mesh sleeve. An increase in internal pressure produces radial expansion, which is converted by the braid into axial shortening and tensile force. This mechanism provides linear contraction similar to that of a biological muscle, which is then converted into elbow torque through the attachment geometry or a mechanical transmission.

Punjabi et al. [54] used a cast liquid-silicone inner membrane and a polyethylene terephthalate braided sleeve in an actuator designed to assist elbow flexion. The actuator was integrated into the exoskeleton as a series of multiple PAMs, with pressure-dependent contraction and stiffness forming the basis of its operation. A similar braided architecture was used in a rehabilitation device assisting upper- and lower-limb flexion and extension, in which the actuator consisted of an elastomeric bladder and an external threaded mesh sleeve [55]. Wu et al. [56] developed a braided pneumatic artificial muscle with programmable body morphology, whose deformation and maximum contraction could be modified through the geometric design of the braided structure.

Abe et al. transformed a thin, textile-integrable version of the McKibben principle into an active fabric. In this design, thin McKibben muscles were incorporated into both the warp and weft directions of the textile, so actuation occurred within the wearable fabric structure rather than in separate muscle bundles [57]. In contrast, Jiang et al. [58] arranged pneumatic muscles according to the anatomical distribution of the human elbow flexor and extensor muscles. The muscle configuration was selected through comparisons of material structures and force modelling.

In multi-degree-of-freedom PAM systems, the linear muscle force is converted into complex motion by a mechanical elbow structure. Yang et al. used PAMs in a three-degree-of-freedom bionic elbow joint and described the mechanism using a pseudo-rigid-body model [59]. These architectures are mechanically closer to pneumatically driven exoskeletons because the soft actuators operate together with a more rigid joint mechanism.

4.1.2. Pennate and Pleated Artificial Muscles

Artificial muscles based on biological pennation or pleated geometry provide an alternative to conventional axial PAM configurations. The purpose of pennation is to combine multiple muscle elements within a low-profile structure to generate greater effective force or elbow torque.

Xie et al. [3] combined the attachment pattern of the brachioradialis muscle with a pennate artificial-muscle architecture. The changing geometric relationship between the pneumatic muscles and the surface of the forearm was described using force-equilibrium and virtual-work models. The resulting device was a low-profile elbow exosuit fitted directly to the arm, in which muscle placement and routing around the body determined the output torque.

Laterally pleated PAMs, in contrast, use side pleats and tensile straps. Inflation of the pleats is converted by the straps into axial contraction, while actuator expansion is directed mainly toward one side. This may reduce radial loading on the wearer's body and limit interference between actuators positioned next to one another [60].

4.1.3. Geometry-Driven Textile Pneumatic Muscles

In a newer generation of textile-based pneumatic muscles, contraction is generated by a programmed geometric textile structure rather than a braided sleeve. The fabric-lattice artificial muscle arranges active air chambers and passive connecting layers in a zigzag lattice. The lattice geometry can convert in-plane pneumatic expansion into linear, radial, or curved contraction, allowing the same basic principle to be configured for the elbow, shoulder, and other applications [61].

The X-crossing PAM uses the relative movement of intersecting deformable elements to produce axial shortening. This differs substantially from braided McKibben muscles because contraction arises directly from the geometric rearrangement of the crossing mechanism. The design can be configured in parallel, asymmetric, and ring-shaped arrangements, providing considerable design flexibility for elbow actuation [62].

Pouch-type artificial muscles use the deformation of welded or woven air pouches to generate contraction. In the asymmetrically expanding pouch actuator developed by Lu et al., the centre of pouch expansion was offset from the tensile axis. As a result, most of the unwanted expansion was directed away from the body, reducing the compressive force applied to the wearer. An important engineering objective of the design was therefore not only to increase contraction, but also to reduce radial pressure on the body-facing side of the actuator [63].

4.1.4. Eversive Muscles

The operating principle of the eversive pneumatic artificial muscle (EPAM) is the opposite of that of a conventional PAM. An EPAM extends under pressure, while the eversion mechanism generates tensile force. In agonistic or antagonistic arrangements, the extending actuator can also drive a rotary joint. This architecture is particularly notable because increasing pressure produces controlled elongation rather than contraction [64].

4.2. Pneumatic Bending and Rotary Actuators

Pneumatic bending and rotary actuators generate angular displacement directly through asymmetric chamber expansion, geometric interaction between cells, or constrained deformation. This motion can be converted into joint rotation either directly or through a short, integrated mechanical transmission. Unlike contractile pneumatic artificial muscles, these architectures do not necessarily require a separate cable transmission or external moment arm because the deformation of the actuator itself follows or produces the joint motion.

4.2.1. Segmented Chamber-Array and Bellows Bending Actuators

In fabric-covered and bellows-type bending actuators, expansion of the pneumatic chambers is directed by inextensible textile layers, folds, cell walls, or geometric contact between adjacent elements. Because expansion is constrained, the actuator produces bending or rotational motion rather than straight-line contraction. The mechanical output is jointly determined by the number, shape, and dimensions of the chambers, the geometry of the folds, the actuator's radius of curvature, and the attachment points on the body.

Thalman et al. [12] arranged several fabric-covered pneumatic soft actuators in a serial array to support elbow flexion. Inflation of the sequentially positioned chambers produced cumulative bending and elbow torque, while the textile covering limited unwanted radial expansion and defined the direction of deformation.

Bellows architectures use folded or cellular structures in which individual segments open under pressure, producing curved motion through their combined geometric change. Varying the size, shape, and arrangement of the cells modifies the relationship among

pressure, bending angle, and output force. Zhang et al. [65] investigated bioinspired bellows actuators for shoulder and elbow applications and showed that structural geometry directly determines the bending and loading characteristics of the actuator.

A similar geometry-optimization approach has been applied to actuators composed of spherical pneumatic chambers. Ma et al. [66] investigated the relationship between bending range and force output by modifying chamber shape and dimensions in an upper-limb exoskeleton incorporating soft pneumatic actuators. In these actuators, contact between adjacent cells and deformation of the individual chambers jointly determine the overall actuator curvature.

The performance of bellows actuators is particularly sensitive to their positioning relative to the body. Sahin et al. compared textile bellows actuators with different cell shapes, dimensions, and numbers using a physical model representing infant anthropometry. The variants were evaluated according to motion smoothness, actuator path length, and the resulting elbow angle. Their findings indicated that actuator dimensions cannot be selected independently of limb size and attachment topology [67].

Ayazi et al. [16,68] examined the force output and attachment points of shoulder and elbow actuators, as well as the influence of the configuration of the adjacent joint. The same bellows actuator generated different forces and ranges of motion depending on its attachment location and the shoulder angle. In a later data-driven study, four shoulder and elbow actuators—including elbow bellows composed of square and circular cells—were compared under different pressure, joint-angle, and attachment configurations. These findings indicate that bellows geometry and the human–robot interface should be treated as a single coupled mechanical system.

4.2.2. Planar Textile-Bag and All-Fabric Actuators

Textile-bag actuators are made from two or more coated fabric layers joined to form an airtight chamber. During inflation, the thickness and shape of the chamber change, while the cutting pattern, seams, welding lines, and inextensible layers determine the direction of deformation. Textile geometry therefore serves not merely as an enclosure but as a functional part of the actuator kinematics.

Shen et al. [15] used a modular pneumatic textile bag made from woven nylon with a TPU coating to support elbow flexion and extension. The textile chamber was combined with 3D-printed connecting elements and a cable transmission. The achievable joint motion was jointly determined by deformation of the textile bag, actuator placement, and the effective length of the transmission cable. The system therefore represents a transition between a direct bending actuator and cable-transmitted pneumatic actuation.

In all-fabric architectures, the chambers, deformation-limiting layers, and force-transmission structures can all be made from textiles. Ma et al. [69] developed a bidirectional pneumatic actuator that used the same basic structure to support flexion and extension of the elbow, wrist, and fingers. Bidirectional operation may reduce the need for the return motion to be produced by gravity, a passive spring, or a separate antagonistic actuator.

Textile-bag and all-fabric actuators provide lightweight and geometrically adaptable solutions, but their performance is sensitive to textile stretching, seam and weld stiffness, relative movement between layers, and attachment to the body. The textile pattern and interface design therefore directly determine the useful bending, force output, and repeatability of the actuator.

4.3. Cross-Cutting System and Architectural Attributes

4.3.1. Rigid–Soft Integrated Architectures

One limitation of pneumatic actuators is that high compliance may result in undesirable deformation, reduced structural stability, or less accurate force direction. To address this issue, several systems combine soft actuators with partially rigid or semi-rigid frames.

The bone-muscle integrated actuator combines deformable pneumatic chambers and frame-like elements within a single 3D-printing process. Zhang et al. [70] created an integrated cavity-and-frame structure from TPU, in which the “bone” and “muscle” functions were not provided by separate components but formed part of a shared additively manufactured structure. Its modularity allowed multiple actuator units to be configured for different joints.

Oguntosin et al. [71] combined pneumatic soft modules with a 3D-printed exoskeleton containing passive and active joints. The soft material and pneumatic actuation provided compliance, while the anatomically aligned exoskeleton enabled targeted transmission of force and torque.

The exoneuromusculoskeleton developed by Nam et al. [72] integrated a soft pneumatic muscle, a mechanical exoskeleton, and neuromuscular electrical stimulation. The mechanical structure therefore served not only to attach the actuator but also as part of a multifunctional rehabilitation system supporting sequential movement of the elbow, wrist, and fingers.

A turtle-shell-inspired exosuit combined deformable pneumatic elements with a shell-like force-distribution geometry. The purpose of the bioinspired structure was to convert the force generated by the pneumatic actuators into joint torque more effectively and to provide greater output within a compact structure [73].

Dimonte et al. [74] combined antagonistic TPU pneumatic muscles with a rigid, single-degree-of-freedom frame. In this hybrid joint, the rigid structure defined the kinematic motion, while the antagonistic soft actuators provided adjustable position and mechanical compliance.

4.3.2. Cable-, Tendon-, and Flexure-Transmitted Pneumatic Actuation

Although pneumatic soft actuators can be positioned directly around the elbow, several systems use indirect force transmission. The linear contraction of a PAM is converted into joint rotation by a cable, steel wire, tendon, or flexure structure.

In a four-degree-of-freedom upper-limb exoskeleton, pneumatic muscles actuated the shoulder and elbow joints through steel cables. The PAMs could be positioned away from the joints, while the cables transmitted mechanical force to the appropriate joint axes [75]. In the dynamic and control-oriented version of the same system, the cable transmission was modelled using a Lagrangian formulation and the principle of virtual work [76].

Zheng et al. [77] used a pneumatic tendon actuator in the elbow module, while a separate fibre-wound flexible actuator was developed for the wrist. This joint-specific selection of actuators demonstrates that not every degree of freedom in a multi-joint rehabilitation robot needs to be driven by the same pneumatic structure.

Tschiersky et al. [13] combined thin McKibben muscles with a flexure-strip structure. The flexible strip converted the small-displacement, high-force muscle contraction into a larger rotational range of motion while reducing concentrated loads at the attachment points. This architecture represents a transition between direct soft actuation and rigid mechanical transmission.

Beck et al. [78] mounted the forearm anchor point of a PAM on a guide rail, allowing a more favourable moment arm to be maintained during elbow movement. This design

demonstrates that variable anchor-point geometry can increase and equalize joint torque transmission without modifying the pneumatic muscle itself.

4.3.3. Modular and Reconfigurable Actuator Systems

The purpose of modularity is to allow actuator length, placement, number, and direction of force transmission to be adapted to different body sizes, rehabilitation tasks, or joints.

Liu et al. [79] combined a mechanically programmable textile sleeve, an internal pneumatic bladder, and a 3D-printed self-locking end closure. The actuators could be attached to an interface composed of semi-rigid cylindrical segments, allowing the same hardware set to be configured for different joints.

A reconfigurable upper-limb exoskeleton used separate soft elbow and wrist modules that could operate either independently or together. This modular architecture allowed the system to be adapted to the patient's functional needs and the objectives of rehabilitation [80].

In the decoupled architecture developed by Greig et al. [5], the actuator and the body-worn force-transmission structure were partially separated. The design aimed to accommodate a broad range of participant body sizes and to transfer as much of the actuator-generated moment as possible to the user.

Jeong and Wang [81] developed a detachable pneumatic linear actuator that could be connected to wearable textiles using distributed temporary-bonding fasteners. These fasteners allowed the actuator to accommodate textile stretching and skin-length changes during bending, while the same actuator unit could be used in both glove-based and elbow-assistive systems.

Zhang and Krishnan [82] used a serially nested PAM architecture to amplify the approximately 17% contraction of individual muscles to a total contraction of nearly 40%. Increasing the number of serial stages did not, however, produce unlimited growth in stroke length, because connector dimensions and muscle-length-dependent operation reduced the useful displacement.

A modular multi-chamber actuator integrated self-sensing, locking, and configurable muscle length [14], while the geometric programmability of fabric-lattice actuators allowed them to be reconfigured for different joints [61]. Modularity can therefore be implemented at least three levels: the number of actuator units, actuator geometry, and placement of the human-robot interface.

4.3.4. Antagonistic and Bidirectional Actuator Configurations

Active support of both elbow flexion and extension requires either two actuators acting in opposite directions or a single pneumatic structure capable of bidirectional operation.

An antagonistic PAM configuration directly imitates the biological biceps-triceps muscle pair. Tsai et al. used paired pneumatic muscles for elbow flexion-extension and shoulder rotation, while other degrees of freedom were actuated by a single PAM [83]. The bi-muscular pneumatic servo-drive developed by Dindorf and Wos [84] was similarly based on two oppositely acting pneumatic muscles and a biological biceps-triceps model.

An antagonistic EPAM joint configuration uses pressure-induced elongation to generate rotary motion [64]. A developed device uses an antagonistic arrangement of two soft pneumatic actuators to produce greater joint torque while maintaining compliant interaction [85]. Dimonte et al. [74] used antagonistic TPU muscles to create a hybrid joint capable of adjusting both position and stiffness.

The all-fabric bidirectional actuator developed by Ma et al. [69] supported motion in opposite directions within a single textile architecture and could be adapted to the

wrist and finger joints as well as the elbow. In a system using both positive and negative pressure sources, inflation and vacuum provided separate operating states for the elbow actuator [86].

In multi-joint systems, actuators are distributed according to different anatomical functions. Proietti et al. [87] developed a soft wearable robot using textile-based actuators to support shoulder elevation and elbow extension. Patel et al. [88] used pneumatic muscles to support forearm pronation and supination in addition to elbow movement. The two-degrees-of-freedom system developed by Tripanpitak et al. [89] combined flexion-extension and pronation-supination for the rehabilitation of peripheral nerve injuries.

Modular PAM exoskeletons may also include wrist and hand modules in addition to the elbow module [72,80]. A multi-joint design increases the potential to support functional movements, but it also requires additional pneumatic channels, valves, and coordinated actuator configurations.

4.3.5. Variable-Stiffness and Functionally Integrated Architectures

A pneumatic system can generate not only motion but also controllable mechanical stiffness. Ma et al. [90] developed a vacuum-actuated, muscle-inspired fibre-jamming structure. Its stiffness was modified by changing the contact state of overlapping flexible fibres and the vacuum pressure. The structure retained a certain degree of extensibility even in its high-stiffness state and provided adjustable impedance for active elbow rehabilitation.

The jamming mechanism may be implemented as a separate actuator or integrated into an artificial muscle. The RSPAM can be locked through positive-pressure jamming, while its multi-chamber pneumatic element provides contraction and self-sensing [14].

Thompson et al. [91] combined linear and helical fibre-reinforced elastomeric actuators. The linear elements generated elbow motion, while the helical elements provided pneumatically adjustable resistance to joint movement. Together, the two actuator types provided greater functional performance than either type alone.

A bioinspired hybrid joint modified both position and stiffness by simultaneously changing the pressure in antagonistic pneumatic muscles [74].

Recent developments in soft pneumatic actuator design further indicate a shift toward multifunctional and reconfigurable architectures. Gunawardane et al. [92,93] demonstrated a fully 3D-printable pneumatic actuator architecture that can be configured to generate different deformation modes within a common structural platform. More recently, modular strain-limiting layers have been used to modify actuator stiffness and trajectory without requiring complete actuator redesign. These concepts are relevant to wearable rehabilitation because the ability to modify deformation mode, stiffness, or trajectory within a common actuator platform could facilitate adaptation to different users, joint geometries, and rehabilitation tasks.

4.4. Materials and Fabrication Methods

Materials and fabrication methods play a central role in determining actuator deformability, pressure capacity, durability, and integration with the wearable structure. The following subsections summarize the main material systems and manufacturing approaches used in pneumatic soft elbow actuators.

4.4.1. Elastomeric Bladders and Braided Sleeves

The inner membrane of a conventional PAM is typically made from an elastomer and surrounded by a braided mesh. The membrane retains the internal pressure, while the braid constrains radial deformation and determines the direction of contraction. Punjabi et al. used a silicone bladder and PET braid [54], whereas Khan et al. [55] used an elastomeric membrane with an external threaded mesh sleeve.

Because of their smaller diameter, thin McKibben muscles can be woven into textiles or combined with flexure structures [13,57]. Their advantages include a small cross-section and the ability to position them close to the body surface, although multiple muscle elements or a complex textile structure may be required to achieve the desired force.

4.4.2. Coated Textiles and Heat-Sealed Chambers

Textile actuators are often made from TPU-coated nylon or other fabrics provided with an airtight layer. Chambers can be formed by sewing, heat sealing, lamination, or adhesive bonding. Shen et al. [15] used TPU-coated woven nylon in an elbow actuator, while Liu et al. [79] combined an internal bladder with a mechanically programmable textile sleeve and a 3D-printed self-locking end closure.

The mechanical operation of fabric-lattice, pouch, and all-fabric actuators is largely determined by the textile pattern itself [61,63,94]. In these systems, the cutting pattern, seams, and placement of inextensible layers form part of the effective actuator kinematics.

4.4.3. Silicone-Cast and Elastomeric Structures

Silicone-cast chambers can provide considerable geometric freedom and relatively uniform deformation. In a paediatric device, the elbow actuator was fabricated by silicone casting, while textile technology was used for the shoulder actuator, indicating that material selection can be adapted to the functional requirements of the individual joint [95].

3D-printed TPU structures allow chambers, flexures, and load-bearing elements to be integrated into a single component. In the bone-muscle integrated actuator, both the cavities and the frame structure were made from TPU through an integrated manufacturing process [70]. In modular textile actuators, 3D printing was used mainly to produce end closures and connecting elements [79].

Dimonte et al. [74] combined pneumatic artificial muscles made from thermoplastic polyurethane with a rigid joint frame. The weldability and textile-like deformability of TPU allow thin and lightweight chambers to be produced, although the kinematics of the overall system are still partly determined by the rigid frame.

4.5. Comparative Synthesis of Pneumatic Actuator, Transmission, and Material Architectures

The structural architectures of pneumatic soft elbow actuators can be organized around two fundamental principles of mechanical output. Linear pneumatic artificial muscles convert pressure-induced radial or geometry-driven deformation into axial contraction or, less commonly, into eversive elongation. Pneumatic bending actuators, in contrast, generate joint rotation directly or through a short mechanical transmission by deforming segmented chambers, bellows elements, or patterned textile layers.

In addition to chamber geometry and operating pressure, the mechanical output of an individual actuator is strongly influenced by the force-transmission topology. Cable-, tendon-, and flexure-based mechanisms can increase the achievable range of motion or allow the actuator to be positioned farther from the joint, whereas rigid–soft integration can provide more controlled force routing and greater kinematic stability. Modular, antagonistic, and reconfigurable architectures address force, stroke-length, and bidirectional actuation requirements by modifying the number, placement, and operating direction of the actuator units. Jamming-based, self-sensing, and variable-stiffness solutions further extend actuator functionality by integrating mechanical locking, adjustable impedance, or state sensing into the actuation system.

From a materials perspective, the reviewed systems range from conventional PAMs with elastomeric bladders and braided sleeves to coated, welded, and geometrically programmed textile chambers, as well as silicone-cast and additively manufactured TPU structures. Textile-based actuators offer low mass and good geometric adaptability, whereas cast

and 3D-printed designs enable more complex chamber geometries, localized reinforcement, and the integration of load-bearing elements.

The main pneumatic soft actuator and system architectures identified in the reviewed elbow rehabilitation devices are summarized in Table 3, including their principal advantages, limitations, and representative studies.

Table 3. Comparison of the soft pneumatic architectures used in elbow rehabilitation.

Architecture	Main Advantage	Main Limitation	Representative Studies
Braided/McKibben PAM	High tensile force and a well-established design.	Limited contraction and radial expansion.	[54–58]
Pennate and pleated PAM	Low-profile design and directed expansion.	Output depends on actuator placement and strap geometry.	[3,60]
Geometry-driven textile muscles	Large contraction and low mass.	Sensitive to fabrication accuracy and geometric parameters.	[61–63]
Eversive muscle	Large change in actuator length.	Specialized attachment requirements.	[64]
Chamber-array and bellows actuators	Direct generation of joint bending.	Strong dependence on chamber geometry.	[12,16,65–68]
Textile-bag and all-fabric actuators	Lightweight, low-profile.	Affected by textile stretching and seam deformation.	[15,69]
Rigid–soft integration	Improved kinematic stability and force routing.	Increased mass and structural rigidity.	[70–74]
Cable-, tendon-, and flexure-mediated actuation	Remote actuator placement, increased range of motion.	Friction and transmission losses.	[13,75–78]
Modular and reconfigurable systems	Adaptability to anthropometry and rehabilitation tasks.	Additional connectors and increased system complexity.	[5,14,61,79–82]
Antagonistic and bidirectional configurations	Active assistance of both elbow flexion and extension.	Multiple pneumatic channels and coordinated control are required.	[69,74,83–89]
Variable-stiffness and multifunctional architectures	Adjustable impedance and integrated functionality.	Greater structural and control complexity.	[14,74,90–93]

Cross-study comparison of material performance was limited because key engineering parameters such as burst pressure, fatigue life, cyclic hysteresis, allowable strain, stiffness, and long-term durability were reported inconsistently. Material descriptions therefore provide useful information on fabrication strategies, but do not currently support quantitative ranking of material systems.

5. Sensing, Intention Detection, and Control Architectures

The control of pneumatic soft elbow rehabilitation devices consists of several interconnected functional levels. The highest level defines the rehabilitation task or the assistance required by the user. This is followed by the generation of the desired joint trajectory, torque, or interaction behavior, and then by joint-level control. At the lowest level, the controller adjusts the pneumatic actuator pressure, valve opening state, or pulse-width-modulated valve command.

These control levels have distinct but interconnected functions. High-level control determines the rehabilitation objective or the assistance required by the user and generates a corresponding motion, force, torque, or impedance reference. Joint-level control

then converts this reference into a desired mechanical command for the actuator. At the lowest level, pressure and valve control regulate chamber pressure and airflow so that the pneumatic actuator can physically realize the commanded mechanical output. In cascaded architectures, the output of the higher-level position, force, torque, or impedance controller therefore serves as the reference for the inner pressure or valve-control loop.

5.1. Sensing Methods

The sensing architecture determines which aspects of the pneumatic actuator, joint motion, human–robot interaction, and user state are available for feedback or reference generation. The following subsections group the reviewed methods into pneumatic state sensing, kinematic sensing, force and interaction sensing, self-sensing actuators, EMG, and alternative user-generated signals.

5.1.1. Pneumatic State Sensing

The most direct state variable of a pneumatic actuator is its internal pressure. Pressure measurements can be used as an internal control variable, a safety constraint, or an input for estimating actuator force or joint torque. Gao et al. [20] incorporated pressure, bending, and force sensing into a pneumatic elbow pad, enabling the pneumatic state and mechanical output of the actuator to be examined simultaneously.

Pressure alone, however, does not uniquely determine the resulting elbow angle or torque. The output is also influenced by actuator deformation, elbow position, external loading, displacement of the attachment points, and material hysteresis. Peternel et al. [96] therefore used different calibration procedures to identify the force, length, pressure relationship of a pneumatic muscle. The selected model was used to calculate the pressure required for a desired muscle force and actuator length, and was subsequently applied to torque and position control of an elbow exoskeleton.

Ayazi et al. [68] used load sensors and joint encoders to investigate the relationship among pressure, joint angle, and actuator force in a paediatric-scale pneumatic exosuit. Among piecewise polynomial, sigmoid, and autoregressive models with exogenous inputs, the ARX model provided the most comprehensive description because it also accounted for temporal behavior. This suggests that static pressure-force curves may not be sufficient for dynamic control.

Zheng et al. [77] identified a static characteristic describing the relationship among force, pressure, and actuator deformation, which served as the basis for a PID controller. Such models may be particularly useful in multi-joint systems, where the operation of pneumatic actuators is coupled kinematically or dynamically.

5.1.2. Measurement of Joint Position and Kinematics

Encoders, inertial measurement units, flex sensors, and optical systems can be used to provide feedback on elbow position. In soft wearable devices, IMUs have the advantage of not requiring a rigid and precisely aligned joint axis.

Wilkening et al. [19] combined signals from two IMUs and four resistive flex sensors. An artificial neural network was used to combine the flex-sensor signals, while sensor fusion was applied to integrate the IMU and bending measurements. The purpose of this redundant sensing arrangement was to complement the more accurate but disturbance-sensitive inertial signals with more robust bending measurements.

Proietti et al. [87] placed three IMUs on the trunk, upper arm, and forearm to estimate shoulder and elbow kinematics. The IMU-based estimates were validated using optical motion capture, and the resulting joint data were then used as feedback for gravity-compensation and trajectory-tracking controllers.

Mucchiani et al. [95] also used IMU-based proprioceptive feedback in a pneumatic upper-limb device developed for children. Based on the measured joint angles, a PD controller adjusted actuator pressure to move the limb toward predefined targets within the workspace.

5.1.3. Force and Interaction Sensing

The role of force sensing depends on whether the system aims to determine the tensile force generated by the actuator, the force within the mechanical transmission, the external contact force at the robot endpoint, or the normal force applied to the wearer's body. Sensor placement therefore determines whether the measurement can be used to estimate actuator force, joint torque, or human–robot contact loading directly.

Tomita et al. [97] placed a load sensor in series with the force transmitted through a Bowden cable. The measured quantity was not the direct contact force applied to the skin, but the tensile force in the transmission responsible for generating elbow flexion. In the two-axis serial PAM robot developed by Huy Anh et al. [98], the controlled mechanical variable was the external contact force at the robot endpoint. This was therefore an endpoint force-control configuration rather than a measurement of actuator tensile force or distributed contact loading on the wearer.

Peternel et al. [96] measured the tensile forces of antagonistic pneumatic muscles within the mechanical force-transmission chain. According to the detailed hardware description of the associated system, the load sensor was positioned between the PAM and the tensile cord leading to the rotary joint. The measured force, pressure, and muscle-length data were used to develop an inverse model that determined the pressure required for a desired PAM force and actuator length.

In contrast, Ayazi et al. [16] measured the normal contact force exerted by the actuator on a physical model representing the body. In the elbow module of the experimental setup, two load cells, each with a capacity of 5 kg, were embedded in the anterior surfaces of the upper arm and forearm. Rigid plates placed over the sensors collected the load exerted by the bellows actuator across a larger surface area. Modelling of the elbow actuator primarily focused on the relationship among the force applied to the forearm, joint angle, and internal pressure.

5.1.4. Self-Sensing Actuators

In self-sensing actuators, deformation is measured using an element integrated directly into the actuator structure. Liu et al. [14] incorporated liquid-metal-based displacement sensing into a multi-chamber, reconfigurable pneumatic artificial muscle. Actuator contraction was estimated without a separate external displacement sensor, and the self-sensing capability was evaluated in closed-loop square-wave tracking experiments.

Integrated sensing can reduce overall system size and simplify sensor placement. However, the relationship between the sensor signal and actuator deformation may be influenced by material fatigue, temperature, mechanical preload, and hysteresis. Within the reviewed corpus, self-sensing pneumatic elbow actuators were less common than systems using external IMUs, encoders, or pressure sensors.

5.1.5. EMG Sensing

Surface electromyography is a sensing modality used to measure the user's neuromuscular state and can serve several different control functions in the reviewed systems. EMG can be used to detect movement onset or direction, determine the required level of assistance, or act as a feedback variable. This should be distinguished from cases in which muscle activity is measured only to evaluate the effects of the device retrospectively; such systems cannot be considered EMG-controlled on that basis alone.

In simpler architectures, EMG generates a discrete activation or mode-switching command. Khan et al. [99] distinguished between continuous passive motion and sEMG-based active-assistive operation, while Punjabi et al. [54] used EMG signals to control the 3/2-way valve connected to the appropriate pneumatic. Jiang et al. [100] generated a synergistic command from biceps and triceps sEMG signals and adjusted the PWM duty cycle of a high-speed on-off valve according to control levels assigned to the EMG envelope. In the two-level distributed architecture developed by Dindorf and Wos [84], the upper level processed bioelectrical signals, while the lower level controlled the bi-muscular pneumatic servo drive. The system also provided visual feedback to the user.

In more advanced systems, EMG serves as a continuous reference or feedback variable. Kako et al. [101] regulated pneumatic assistance so that biceps activity followed a desired level during object lifting. The controlled biological output was therefore muscular effort rather than joint position. Wu and Lee [102] used an LSTM network to estimate muscle force from time- and frequency-domain features of sEMG signals and generated a dynamic assistance ratio based on the estimated force. Chen and Chen [30] used a feedforward neural network processing EMG and joint-angle features to estimate the desired elbow motion, which then served as the reference for a cascaded position-torque-pressure controller.

5.1.6. Alternative User-Generated Signals

In addition to EMG-based movement-intention estimation, Tripanpitak et al. [89] investigated the potential use of pain-related evoked potentials, referred to as PREP signals. The authors considered PREP as a possible feedback signal for limiting the range of motion according to pain level. The study therefore presented the concept of a rehabilitation architecture that could adapt to the user's pain state.

Oguntosin and Abdulkareem [86] used eye tracking to control positive- and negative-pressure sources. Based on the gaze position detected through a graphical interface, the system activated inflation or deflation of the pneumatic actuator. This approach does not provide physiological state feedback, but instead serves as an alternative command-input method that may be relevant for users with severely limited upper-limb function.

5.1.7. Comparative Summary of Sensing Architectures

The reviewed sensing architectures monitor different aspects of the actuator, joint, and user. Pressure sensing supports pneumatic regulation and force estimation, whereas kinematic and force sensors provide more direct information about joint motion and mechanical interaction. Self-sensing can reduce the need for external sensors, while EMG enables intention detection and assistance adaptation. However, each modality is affected by specific limitations, including hysteresis, drift, placement sensitivity, and user-dependent variability. Table 4 summarizes the main sensing modalities used in pneumatic soft elbow rehabilitation devices and compares their measured variables, functional roles, and limitations.

Table 4. Comparative overview of sensing modalities in pneumatic soft elbow rehabilitation.

Sensing Modality	Measured Signal and Measurement Location	Primary Function	Main Disadvantage or Limitation	Representative Studies
Pneumatic state sensing	Internal pressure, actuator deformation, or actuator length.	Pressure control and force or torque estimation.	Pressure does not uniquely determine the mechanical output.	[20,68,77,96]
Joint and kinematic sensing	Joint angle, segment orientation, or local bending.	Trajectory tracking and posture estimation.	IMU drift and sensor-placement errors.	[19,87,95]

Table 4. Cont.

Sensing Modality	Measured Signal and Measurement Location	Primary Function	Main Disadvantage or Limitation	Representative Studies
Force and interaction sensing	Transmission tensile force, PAM force or actuator-body normal force.	Determination of actuator force, joint torque, and contact load; force control.	Results strongly depend on measurement location and direction.	[68,96–98]
Integrated actuator self-sensing	A liquid-metal-based deformation sensor embedded in the actuator.	Deformation feedback without an external position sensor.	Hysteresis, drift, fatigue, and temperature dependence.	[14]
Discrete EMG input	Biceps and triceps surface electromyography (sEMG).	Movement initiation and direction or operating-mode switching.	Sensitivity to noise, electrode placement, and threshold selection.	[54,84,99,100]
Continuous EMG estimation	Muscle activity, muscle force, or motion reference estimated from EMG.	Determination of the assistance level and provision of biological feedback.	Inter-subject variability, calibration requirements.	[30,101,102]
PREP-based pain monitoring	Electrically evoked potentials recorded using EEG.	Monitoring of pain state and potential restriction of the range of motion.	Requires external stimulation; supported only by preliminary offline validation.	[89]
Gaze-based command input	Gaze position detected on a graphical user interface.	Inflation or deflation command input.	Dependent on calibration, user attention, and the visual interface.	[86]

5.2. Joint- and Actuator-Level Control Architectures

Following the sensing and reference-generation stages, the role of the control architecture is to produce the desired joint motion, mechanical assistance, or human–robot interaction behavior. The reviewed systems used different controlled variables: some regulated joint position or trajectory, while others controlled actuator force, joint torque, mechanical impedance, or actuator-level pressure. At the lowest level of the hierarchy, pressure and valve control convert the desired mechanical command into pneumatic actuation. Control objective and controller formulation are treated as orthogonal dimensions. Section 5.2 therefore classifies systems according to the physical variable being regulated, whereas Section 5.3 classifies the mathematical or algorithmic method used to achieve that regulation.

In this review, control architectures are classified according to the primary mechanical variable regulated at the joint or actuator level. Position or trajectory control regulates joint configuration or motion; force control regulates a measured linear force at the actuator, transmission, or interaction interface; torque control regulates or estimates rotational mechanical output about the elbow joint; and impedance control regulates the dynamic relationship between motion and interaction force rather than either variable independently. Pressure and valve control are treated separately as actuator-level implementation layers that realize the mechanical commands generated by these higher-level controllers.

5.2.1. Position and Trajectory Control

The objective of position and trajectory control is to track a desired joint angle, workspace target, or rehabilitation trajectory. Feedback is typically obtained from an

encoder, IMU, or another kinematic sensor, while the controller output modifies the pneumatic actuator either directly or through a lower-level pressure-control loop.

Yang et al. [59] calculated the desired joint configuration from the inverse kinematic model of a three-degree-of-freedom bionic elbow joint actuated by PAMs, and then used PID control to track the prescribed rotation angles. Zhao et al. [35] applied PI-based angle control in a silicone pneumatic exosuit. Mucchiani et al. [95] used IMU-based joint-angle estimates as feedback and adjusted actuator pressure with a PD controller so that the limb reached predefined targets within the workspace.

Zheng et al. [77] combined a feedforward model based on the static force-pressure-deformation characteristics of the pneumatic actuator with PID feedback. The feedforward component provided an initial estimate of the pressure required to generate the desired mechanical output, while the feedback loop corrected model errors and external deviations. The approach was also used to reduce coupling between the elbow and wrist modules.

Olvera-Rosales et al. [103] investigated position control under a limited pneumatic pressure range. The controller was adapted to the actuator's physically achievable workspace and torque capacity. This highlights that algorithmic compensation of the dynamics cannot replace appropriate physical sizing of the actuator and selection of its operating pressure range.

5.2.2. Force and Torque Control

Force and torque control aim to regulate the assistance transferred to the user or the mechanical environment. The controlled variable may be actuator tensile force, transmission force, joint torque, or endpoint contact force. These quantities should not be treated as equivalent, as they correspond to different measurement locations and have different mechanical meanings.

Peternel et al. [96] identified the force-length-pressure relationship of a pneumatic muscle. The inverse model was used to calculate the pressure required for a desired actuator force and muscle length, while joint torque was determined from the linear tensile force and the mechanical moment arm. The model therefore established a relationship between the pneumatic state of the actuator and the desired mechanical output.

Huy Anh et al. [98] applied hybrid force-position control in a two-axis PAM-driven robot. The system followed a prescribed trajectory while also regulating the external contact force at the endpoint. This allowed motion and contact conditions to be controlled simultaneously under varying external loads.

Tomita et al. [97] implemented transmission-force tracking in a pneumatic motion-teaching suit. A load sensor positioned in series measured the tensile force transmitted through the Bowden cable, while feedback force control was supplemented by compensation for the delay of the pneumatic drive. In this case, the controlled variable was not the direct skin-contact force but the transmission force responsible for producing elbow flexion.

5.2.3. Impedance, Stiffness, and Compliance Control

Strict position control may be undesirable when the system strongly resists the user's voluntary movement or external disturbances. Impedance control, in contrast, defines the dynamic relationship between the robot and the user rather than enforcing only a geometric trajectory.

Liu et al. [104] applied impedance control based on a virtual coordination tunnel. An allowable region was defined around the coordinated elbow-wrist trajectory generated by a CNN-LSTM model. The controller provided assistance as the user approached the boundaries of this region, while allowing active deviations within it. The system therefore

did not impose a single rigid trajectory, but instead provided the mechanical support required to maintain the desired coordination pattern.

Dimonte et al. [74] achieved partially independent position and stiffness control using antagonistic pneumatic muscles. The pressure difference between the actuators primarily modified the equilibrium position of the joint, while their common pressure level adjusted the mechanical stiffness. This behavior resembles the coactivation of biological agonist–antagonist muscle pairs.

Ma et al. [90] used a vacuum-actuated fibre-jamming structure to modify the mechanical stiffness of the wearable system. Changing the vacuum pressure altered the contact and friction between the flexible fibres and thereby changed the stiffness of the structure. This primarily represents commanded physical stiffness modulation. It should only be considered full impedance control when stiffness is regulated in a closed feedback loop based on a measured interaction variable.

5.2.4. Pressure- and Valve-Level Control

The mechanical output of a pneumatic actuator is ultimately modified by controlling chamber pressure and airflow through the valves. Commands generated by joint-level position, force, or impedance controllers must therefore be implemented by a lower-level pressure or valve controller.

The reviewed systems used proportional valves, high-speed on-off valves, pulse-width modulation, and model-based pressure references. Mucchiani et al. [95] adjusted actuator pressure according to the joint error estimated using IMUs. Zheng et al. [77] calculated the pressure required for the desired mechanical output from a static actuator model and then corrected deviations using feedback control.

Jiang et al. [100] modified the PWM duty cycle of a high-speed on-off valve according to discrete control levels derived from EMG. In this architecture, the user's muscle activity directly determined a valve-control command rather than a continuous pressure reference.

Dan et al. [105] applied fuzzy correction to estimate and adjust the effective flow coefficient of an on-off valve while accounting for nonlinearities in the filling and exhaust processes. Chen and Chen [30] used cascaded position-torque-pressure control, in which the higher-level motion reference was first converted into a torque demand and then into a pneumatic pressure command.

5.2.5. Comparative Summary of Control Architectures

The control architectures of the reviewed systems can be organized around four main target variables: joint position or trajectory, force or torque, mechanical impedance or stiffness, and actuator-level pressure or valve state. Some systems used direct position feedback, whereas others incorporated model-based feedforward control, hybrid force–position control, antagonistic stiffness regulation, or cascaded position–torque–pressure architectures. The main control architectures, their controlled variables, characteristic feedback signals or inputs, architectural features, principal limitations, and representative studies are summarized in Table 5.

These architectures are not mutually exclusive categories but often represent successive levels within a control hierarchy. In an advanced system, a higher-level position, force, or impedance controller generates a mechanical reference that is implemented by lower-level pressure and valve control. System performance is therefore determined not only by the selected high-level control objective, but also by sensing accuracy, the validity of the actuator and interface models, and the dynamic limitations of the pneumatic subsystem.

Table 5. Comparative overview of the main control architectures in pneumatic soft elbow rehabilitation.

Control Architecture	Controlled Variable	Typical Feedback or Input	Architectural Form	Main Limitation	Representative Studies
Position and trajectory control	Joint angle, end-effector position, or prescribed trajectory.	Encoder, IMU, or estimated joint angle.	Direct feedback control or a position loop supplemented by model-based feedforward compensation.	Sensitive to load variations, hysteresis, and the physical pressure limits of the actuator.	[35,59,77,95,103]
Force and torque control	PAM force, transmission tensile force, joint torque, or endpoint contact force.	Load cell, force-length-pressure model, or endpoint force measurement.	Direct force loop, model-based torque control, or hybrid force-position control.	The result depends on the measurement location and on whether actuator, transmission, or contact force is controlled.	[96–98]
Impedance, stiffness, and compliance control	Mechanical impedance, equilibrium position, stiffness, or allowable trajectory deviation.	Joint position, coordination error, actuator pressure, or stiffness command.	Virtual tunnel, antagonistic co-activation, or physical stiffness modulation.	Commanded stiffness does not necessarily constitute closed-loop impedance control.	[74,90,104]
Pressure- and valve-level control	Chamber pressure, valve state, PWM duty cycle, or mass flow rate.	Pressure signal or a higher-level position, torque, or EMG command.	Low-level pressure loop, on-off or proportional valve control, or a cascaded architecture.	Valve dead time, flow nonlinearity, air compressibility, and asymmetric filling and exhausting dynamics.	[30,77,95,100,105]

5.3. Control Methods for Compensating Nonlinearities and Uncertainties

Pneumatic soft actuators are characterized by strong nonlinearities, hysteresis, pressure- and load-dependent behavior, and time-varying mechanical and pneumatic parameters. Additional uncertainty may arise from displacement of the wearable interface, the deformable connection between the actuator and the body, and changes in the user's mechanical properties. The reviewed studies applied classical model-based, adaptive, sliding-mode, fuzzy, and neural methods to compensate for these effects.

5.3.1. Classical Feedback and Model-Based Feedforward Control

PI, PD, and PID controllers are straightforward to implement, require relatively little computation, and can provide adequate performance within a limited operating range. Yang et al. [59], Zhao et al. [35], and Mucchiani et al. [95] applied classical feedback control to track joint positions or workspace targets.

Classical feedback can also be combined with model-based feedforward control. Zheng et al. [77] calculated an initial pressure command from the static actuator characteristics and then used PID feedback to correct the remaining error. Peternel et al. [96] used an inverse model of the force-length-pressure relationship to determine the pressure required for a desired actuator force.

Xu et al. [106] applied an adaptively updated inverse Prandtl–Ishlinskii model for feedforward compensation of pneumatic-muscle hysteresis. The parameters of the inverse

model were updated online using a projection algorithm, making the compensation less dependent on a single previously identified static hysteresis characteristic.

5.3.2. Adaptive and Model-Reference Control

Adaptive control aims to adjust the controller parameters online in response to changes in system dynamics. This is particularly relevant when actuator behavior varies with pressure, joint position, external loading, or the mechanical properties of the user.

Irshaidat et al. [23] applied model-reference adaptive control to an elbow exoskeleton driven by a bending pneumatic muscle. The desired closed-loop dynamics were defined by a reference model, while the adaptation mechanism adjusted the controller parameters to match the behavior of the actual system.

Zhang et al. [107] developed an adaptive fault-tolerant controller for a PAM-driven upper-limb rehabilitation exoskeleton. The method aimed to maintain robust trajectory tracking under actuator faults and constrained control inputs. Its adaptive compensation did not require prior knowledge of the unknown system parameters or the magnitude of the actuator fault.

5.3.3. Sliding-Mode and Prescribed-Performance Control

Sliding-mode control is frequently applied in PAM systems because its formulation can provide robustness to bounded modelling uncertainties and external disturbances under the assumptions of the controller design. However, conventional switching laws may produce chattering, meaning high-frequency oscillations in the control signal that can lead to pressure fluctuations and undesirable interaction behavior.

Zhao et al. [108] combined model-free adaptive control, a prescribed-performance approach, and terminal sliding-mode control. The prescribed-performance transformation imposed predefined transient and steady-state error bounds, while the model-free adaptive component reduced the need for a complete dynamic model.

Zhao et al. [109] applied a second-order super-twisting sliding-mode algorithm with adaptive dynamic compensation and a prescribed-performance function. The purpose of the higher-order sliding-mode method was to preserve robustness while reducing chattering and producing a smoother control signal.

Nguyen et al. [110] supplemented adaptive fast terminal sliding-mode control with fuzzy approximation and a PID compensator. The fuzzy component approximated unknown nonlinear system terms, while the PID compensator addressed approximation errors and residual hysteresis effects.

Chen et al. [75,76] applied fuzzy and adaptive fuzzy sliding-mode control to a multi-degree-of-freedom cable-transmitted PAM drive. Guo et al. [111] supplemented a sliding-mode controller with an RBF neural network to estimate uncertain dynamic terms and external disturbances online.

5.3.4. Fuzzy, Neural, and Hybrid Compensation

Fuzzy and neural methods can approximate complex nonlinear relationships without requiring a complete analytical model. These approaches may be used as standalone controllers, online model estimators, or compensating elements within classical and robust control architectures.

Dan et al. [105] developed a fuzzy indirect adaptive robust controller. A modified fuzzy RBF network estimated model uncertainties and unknown control parameters, while fuzzy correction adjusted the flow coefficient of a high-speed on-off valve in real time. The approach therefore addressed uncertainty in both the joint mechanics and the pneumatic flow subsystem.

Huy Anh et al. [98] used an adaptive dynamic neural network to identify the nonlinear dynamics of a two-axis PAM-driven robot. The estimated model was then incorporated into a hybrid controller combined with a PID component. The neural model approximated both inter-axis coupling and the influence of varying external contact loads.

In the hierarchical system developed by Wu and Lee [102], an sEMG-based LSTM model estimated the user's muscle force and generated a dynamic assistance ratio, while a two-channel fuzzy neural controller compensated for pneumatic-muscle nonlinearities and hysteresis at the actuator level. Neural methods therefore operated at two distinct levels: the higher level estimated the user's state, while the lower level compensated for actuator dynamics.

5.3.5. Comparative Summary of Control Methods

The reviewed control methods range from classical PI, PD, and PID feedback to adaptive, sliding-mode, fuzzy, and neural compensation. Classical approaches require relatively little computation and are straightforward to implement, but they have limited ability to adapt to pneumatic-actuator hysteresis, load dependence, and time-varying dynamics. Adaptive and sliding-mode methods aim to provide greater robustness, while fuzzy and neural approaches enable online or data-driven approximation of difficult-to-model nonlinearities, system coupling, and actuator faults. The main control methods and representative studies are summarized in Table 6.

Table 6. Comparative overview of the main control methods in pneumatic soft elbow rehabilitation.

Method Family	Compensation Principle	Typical Control Structure	Addressed Uncertainty or Nonlinearity	Main Limitation	Representative Studies
Classical feedback and model-based feedforward control	PI, PD, or PID regulation of the tracking error; estimation of the nominal command from an actuator or system model.	PID feedback loop or combined feedforward-feedback structure.	Limited model error, static actuator characteristics, and moderate disturbances.	Limited adaptability under strongly nonlinear, hysteretic, or variable-load operating conditions.	[34,58,76,92,93,103]
Adaptive, model-reference, and fault-tolerant control	Online adjustment of controller parameters or fault-compensation terms.	Reference-model-based adaptation or adaptive fault compensation.	Time-varying parameters, unknown system dynamics, actuator faults, and input constraints.	Stability and performance depend on the adaptation law and on the assumed bounds of faults and uncertain parameters.	[23,107]
Sliding-mode and prescribed-performance control	Forcing the system state toward a sliding surface while maintaining the tracking error within prescribed bounds.	Terminal or super-twisting sliding-mode control, often combined with adaptive or fuzzy compensation.	Model uncertainty, external disturbances, and transient and steady-state tracking errors.	Chattering, complex parameter tuning, and uncertainty in transferring formal error bounds to the physical system.	[75,76,108–111]
Fuzzy, neural, and hybrid compensation	Data-driven or rule-based approximation of nonlinear relationships.	Fuzzy RBF network, adaptive neural model, LSTM, fuzzy-neural controller, or neural-PID hybrid.	Hysteresis, inter-axis coupling, valve-flow nonlinearities, user state, and actuator dynamics.	Data and tuning requirements, complex stability analysis, and limited extrapolation beyond the training range.	[30,98,102,105]

A direct quantitative ranking of these control methods was not possible because the reviewed studies used different actuator architectures, pressure ranges, trajectories,

external loads, sampling conditions, and performance metrics. Reported values of tracking error, settling time, or robustness therefore cannot be compared directly across studies. Consequently, Table 6 provides a functional rather than performance-based comparison. Future benchmarking should evaluate different controllers on identical hardware and under standardized trajectories, loads, pressure limits, and evaluation metrics.

6. Discussion

6.1. Discussion and Summary of Evidence

The findings of this scoping review indicate that pneumatic soft elbow rehabilitation devices cannot be understood solely in terms of the use of a particular actuator technology. Their operation is jointly determined by the energy-converting actuator, mechanical transmission, body-worn interface, sensing system, and multilevel control architecture. Because these subsystems are closely interconnected, the mechanical advantages of the actuator can only be realized at the system level when force transmission, attachment, state estimation, and control are adapted to the characteristics of the deformable pneumatic structure.

RQ1: Four main actuation technology families were identified in soft and compliant elbow rehabilitation devices: pneumatic soft actuators, motor-driven cable and tendon systems, series-elastic or variable-stiffness electromechanical actuators, and shape-memory-alloy-based systems. These technologies differ primarily in terms of the subsystem in which energy conversion, mechanical compliance, and force transmission are implemented. Pneumatic soft actuators occupy a distinct position because mechanical power generation and intrinsic compliance are often integrated within the same deformable chamber or artificial muscle. These characteristics have motivated the use of pneumatic actuation in wearable rehabilitation, particularly where compliant and body-conforming mechanical interfaces are desired. However, the reviewed studies do not provide standardized quantitative evidence demonstrating lower mass or superior anatomical adaptability relative to alternative actuator families.

RQ2: The structural architectures of pneumatic soft elbow actuators are organized around two fundamental mechanical output principles. Linear pneumatic artificial muscles convert pressure-induced radial or geometry-driven deformation into axial contraction or, less commonly, elongation. Bending and rotary actuators, in contrast, generate joint rotation directly through the controlled deformation of chambers, bellows elements, or patterned textile layers. These two basic principles are complemented by cable-, tendon-, and flexure-transmitted configurations, rigid–soft integrated structures, and modular, reconfigurable, antagonistic, and bidirectional arrangements. Jamming-based, self-sensing, and variable-stiffness solutions indicate a broader development trend from simple actuation toward increasingly functionally integrated structures.

RQ3: The most commonly used sensing modalities were internal pressure, joint position and kinematics, and mechanical forces measured at different locations. When interpreting force measurements, it is important to distinguish among actuator tensile force, transmission force, joint torque, endpoint contact force, and the normal force between the actuator and the body. EMG-based systems also served different functions: the signal could act as a discrete activation command, a continuous assistance reference, an estimate of muscle force, or a biological feedback variable. Self-sensing actuators, PREP-based pain monitoring, and gaze-based command input were less common but represented functionally distinct approaches.

To provide an integrated overview of the findings across the three research questions, Table 7 consolidates the principal synthesis domains, key evidence themes, and representative references identified in the reviewed literature.

Based on the controlled variable, the reviewed architectures included position and trajectory control, force and torque control, impedance and stiffness control, and pressure- and valve-level control. Pneumatic-actuator hysteresis, load dependence, time-varying parameters, and flow nonlinearities were addressed using classical feedback, model-based feedforward control, adaptive and fault-tolerant methods, sliding-mode and prescribed-performance approaches, and fuzzy, neural, and hybrid compensation. Although advanced algorithms may improve robustness, they cannot replace appropriate physical actuator sizing, reliable sensing, and a properly designed inner pneumatic control loop.

Importantly, demonstration of actuator force, range of motion, trajectory tracking, or pressure-control performance should not be interpreted as evidence of clinical rehabilitation effectiveness. Most of the reviewed studies primarily established technical feasibility, whereas evidence for improved motor recovery, functional independence, long-term usability, or other clinical outcomes remains considerably more limited.

Overall, the reviewed literature shows a progression in pneumatic soft elbow rehabilitation from simple pressure-activated actuators toward hierarchically controlled, functionally integrated systems that also account for the user's state. The main direction of technological development is not limited to increasing force or range of motion, but includes the combined implementation of programmable deformation, anatomical adaptability, integrated sensing, adjustable mechanical behavior, and user-specific assistance.

Table 7. Integrated summary of findings across the three research questions.

Synthesis Domain		Key Evidence Theme	Representative Studies
RQ1	Actuation technologies	Pneumatic Soft Actuation	[14,19,20,22,23]
		Motor-Driven Cable and Tendon Actuation	[6,7,37,42–44]
		Series-Elastic and Variable-Stiffness Actuation	[8,9,45,46,48–51]
		Shape-Memory-Alloy Actuation	[10,11,52,53]
RQ2	Pneumatic actuator architectures	Linear Pneumatic Artificial Muscles	[3,54–58,60–64]
		Pneumatic Bending and Rotary Actuators	[12,15,16,65–69]
		Cross-Cutting System and Architectural Attributes	[5,13,14,61,69–93]
		Materials and Fabrication Methods	[13,15,54,55,57,61,63,70,74,79,94,95]
RQ3	Sensing and control architectures	Sensing Methods	[14,19,20,30,54,68,77,84,86,87,89,95–102]
		Joint- and Actuator-Level Control Architectures	[30,35,59,74,77,90,95–98,100,103–105]
		Control Methods for Compensating Nonlinearities and Uncertainties	[23,30,34,58,75,76,92,93,98,102,103,105,107–111]

6.2. Challenges and Constraints

The main challenge in pneumatic soft elbow rehabilitation is the accumulation of uncertainty across the complete human–robot system. Actuator hysteresis, pressure-dependent behavior, external loading, attachment migration, soft-tissue deformation, and anthropometric variability all influence the resulting force and joint torque. Consequently, models and controllers validated under fixed laboratory conditions may not transfer directly across users or sessions.

The wearable interface and sensing architecture are particularly important sources of uncertainty. Pressure alone does not reliably describe the mechanical state, IMU measurements may be affected by drift and placement errors, and force measurements are difficult to compare unless the measured quantity and sensor location are clearly defined. EMG-based intention estimation remains sensitive to fatigue, electrode placement, and user-specific calibration, while integrated self-sensing still requires stronger evidence regarding long-term drift and durability.

The clinical evidence also remains limited. Most studies focused on technical feasibility, short-duration laboratory experiments, or healthy participants. Tracking accuracy, actuator force, and pressure regulation do not by themselves demonstrate improved motor recovery, usability, or patient adherence. In addition, inconsistent reporting of actuator geometry, operating pressure, attachment configuration, sensor placement, and validation conditions limits reproducibility and cross-study comparison.

6.3. Conclusions and Recommendations

Pneumatic soft elbow rehabilitation should be treated as a system-level design problem involving the actuator, transmission, wearable interface, sensing, and control. No single actuator, sensing modality, or control strategy can currently be considered universally optimal, because suitability depends on the rehabilitation task, required torque, patient characteristics, attachment tolerance, and pneumatic infrastructure.

Future work should prioritize co-design of the actuator, transmission, and attachment geometry, together with multimodal state estimation that combines pressure, kinematic, and force measurements. Hierarchical control should integrate user-intention or muscular-effort information with joint-level control and reliable inner pressure and valve loops. Advanced adaptive or learning-based methods should also respect explicit physical and safety constraints.

More standardized and clinically relevant validation is required. Future studies should compare controllers on identical hardware, report actuator and interface parameters consistently, and include longer-duration experiments with target patient populations. Technical performance should be linked to clinically meaningful outcomes such as comfort, usability, reduced muscular effort, functional improvement, and rehabilitation effectiveness. The current evidence base is dominated by technical and proof-of-concept validation. Patient-based studies remain comparatively limited, and the available literature does not support broad conclusions regarding clinical effectiveness.

6.4. Limitations

This scoping review has several limitations. First, its purpose was to map the extent, concepts, and technological architectures of the literature rather than to perform a formal critical appraisal of the methodological quality of the included studies. The manuscript therefore does not establish a quality-based ranking of individual actuators or control methods.

Second, relevant publications may have been missed because of the selected databases, search terms, language restrictions, and differences in indexing practices across databases. The focus on peer-reviewed literature may also have excluded prototypes and industrial developments reported in patents, technical reports, or non-peer-reviewed sources. Study screening was performed by a single reviewer rather than independently in duplicate, which may have increased the risk of selection bias despite the use of predefined eligibility criteria and conservative retention of uncertain records.

Finally, the studies used different actuator geometries, pressure ranges, loads, measurement locations, rehabilitation tasks, and validation environments. This heterogeneity

prevented quantitative synthesis and direct ranking of the reported mechanical and control results. The conclusions therefore primarily reflect the identified technological and functional patterns and should not be interpreted as comparative evidence of the clinical effectiveness of individual devices.

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