

Article

Design, Simulation and Optimization of a Novel Knee-Rehabilitation Mechanism with Passive-Self-Alignment Segmented Redundant Joints for Stroke Patients

Meng Gao ^{1,*}, Hujiang Wang ¹, Yaqi Wang ², Da Jiang ¹, Wen Zhang ¹, Wentao Feng ³ and Fuqun Zhao ^{4,*}

¹ National Engineering Laboratory for Wheeled Vehicle, China North Vehicle Research Institute, Beijing 100072, China; wanghujiang@noveri.com.cn (H.W.); ziangdar@sina.com (D.J.); lxankai@163.com (W.Z.)

² Department of General Surgery, Peking Union Medical College Hospital, Chinese Academy of Medical Sciences and Peking Union Medical College, Beijing 100730, China; wangyaqi6666@sina.com

³ Beijing North Vehicle Group Corporation, Beijing 100072, China; fengwentao1106@163.com

⁴ Department of Mechanical Engineering, Beijing Jiaotong University, Beijing 100044, China

* Correspondence: gao_meng2018@163.com (M.G.); fqzhao@bjtu.edu.cn (F.Z.)

Abstract

With the increasing number of stroke patients, there is a growing demand for lower-limb rehabilitation exoskeletons. While current mechanisms are preferred for their light weight and dexterous design in limited environments, the alignment of the structures and motion are still not matched perfectly to human movements. This study develops a novel structure and configuration optimization method for knee part rehabilitation with special passive self-alignment modules. The driving segment is mechanically coupled to the patients' lower limb. All components are designed with high rigidity and fully constrained to ensure smooth and continuous motion. Then, the kinematics are systematically derived to establish the foundation for the control system. Next, the application of the particle swarm optimization algorithm determines the optimal parameters for each revolute joint during the bending motion, and reduces the non-ideal S-shaped motion deformation curve caused by the offset of the joint rotation center and the load at the end effector successfully. The final results demonstrate that the optimized SRE achieves 97.5% motion accuracy under large-angle knee movement. This work presents simulation-only validation, and clinical testing remains future work. The proposed mechanism provides a promising solution for post-stroke rehabilitation, and is also applicable to geriatric lower-limb weakness and orthopedic postoperative recovery.

Keywords: rehabilitation exoskeleton; PSO algorithm; self-alignment



Academic Editor: Ashwin Ashok

Received: 30 April 2026

Revised: 7 June 2026

Accepted: 24 June 2026

Published: 1 July 2026

Copyright: © 2026 by the authors.

Licensee MDPI, Basel, Switzerland.

This article is an open access article distributed under the terms and

conditions of the [Creative Commons Attribution \(CC BY\)](https://creativecommons.org/licenses/by/4.0/) license.

1. Introduction

Knee dysfunction resulting from stroke, spinal cord injury, or analogous disorders generally leads to abnormal muscular activities, such as mistimed or insufficient muscle contractions, significantly affecting gait function [1]. Stroke, a common cause of mobility impairment, frequently leads to unilateral limb weakness in patients [2], who may show a semicircular leg-dragging gait during locomotion [3]. Weak or paralyzed knee stabilizer muscles further undermine weight-bearing stability, increasing the risk of collapse during the stance phase [4,5]. Therefore, knee exoskeletons have emerged as a feasible intervention to correct pathological gait in these patients [6,7]. However, the increasing demand for personalized long-term rehabilitation has increased the workload of therapists and

treatment costs [8]. To address these issues and facilitate rehabilitation at home, various robotic exoskeleton designs have been developed [9]. Lower-limb exoskeletons have broad applications beyond post-stroke rehabilitation, including geriatric muscle weakness and orthopedic postoperative recovery. Compared with active exoskeletons, which require powerful actuators, passive exoskeletons offer the advantages of being light, inexpensive, and safe, suitable for early rehabilitation. AI-based optimization for exoskeleton design is still emerging, and this research contributes to the advancement of PSO-based customization for knee rehabilitation exoskeletons.

Driven by technological advances and the growing demand for rehabilitation of the lower extremities, research in this field has expanded considerably [10–12]. Continuous Passive Motion (CPM) devices, which were introduced in the 1970s, initially represented the early generation of rehabilitation equipment [13,14]. For example, the 1988 BioDex system provided programmable, force-regulated support for muscular rehabilitation [15,16]. Subsequent designs included the NeXOs (characterized by multi-segment structures for passive/assisted motion [17,18]) in 2009, the 3-degrees-of-freedom Physiotherabot (supporting both passive and active rehabilitation modes [19,20]), and the Gait Trainer GT-I (facilitating gait simulated training [21,22]) in 2011.

The Lokomat, a well-recognized wearable exoskeleton, allows patients to walk on a treadmill while adjusting their gait through human–device interaction [23,24]. Similar systems such as AutoAmbulator and PAM/POGO also reduce physical exertion during rehabilitation [25,26]. Nevertheless, existing solutions still face the following notable limitations: rigid mechanical structures that cannot adapt to individual patient anthropometry, inadequate intelligent control modules to track recovery progress in real time, and lack of personalized rehabilitation methods [27,28]. Focusing on the actuator design of soft rehabilitation robots, Lai et al. proposed a novel pneumatic actuator of the lattice structure for wrist rehabilitation robots and completed systematic modeling and performance evaluation, providing a new structural solution for the design of flexible driving units of rehabilitation robots [29]. In terms of lightweight and portable wearable rehabilitation devices, Li et al. developed an untethered adaptive thumb exoskeleton that provides precise and delicate rehabilitation assistance for finger movements and expands the application scenarios of small-scale rehabilitation exoskeletons [30]. To improve the active assistance performance of the upper-limb rehabilitation equipment, Xie et al. constructed a cable-driven rehabilitation robot based on the characteristics of human muscle synergy that effectively improves the precision of myoelectric control and the coordination ability between humans and robots during upper-limb rehabilitation training [31]. In order to meet the adaptive walking requirement of lower-limb rehabilitation devices, Best et al. proposed a data-driven variable impedance control method for knee–ankle-powered prostheses, which enabled the robotic device to adapt to different walking speeds and road slopes [32]. In terms of lightweight passive rehabilitation assistance, Williams et al. compared the rehabilitation effects of spring resistance and motorized resistance structures through the design and validation of a portable ankle-assisted device, providing a practical reference for the lightweight design of lower-limb rehabilitation exoskeletons [33]. For neurological rehabilitation scenarios, Dell’Eva et al. developed a hybrid robotic rehabilitation device that integrates multiple functional modules, which can effectively assist gait training for patients with neurological movement disorders [34]. In summary of the current progress of wearable rehabilitation robots, Cha et al. comprehensively reviewed the research status of gait-assisted wearable robots and noted the existing limitations and future development trends in the field of robotic gait rehabilitation [35]. In terms of robust control of lower-limb exoskeletons, Wang et al. designed an extended state observer-based sliding mode control strategy for torque-sensor-free hip exoskeletons, which improves the motion stability and

anti-interference performance of lower-limb rehabilitation robots [36]. In addition, effective human-to-robot collaboration remains a crucial challenge. Exoskeletons must dynamically adapt to human motion patterns while ensuring safety and bio-compatibility, which requires advanced control strategies that meet the requirements for rehabilitation robotics. Although self-aligning knee exoskeletons have been proposed [25,28], most achieve only 1D/2D alignment and lack effective deformation suppression. Recent work on soft actuators and adaptive exoskeletons further highlights the need for lightweight, compliant designs.

In addition, structural adaptability and self-alignment design have become key research directions for knee rehabilitation exoskeletons, primarily to resolve the kinematic mismatch between robotic structures and human knee joints and reduce unwanted forces in the human–robot interaction [37]. Li et al. designed a self-aligning knee exoskeleton with auxiliary hip rotational degrees of freedom that compensates for spatial misalignment of the joint and improves the kinematic compatibility of lower-limb rehabilitation equipment [38,39]. Yang et al. investigated the structural optimization of adaptive knee exoskeletons, providing a practical design approach to improve the motion match between mechanical structures and human knee movement during rehabilitation training [40]. Liu et al. proposed a lightweight knee exoskeleton inspired by human ligament biomechanical characteristics, which effectively reduces interaction force and enhances wear comfort during walking rehabilitation [41]. Focusing on daily rehabilitation scenarios, Kittisares et al. developed a dual four-bar-linkage knee exoskeleton with ergonomic characteristics, implemented adaptive motion adaptation for stair rehabilitation, and expanded the forms of structural design of knee assistive devices [42].

As an efficient intelligent optimization algorithm, particle swarm optimization (PSO) has been widely applied to structural design, parameter tuning, and performance optimization of lower-limb rehabilitation exoskeletons. Song et al. adopted a multi-objective PSO method to optimize the layout of plantar pressure sensors for lower-limb exoskeletons, improving gait detection accuracy and laying the foundation for precise knee rehabilitation motion control [43]. Zhang et al. integrated the PSO algorithm with an extreme learning machine to build a gait recognition model that effectively improves the system adaptability of the exoskeletons to complex knee motion states in different rehabilitation stages [44]. To enhance the control stability of exoskeleton systems, Li et al. used PSO to realize adaptive real-time tuning of control parameters, optimize trajectory tracking performance of rehabilitation robots, and improve anti-interference capability [45]. Wang et al. combined PSO with a neural network of radial basis functions to construct a hybrid control strategy that optimizes the performance of the output torque of the knee exoskeletons and promotes stable and efficient rehabilitation actuation [46]. Tian et al. utilized the PSO algorithm to optimize the kinematic parameters of the knee rehabilitation mechanisms, achieving optimal structural synthesis and effectively improving the motion transmission efficiency of knee joint assistance systems [47].

This paper introduces an ergonomic exoskeleton design featuring a flexible arctic hybrid structure, achieving an optimal balance between active controllability and passive adaptability. Then, it details the gait data acquisition process and the structural design of the exoskeleton. Next, this paper analyzes the performance characteristics of the proposed system, assesses its compatibility with human motion, and addresses the non-ideal S-shaped curve issue. Finally, it summarizes key conclusions and presents promising directions for future research on intelligent control optimization for rehabilitation of exoskeletons. All results are simulation-only, and human/clinical trials will be shown in future work.

The main novelties are summarized as follows:

1. A segmented redundant joint mechanism is proposed to realize full 3D passive self-alignment, while most existing designs only achieve 1D/2D alignment.

2. A hybrid rigid–flexible structure with distributed torsion springs is developed to suppress non-ideal S-shaped motion deformation, which has not been addressed in previous knee exoskeletons.
3. A PSO-based multi-joint parameter optimization method is proposed, which achieves 97.5% motion accuracy under large-angle knee movement, outperforming current prototypes.

2. Data Collected from the Motion Capture System

To address the limitations in personalized adaptation and human–robot interaction (HRI) identified within existing rehabilitation exoskeletons, a high-precision 3D motion capture system was utilized to acquire lower-limb motion data for the design and optimization of exoskeletons. Fifteen retroreflective markers, positioned according to the standard Helen Hayes marker set protocol, were affixed to the lower extremities of the participants, as shown in Figure 1. The participants performed treadmill walking tasks and leg raisers in a designated testing area. Six peripheral infrared cameras were synchronized to capture the complete motion cycles of the lower limb during ambulatory and seated conditions. These captured kinematic data constituted the foundational input for subsequent exoskeleton development. The database includes eight healthy participants (age: 24 ± 3 years; height: 172 ± 5 cm; weight: 68 ± 7 kg). “Maximum Angle” represents the maximum flexion angle, “Maximum Moment” is the maximum torque, and “Rotational Angle of Testers” denotes the average natural knee movement angle—these are the raw experimental data directly measured from the motion capture tests of the eight participants. All values are averaged from the measured data. The control strategy references the physiological knee motion of healthy subjects to guide functional recovery. Table 1 summarizes the maximum angles and torques measured in sitting posture, which define the design requirements for standing and sitting motions.



Figure 1. Motion capture experiments (Numbers 1–6 denote the placement positions of the six cameras.).

The captured motion data allowed quantification of knee joint range of motion (ROM) and real-time torque fluctuations, with the key metrics summarized in Table 1. A central emphasis of this study is the mechanical response of the knee joint under exoskeleton-assisted control, a dimension that is often under-emphasized in conventional designs. Many existing lower-limb exoskeletons prioritize basic kinematic tasks but neglect the dynamic requirements of HRI, which represents a significant gap for rehabilitation systems. For stroke patients, who generally have muscular weakness, this oversight restricts the utility of the exoskeleton. Weak knee stabilizer muscles impede independent joint stabilization, flexion/extension, and locomotion. As a result, exoskeletons are required to provide adaptive compensatory stabilizing torque around the knee to support standing and walking.

Table 1. Range of angle and torque of the knee joint.

	Maximum Angle	Maximum Moment	Rotational Angle of Testers
Walking	67°	36.8 Nm	42°
Sitting	117°	62.5 Nm	102°

3. Modeling Design of Knee Rehabilitation Mechanism

3.1. Biomechanical Description of Knee Joint Configuration

A thorough understanding of the biomechanics of the knee joints serves as the cornerstone for the design of intelligent-adaptability exoskeletons, especially for the alignment of the mechanical structure with the dynamic motion of the human lower limb. For a healthy adult, the gait cycle primarily comprises two distinct phases: the stance phase and the swing phase. The stance phase can be further subdivided into the weight-acceptance stage and the single support stage. Additionally, knee motion exhibits six degrees of freedom (DOF), comprising three rotational movements—flexion/extension, internal/external rotation, and abduction/adduction—alongside three translational movements: anterior/posterior, medial/lateral, and superior/inferior. Although flexion and extension are the main functional movements, internal/external rotation and anterior/posterior translation are crucial to maintaining joint stability during locomotion. The complete rigid restriction of all secondary translational and rotational motions will increase joint reaction forces and further cause mechanical wear or discomfort; by contrast, our scheme only restricts the anterior/posterior translation of the X-axis while retaining free movement of the YZ-plane, effectively avoiding redundant extra joint load.

Figure 2 illustrates points A and B, which represent the exoskeleton and the lower limb, respectively. During bending experiments, the changes in the rotation angle cause the rotation center of both the exoskeleton and the knee to shift from their original positions. Each time the angle changes, the knee's rotation center (O) produces a micro-variation in its position along the three translation axes: dl (along the x -axis), dh (along the z -axis), and $d\omega$ (along the y -axis). Then, the angles change accordingly, including angle α (rotates about the x -axis), angle θ (rotates about the z -axis), and angle ω (rotates about the y -axis). This misalignment significantly impacts the functionality of the exoskeleton. To address this, it is necessary to passively achieve self-alignment of the rotation center of the knee joint.

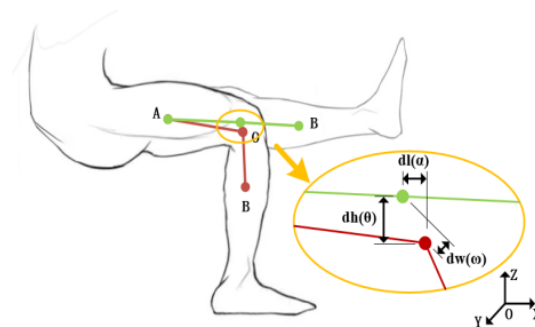


Figure 2. Knee flexion system (sitting posture; standing follows opposite loading, maximum rotation center offset occurs here, detailed in Section 3.4. The symbols $dh(\theta)$, $d\omega(\omega)$, and $dl(\alpha)$ represent the deviation values along the x -, y -, and z -axes in space, respectively.)

For patients with lower-extremity impairments, surgical interventions are still only applicable in select cases, such as those that do not respond to conventional prosthetics. Wearable robotic exoskeletons have emerged as noninvasive alternatives, facilitating prosthetic adaptation and mobility restoration for individuals with spinal cord injuries or bilateral transfemoral amputations. However, existing designs often lack bio-mechanically aligned structures to support intelligent human–robot interaction (HRI), and this gap is addressed by the control-oriented modeling in this study.

Thus, a comprehensive understanding of how the exoskeleton differs from human limbs is necessary. The self-alignment mechanism ensures that the components of the exoskeleton automatically adjust to the shifting rotation center of the knee, maintaining proper alignment with the natural movement of the knee joint.

3.2. Configuration Synthesis of the Knee Joint

The knee exhibits six-degrees-of-freedom motion, but anterior–posterior (X-axis) translation is the main cause of human–exoskeleton misalignment. The medial–lateral (Y-axis) and proximal–distal (Z-axis) translations are small and can be passively compensated for by the flexible segments of the proposed SRE. Therefore, restricting only X-axis translation effectively reduces complexity while maintaining alignment capability. To achieve bidirectional human–machine adaptation, which serves as the core requirement for rehabilitation of exoskeletons, a sub-mechanism was devised to facilitate full rotational motion and translational movement strictly confined to the YZ plane. This sub-mechanism is designed to impose no constraint forces along the Y or Z axes and introduces no constraint couples about any axis. The fundamental constraint is solely the restriction of translational motion along the X-axis. This design guarantees that the exoskeleton maintains flexibility to adapt to natural human motion while providing reliable structural support.

Using these flexibility and constraint characteristics fully, the lower-limb exoskeleton mechanisms can adapt to the various motions of individuals without sacrificing stability while maintaining a certain level of stability, thus improving its applicability for stroke patients with different gait impairments.

The basic formula of the constraint-screw system for the sub-mechanism is defined as follows:

$$\mathbf{S}_a^r = \left\{ \mathbf{S}_{a1}^r = (1, 0, 0, 0, 0, 0)^T \right\} \quad (1)$$

This is a constraint force along the X-axis. The standard base of the motion-screw system for the sub-mechanism can be obtained as follows:

$$\mathbf{S}_a = \begin{cases} \mathbf{S}_{a1} = (1, 0, 0, 0, 0, 0)^T \\ \mathbf{S}_{a2} = (0, 1, 0, 0, 0, 0)^T \\ \mathbf{S}_{a3} = (0, 0, 1, 0, 0, 0)^T \\ \mathbf{S}_{a4} = (0, 0, 0, 1, 0, 0)^T \\ \mathbf{S}_{a5} = (0, 0, 0, 0, 1, 0)^T \end{cases} \quad (2)$$

Then, three new screws, $\mathbf{S}'_{a1}$, $\mathbf{S}'_{a2}$ and $\mathbf{S}'_{a3}$, can be obtained using the linear combination of screws $\mathbf{S}_{a1}$, $\mathbf{S}_{a2}$ and $\mathbf{S}_{a3}$, which can be expressed as follows:

$${}^1\mathbf{S}_a' = \begin{cases} \mathbf{S}'_{a1} = (l_1, m_1, n_1, 0, 0, 0)^T \\ \mathbf{S}'_{a2} = (l_2, m_2, n_2, 0, 0, 0)^T \\ \mathbf{S}'_{a3} = (l_3, m_3, n_3, 0, 0, 0)^T \end{cases} \quad (3)$$

which presents a 3R spherical subchain or a spherical joint. However, spherical joints are excluded due to insufficient structural stiffness and poor load-bearing capacity for rehabilitation tasks. Helical pairs are avoided because they introduce undesired axial motion, which conflicts with the requirements of pure knee flexion/extension. Thus, the equation can be used to construct a 3R spherical subchain. This can be used to obtain new screws as follows:

$${}^1\mathbf{S}_a'' = \begin{cases} \mathbf{S}''_{a1} = (l'_1, m'_1, n'_1, 0, b'_1, c'_1)^T \\ \mathbf{S}''_{a2} = (l'_2, m'_2, n'_2, 0, b'_2, c'_2)^T \\ \mathbf{S}''_{a3} = (l'_3, m'_3, n'_3, 0, b'_3, c'_3)^T \end{cases} \quad (4)$$

This formula denotes three successive revolute joints whose axes do not pass through a common point. That is, the submechanism may consist of three successive revolute

joints that are not a 3R spherical subchain. In addition, screws $\mathbf{S}'_{a2}$ and $\mathbf{S}'_{a3}$ can form a 2R spherical subchain when screw $\mathbf{S}'_{a1}$ is transformed into screw $\mathbf{S}''_{a1}$, whose axis is along the X-axis in the local coordinate frame $O_a - XYZ$. Similarly, screws $\mathbf{S}''_{a2}$ and $\mathbf{S}''_{a3}$ can be treated as two successive revolute joints that are not a 2R spherical subchain.

In addition, the linear combination of screws $\mathbf{S}_{a1}$, $\mathbf{S}_{a4}$, and $\mathbf{S}_{a5}$ produces a new screw $\mathbf{S}''_{a1}$. It should be noted that screw $\mathbf{S}_{a2}$ cannot be used to combine with screw $\mathbf{S}_{a4}$ to avoid the presence of helical pairs. Similarly, $\mathbf{S}_{a3}$ cannot be used to combine with the screw $\mathbf{S}_{a5}$. Then, the new screw $\mathbf{S}''_{a2}$ can be transformed by the linear combination of $\mathbf{S}_{a2}$ and $\mathbf{S}_{a5}$, and the new screw $\mathbf{S}''_{a3}$ can be obtained using the linear combination of $\mathbf{S}_{a3}$ and $\mathbf{S}_{a4}$. The two screws $\mathbf{S}'_{a4}$ and $\mathbf{S}'_{a5}$ can be obtained after the linear combination of screws $\mathbf{S}_{a4}$ and $\mathbf{S}_{a5}$, which can be expressed as

$${}^2\mathbf{S}'_a = \begin{cases} \mathbf{S}'_{a4} = (0, 0, 0, 0, m_4, n_4)^T \\ \mathbf{S}'_{a5} = (0, 0, 0, 0, m_5, n_5)^T \end{cases} \quad (5)$$

The formulas exhibit two prismatic joints whose translation directions parallel the plane determined by the Y- and Z-axes. The screws $\mathbf{S}_{a4}$ and $\mathbf{S}_{a5}$ can also be used to construct two screws $\mathbf{S}''_{a4}$ and $\mathbf{S}''_{a5}$ by linear combination with screw $\mathbf{S}_{a2}$, as follows:

$${}^2\mathbf{S}''_a = \begin{cases} \mathbf{S}''_{a4} = (1, 0, 0, 0, b_4, c_4)^T \\ \mathbf{S}''_{a5} = (1, 0, 0, 0, b_5, c_5)^T \end{cases} \quad (6)$$

These new screws can be applied to design two revolute joints whose axes are along the X-axis. In addition, the formula indicates that there is no prismatic joint in the sub-mechanism.

It can also be found that the rotation axes of revolute joints are along the X-axis, intersect at one common point, or are based on the values presented in the previous equations. Furthermore, the number of revolute joints whose axes are along the X-axis cannot be greater than three because there are only three screws that can be used to obtain such revolute joints. Similarly, the number of other types of successive revolute joints is also not greater than three based on later equations.

Finally, the structural characteristics of the sub-mechanism in this case are:

1. There must be two or three successive revolute joints, i.e., $({}^i\mathbf{R}^j\mathbf{R})_N$, $({}^i\mathbf{R}^j\mathbf{R}^k\mathbf{R})_N$, ${}^{u1}\mathbf{R}{}^{u2}\mathbf{R}$ or ${}^{u1}\mathbf{R}{}^{u2}\mathbf{R}{}^{u3}\mathbf{R}$;
2. The revolute joint(s), except those in the successive revolute joints, must be ${}^i\mathbf{R}$;
3. The prismatic joint(s) must be ${}^v\mathbf{P}$ or ${}^{v1}\mathbf{P}{}^{v2}\mathbf{P}$.

The superscripts such as $i, j, u1, u2$ denote different spatial orientations of the joint axes; R stands for a revolute joint, P stands for a prismatic joint; and subscript N indicates successive revolute joints whose axes do not intersect at a common point. All screw expressions are unified into standard six-dimensional column vectors with consistent bold formatting.

An enumeration of such sub-mechanisms is provided in Table 2 based on these structural characteristics. These sub-mechanisms consist of two revolute joints forming a 2R spherical subchain, as well as three successive revolute joints that are ${}^x\mathbf{R}{}^x\mathbf{R}{}^x\mathbf{R}$.

3.3. Modeling Design of SRE Rehabilitation Model

Based on the biological configuration characteristics of the knee joint, a segmented redundant exoskeleton (SRE) was proposed to overcome the limitations of traditional rigid lower-limb exoskeletons, such as large volume, high energy consumption, and misalignment between robotic and human joints. After a comprehensive comparison of all candidate

topologies in Table 2 in terms of structural compactness, load capacity, and passive self-alignment performance, the final manufactured SRE mechanism adopts the configuration $xR^xR^{u1}R^{u2}R^{u3}R$ selected from the category without prioritization as the core structural layout. The proposed SRE integrates rigid and flexible components into a hybrid system, specifically designed to improve the efficacy of knee rehabilitation, as shown in Figure 3. The new integrated rehabilitation structure combines two benefits. On the one hand, the weight and volume of the wearable device are significantly reduced, thus ensuring a high degree of lower-limb comfort and movement for patients. On the other hand, this kind of structure can follow the motion of the knee joint of the patient and adapt to it; therefore, less mechanical interference is offered, the HRI response time is greatly reduced, and dangerous secondary injuries can be effectively prevented. The rigid part provides sufficient structural strength and load-bearing capacity, while the flexible parts can change with kinetic changes in movement. And this is the perfect combination that makes the rehabilitation support force more compliant with body functions.

Table 2. The enumeration of the sub-mechanism restricting translational mobility along the X-axis.

Enumeration	
Two prismatic pairs	$v^1Pv^2Pu^1Ru^2Ru^3R$, $u^1Ru^2Ru^3Rv^1Pv^2P$, $v^1Pv^2PxRu^1Ru^2R$, $v^1RxRv^2Pu^1Ru^2R$, $xRv^1Pv^2Pu^1Ru^2R$, $u^1Ru^2Rv^1Pv^2PxP$, $u^1Ru^2RxRv^1Pv^2P$, $u^1Ru^2Rv^1PxRv^2P$, $v^1Pv^2P(iR^jR^kR)_N$, $(iR^jR^kR)_N^v^1Pv^2P$, $v^1Pv^2PxP(iR^jR)_N$, $v^1PxRv^2P(iR^jR)_N$, $xRv^1Pv^2P(iR^jR)_N$, $(iR^jR)_N^v^1Pv^2PxR$, $(iR^jR)_N^xRv^1Pv^2P$, $(iR^jR)_N^v^1PxRv^2P$
One prismatic pair	$vPxRu^1Ru^2Ru^3R$, $xPvPu^1Ru^2Ru^3R$, $u^1Ru^2Ru^3RxPvP$, $u^1Ru^2Ru^3RvPxR$, $vPxRxRu^1Ru^2R$, $u^1Ru^2RxRxRvP$, $xRvPxRu^1Ru^2R$, $xRxRvPu^1Ru^2R$, $u^1Ru^2RxPvPxR$, $u^1Ru^2RvPxRxR$, $vPxRxP(iR^jR)_N$, $v^1PxR(iR^jR^kR)_N$, $xRvP(iR^jR^kR)_N$, $(iR^jR^kR)_N^xRvP$, $(iR^jR^kR)_N^vPxR$, $vPxRxR(iR^jR)_N$, $(iR^jR)_N^xRxRvP$, $xRvPxR(iR^jR)_N$, $xRxRvP(iR^jR)_N$, $(iR^jR)_N^xRvPxR$, $(iR^jR)_N^vPxRxR$
No prismatic pair	$xRxRu^1Ru^2Ru^3R$, $u^1Ru^2Ru^3RxRxR$, $xRxRxRu^1Ru^2R$, $xRxR(iR^jR^kR)_N$, $(iR^jR^kR)_N^xRxR$, $xRxRxR(iR^jR)_N$, $(iR^jR)_N^xRxRxR$

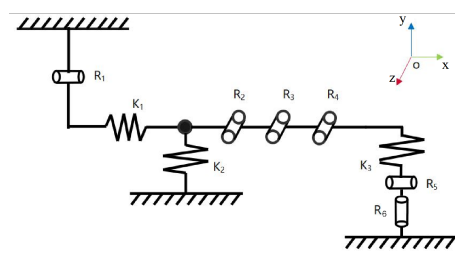


Figure 3. Establishment of the knee joint deviation system.

3.4. Adaptive Clutch Design for Precise Motion Control

An innovative clutch mechanism was designed to enable seamless knee flexion and extension, with its structural features optimized for compatibility with intelligent control algorithms, as shown in Figure 4. The first revolute joint of the mechanism is fixed to the thigh, while the last revolute joint is attached to the calf. Intermediate revolute joints (connected via revolute pairs) are actuated by a motor-driven string system.

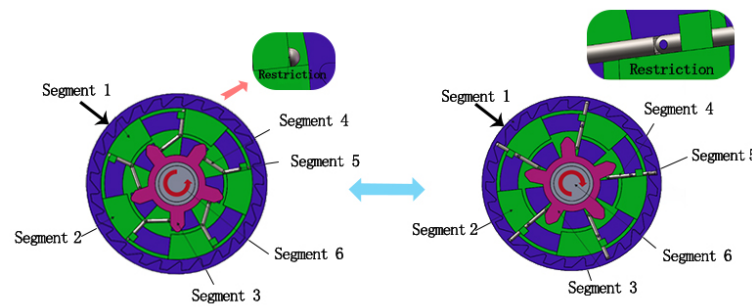


Figure 4. Design of the clutch system (The blue part indicates the outer shell, the green part denotes the passive rotation module, and the pink part corresponds to the active drive module.).

In particular, the primary load in both the standing and sitting postures comes from the lower leg, but the loading conditions are opposite: in standing, the leg is unloaded when extended and loaded when flexed; in sitting, the leg is unloaded when flexed and loaded when extended. The maximum rotation center offset also occurs with this posture. Therefore, a novel clutch mechanism capable of achieving maximum rotation angle and maximum torque can meet the requirements of the two aforementioned application scenarios.

The motor is mounted with flexible positioning to allow slight slippage along the thigh, adapting to individual anatomical variations and gait patterns. The motor mounted on the thigh can drive a rope pulled through the segments to make the limbs move, a clockwise turn brings about knee flexion, and a counterclockwise one triggers knee extension. In the design process, except for the first and last links, which are directly fixed to the lower extremities of the patients, all intermediate joint links can rotate freely and the natural movement of the leg is utilized. The motor shifting function can be used to realize muscle training in stroke patients using the healthy patella as a fulcrum point and keeping their legs closely aligned with the rotation axis of the knee joint through passive means. The connecting segment gradually tapers from thigh to calf, and in order to adapt to different thigh and calf sizes, an optimum fit and accurate transmission are needed. The accurate placement of the rotational center relative to the interface of the lower limb and the mechanism provides a solid foundation for the improvement and optimization of the control system. The general model of the knee rehabilitation mechanism is shown in Figure 5.

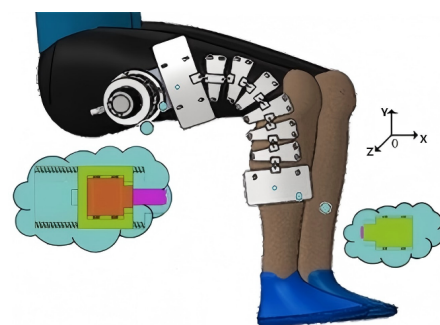


Figure 5. Model of the knee rehabilitation mechanism.

In particular, there are differences in the number of joints in different models. The conceptual model in Figure 3 uses six revolute joints to verify the feasibility of the principle in the initial design stage. When establishing the Adams simulation model with real human lower-limb dimensions, an additional revolute joint was added to better match the actual anatomical characteristics. In the kinematic and PSO optimization models, eight variables are considered, including two extra redundant revolute joints, to accommodate multi-directional rotational compensation during knee movement.

4. Kinematic Analysis

The proposed Segmented Redundant Exoskeleton (SRE) incorporates a multi-axis configuration. This design elevates human-machine adaptability; however, it also imposes substantial complexity on its kinematic model in Cartesian space. This complexity emerges as a crucial bottleneck for real-time motion planning and intelligent control, which are essential for next-generation rehabilitation robots. Therefore, simplifying the kinematic model while maintaining control precision is a fundamental requirement for the practical application of the SRE.

4.1. Forward Kinematic Modeling with Simplified D-H Parameters

Given that the SRE is a combination of successive rotational joints, the Denavit–Hartenberg parameters are applied to specify the coordinate system of each link in segment m , where $Z_{s,i-1}$ is the unit vector of the i_{th} rotational axis of segment m . The Denavit–Hartenberg parameters are listed in Table 3, where a_1 is 0 degrees due to its special position constraints.

Table 3. D-H parameters.

Joint i	a_i (mm)	α_i (°)	d_i (mm)	θ_i (°)
1	L	0	0	θ_1
2	L	0	0	θ_2
3	L	0	0	θ_3
4	L	0	0	θ_4
5	L	0	0	θ_5
6	L	0	0	θ_6
7	L	0	0	θ_7

A standard DH coordinate-frame schematic is supplemented here to clarify the coordinate definition rules. For the standard DH convention, the Z_{i-1} axis is aligned with the rotation axis of the i -th revolute joint, while the X_{i-1} axis lies along the common perpendicular between Z_{i-1} and Z_i . All eight connecting links are listed in the full DH parameters. The standard DH coordinate-frame diagram is shown in Figure 6.

The kinematic relationship of segment m is deduced on the basis of the D-H parameters as follows:

$$T_s = {}_1^{s,0}T_{s,1} {}_1^{s,1}T_{s,2} {}_1^{s,2}T_{s,3} \cdots {}_1^{s,5}T_{s,6} {}_1^{s,6}T_{s,7} {}_1^{s,7}T_{s,8} = f(\theta_{2s-1}, \theta_{2s}) \quad (7)$$

where ${}_1^{s,i-1}T_{s,i}$ denotes the homogeneous transformation matrix between the coordinate frames of adjacent links $i-1$ and i within segment s , $f_s(\theta_{2s-1}, \theta_{2s})$ represents the simplified forward kinematic function corresponding to segment s , and θ_{2s-1} and θ_{2s} are the joint angles of the segments.

According to the forward kinematic analysis, the precise attitude of the end effector can be calculated for the SRE. To further decrease the computational complexity for real-time control, the rotation matrix of the end effector is expressed using Euler angles as follows:

$$f_{\text{SRE}} = {}^0T_8 = \begin{bmatrix} \cos \theta_{123\dots 7} & 0 & \sin \theta_{123\dots 7} & 0 \\ 0 & 1 & 0 & 0 \\ -\sin \theta_{123\dots 7} & 0 & \cos \theta_{123\dots 7} & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (8)$$

where $\sin \theta_{123\dots 7} = \sin(\theta_1 + \theta_2 + \theta_3 + \cdots + \theta_7)$, $\cos \theta_{123\dots 7} = \cos(\theta_1 + \theta_2 + \theta_3 + \cdots + \theta_7)$, $f_{\text{SRE}}(\theta)$ is the simplified forward kinematic equation of the SRE, and θ is a column vector combining $\theta_1 \sim \theta_7$.

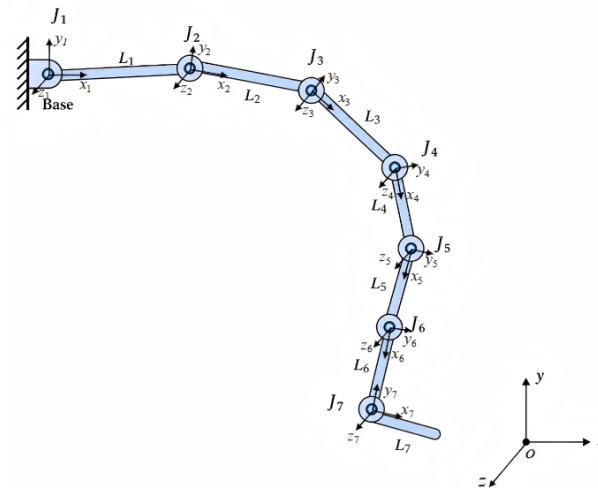


Figure 6. The standard DH coordinate-frame diagram.

4.2. Inverse Kinematic Solution Based on Simplified Jacobian Matrix

Due to the multiple actual rotational axes of the SRE, the Jacobian matrix derived by traditional methods would have an excessively high dimensionality, leading to a significant increase in computational load, and thus impedes real-time control responsiveness. To address this issue, the Jacobian matrix was simplified to derive an efficient inverse kinematic solution, with the aim of balancing calculation speed and control accuracy. First, the linear velocity v and the angular velocity ω of the SRE's linkage segment m are expressed as follows:

$$\begin{bmatrix} v_1 \\ w_1 \end{bmatrix} = \begin{bmatrix} e_{s,1}^\times r_{s,1} \dot{\theta}_{2s-1} + e_{s,2}^\times r_{s,2} \dot{\theta}_{2s-1} + \dots + e_{s,8}^\times r_{s,8} \dot{\theta}_{2s} \\ e_{s,1} \dot{\theta}_{2s-1} + e_{s,2} \dot{\theta}_{2s-1} + \dots + e_{s,7} \dot{\theta}_{2s-1} + e_{s,8} \dot{\theta}_{2s} \end{bmatrix} = \begin{bmatrix} v_{2s-1} & v_{2s} \\ w_{2s-1} & w_{2s} \end{bmatrix} \quad (9)$$

where $e_{s,i}^\times$ denotes the antisymmetric matrix of $e_{s,i}$ and also expresses the unit vector of the i_{th} axis, which includes the segment m and refers to the vector from the joint i_{th} to the end effector. Then, v_{2s-1} , v_{2s} , w_{2s-1} , and w_{2s} , respectively, represent the contributions of the two key joint variables of segment m to the linear velocity and angular velocity of the end effector, which are further decomposed as

$$\begin{cases} w_{2s-1} = e_{s,3} + e_{s,4} + \dots + e_{s,7} + e_{s,8} = R_{s-1} f_{w,2s-1} \\ w_{2s} = e_{s,1} + e_{s,2} + \dots + e_{s,5} + e_{s,6} = R_{s-1} f_{w,2s} \end{cases} \quad (10)$$

$$\begin{cases} v_{2s-1} = e_{s,3}^\times r_{s,3} + e_{s,4}^\times r_{s,4} + \dots + e_{s,7}^\times r_{s,7} + e_{s,8}^\times r_{s,8} \\ v_{2s} = e_{s,1}^\times r_{s,1} + e_{s,2}^\times r_{s,2} + \dots + e_{s,5}^\times r_{s,5} + e_{s,6}^\times r_{s,6} \end{cases} \quad (11)$$

where $e_{s,i}$ is the unit direction vector of the rotational axis of the i -th joint; $r_{s,i}$ stands for the position vector originating from the base coordinate origin to the central position of the i -th joint; and R_{s-1} refers to the orientation rotation matrix of the preceding segment $s-1$.

From the equations, R_{s-1} is the attitude matrix of the segment $m-1$, and $f_{w,2s-1}$ and $f_{w,2s}$ are the simplified angular velocity functions of the segment s , expressed as

$$f_{w,2s-1} = {}_1^{s-1}e_{s,3} + {}_1^{s-1}e_{s,4} + {}_1^{s-1}e_{s,7} + {}_1^{s-1}e_{s,8} = f_{w,2s-1}(\theta_{2s-1}, \theta_{2s}) \quad (12)$$

$$f_{w,2s} = {}_1^{s-1}e_{s,1} + {}_1^{s-1}e_{s,2} + {}_1^{s-1}e_{s,5} + {}_1^{s-1}e_{s,6} = f_{w,2s}(\theta_{2s-1}, \theta_{2s}) \quad (13)$$

To further simplify the linear velocity expression, the position vector $r_{s,i}$ (from the base to the i -th joint center of the segment m) is derived from the segment position analysis as in Figure 7.

$$r_{s,i} = -r_{Os,i} - P_{s-1} + P_e \quad (14)$$

where v expresses the vector matrix from the base to the i_{th} joint center of the segment m . Thus, the equation can be reformulated as

$$v_{2s-1} = R_{s-1} \left(f_{v1,2s-1} R_{s-1}^{-1} (P_e - P_{s-1}) - f_{v2,2s-1} \right) \quad (15)$$

where R_{s-1} , P_{s-1} and P_e can be obtained from the simplified forward kinematics from before $f_{v1,2s-1}$ and $f_{v2,2s-1}$ can be defined as

$$f_{v1,2s} = e_{s,1}^{\times} + \dots + e_{s,6}^{\times} = f_{v1,2s}(\theta_{2s-1}, \theta_{2s}) \quad (16)$$

$$f_{v2,2s} = {}_1^{s-1}e_{s,11}^{\times} {}_1^{s-1}r_{Os,1} + \dots + {}_1^{s-1}e_{s,61}^{\times} {}_1^{s-1}r_{Os,6} = f_{v2,2s}(\theta_{2s-1}, \theta_{2s}) \quad (17)$$

where v_{2s-1} , v_{2s} , ω_{2s-1} and ω_{2s} can then be obtained using simplified forward kinematics. The equations can be further rewritten as follows according to the Jacobian definition:

$$J_s \dot{\theta}_s = \begin{bmatrix} v_s \\ w_s \end{bmatrix} = \begin{bmatrix} v_{2s-1} & v_{2s} \\ w_{2s-1} & w_{2s} \end{bmatrix} \begin{bmatrix} \dot{\theta}_{2s-1} \\ \dot{\theta}_{2s} \end{bmatrix} \quad (18)$$

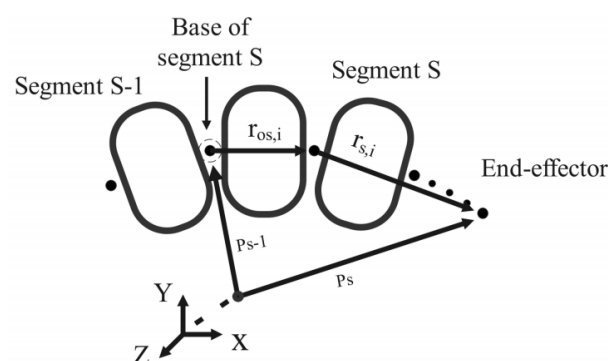


Figure 7. Position analysis of the segments.

Thus, the simplified Jacobian matrix of segment m is

$$J_s = \begin{bmatrix} v_{2s-1} & v_{2s} \\ w_{2s-1} & w_{2s} \end{bmatrix} \quad (19)$$

Similarly, the overall simplified Jacobian matrix of the SRE is further integrated as

$$J = \begin{bmatrix} v_1 & v_2 & \dots & v_8 \\ w_1 & w_2 & \dots & w_8 \end{bmatrix}^{6 \times 8} \quad (20)$$

Finally, the inverse kinematics of SRE can be obtained by iterating the Jacobian matrix, which reduces the complexity of the calculation process and shortens the time of the control algorithm. SRE has 8 joint variables and fewer task DoFs, resulting in some redundant DoFs. The PSO algorithm is adopted to resolve redundancy, minimizing joint torque and avoiding singular configurations. These results provide the core evaluation basis for the subsequent PSO algorithm. Importantly, redundant joints are inherent in the kinematic model, and the PSO algorithm is specifically utilized to correct and optimize these redundant degrees of freedom, ensuring stable and rational motion of the mechanism.

5. Simulation and Optimization Analysis

Based on the simplified kinematic model of the Segmented Redundant Exoskeleton (SRE), Adams simulation software was utilized to develop the SRE model to evaluate its mechanical performance under realistic rehabilitation scenarios. To balance computational efficiency with the retention of the core mechanical characteristics of the SRE, only seven revolute joints per SRE segment within the Adams model were configured, consistent with the calculated kinematic simplification formulas.

The SRE model incorporates real gravity, contact conditions, and elastic conditions in the simulation software. The load in the end effector of the model corresponds to the weight of a single lower limb for a healthy adult, simulating the actual weight load experienced during human movement. Then, the torsion spring assembly is installed to evenly distribute the driving force on both sides of the operating point, enabling transmission of force. Similarly, the parameters of the torsion springs previously set in the rotation joint are adjusted to mitigate impacts of collision during motion. Finally, pulleys are laterally installed on the module to guide the string's trajectory.

Under the specified loading conditions, the simulation results revealed a distinct “S-shaped” motion trajectory of the SRE mechanism, as shown in Figure 8, representing a non-ideal deformation that deviates from the kinematics of the natural knee joint. This figure further depicts the posture of each revolute joint during non-ideal motion toward the target trajectory. This “S-shaped” deformation is mainly due to two factors: (1) ropes connected to the distal end effector through steering levers generate a higher torque on the distal segment of the mechanism relative to the proximal segments; (2) the combined distal traction and the flexibility of multiple intermediate revolute joints induce a non-linear structural deformation.

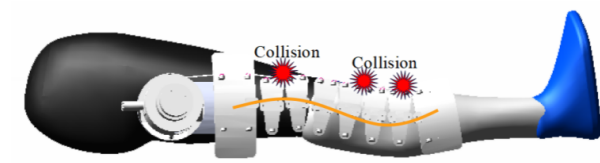


Figure 8. Failure “S-shaped” curve of the SRE model.

This deformation compromises motion accuracy. The optimized trajectory reduces deviation and shows potential clinical significance in improving knee coordination. To quantify collision-induced deviations, the rotational angles varying in time with each revolute joint prior to optimization are analyzed in Figure 9. Inflection points correspond to collision points, initially manifesting in distal segments and progressively propagating proximally.

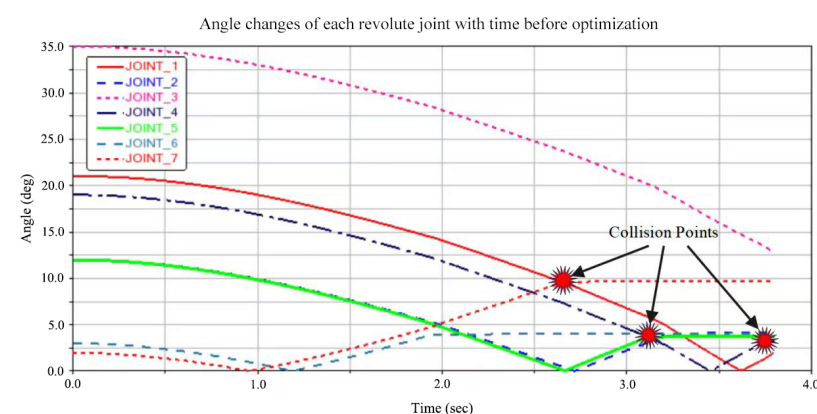


Figure 9. Angle changes in each revolute joint with time before optimization.

Manual adjustment of the torsion spring parameters to mitigate the “S-shaped” curve is inefficient and impedes rapid identification of optimal configurations, presenting critical limitations for the application of SRE in practical intelligent rehabilitation control systems. To minimize “S-shaped” deformation and reduce its curvature—thereby aligning SRE motion with the natural biomechanics of the knee biomechanics—the particle swarm optimization (PSO) algorithm was deployed within MATLAB R2023b to effectively optimize the torsion spring parameters. This data-driven optimization approach ensures that the SRE’s mechanical performance is tailored to real-time control requirements, a key design criterion for rehabilitation robots.

6. Motion Trajectory Optimization

6.1. Particle Swarm Optimization Algorithm

Based on the previous theoretical analysis, when no external load is applied to the SRE mechanism, the optimal solution is achieved when each rotational axis rotates at the same angle. However, under the influence of the load and the gravitational-magnetic field, it becomes necessary to employ the particle swarm optimization (PSO) algorithm to address this issue.

PSO is a highly effective intelligent optimization algorithm that discovers optimal solutions by iteratively configuring them based on both each particle’s individual best values and the swarm’s overall best value. Using an initial random solution, the algorithm seeks the optimal configuration that achieves the desired objective through an objective function. Additionally, PSO has a distinctive memory mechanism that greatly improves optimization efficiency, which makes it stand out from the alternatives.

Then, we can translate the base coordinate system to the working coordinate system of the robot, and this translation formula can be expressed as follows:

$$P_b = RP_t + T \quad (21)$$

where P_b represents the coordinate of the point P in the base coordinates of the robot, which can change with robot position changes; P_t represents the coordinate of point P in the working coordinates of the robot, which is not affected by the posture of the robot end effector; R represents the rotation matrix from the working coordinates to the base coordinates; and T represents the column vector from the working coordinates to the base coordinates. R and T were calculated in the kinematic analysis before.

The most important advantage of the PSO algorithm is the application of velocity update and position update formulas, which can be expressed as follows:

$$v_i = \omega \cdot v_i + c_1 \cdot r_1 \cdot (p_{best.i} - x_i) + c_2 \cdot r_2 (g_{best.i} - x_i) \quad (22)$$

where i is a particle; ω is the inertia factor; c_1 and c_2 represent different acceleration coefficients; and r_1 and r_2 represent two independent random numbers generated in the interval $[0, 1]$.

The key PSO parameters are stated as follows: swarm size = 50; maximum iterations = 300; inertia weight $\omega = 0.7$; and acceleration coefficients $c_1 = c_2 = 2.0$. The explicit objective function for minimizing S-shaped deformation is defined as

$$F = \frac{180}{\pi} \cdot \text{abs}(\theta_{\text{actual}} - \theta_{\text{ideal}})$$

where θ_{actual} is the actual joint angle and θ_{ideal} is the ideal straight-line trajectory.

From this formula, the i_{th} particle is directed toward the locally optimal individual particle p_{best} and the globally optimal individual particle g_{best} .

$$x_i = x_i + \lambda \cdot v_i \quad (23)$$

Here, v_i represents the current velocity of the particle and λ represents the applied time length.

In the process of the PSO algorithm, with the information of the best global point and the best local point provided to each particle, the PSO convergence speed can be rapidly improved, thus greatly improving the optimization efficiency.

Therefore, the configured position, speed, and fitness values were taken as the reference values, and the difference between the angles of each joint was reduced as much as possible. Multiple iterations were performed to find the optimal solution (we set eight joints in our simulation). Thus, we set the objective function as follows:

$$F = (\text{ABS}(\text{AX}(\text{MAR A8}, \text{MAR A0}))) \times 180 / \text{PI} \quad (24)$$

We define three core optimization variables: torsion spring stiffness k , ranging from 0.5 to 2 Nm/deg; preload angle θ_0 within 0° to 30° ; and installation position x varying from 5 to 20 mm. In Equation (24), AX stands for angle error, while MAR A8 and MAR A0 represent actual and reference joint angles, respectively. That is, the closer the shape of the mechanism is to a horizontal state after the operation, the closer it is to the real state of the patient's lower limbs. The $\text{Pi}/2$ radians are included in this equation because the rotation angle is opposite to the specified positive direction.

After setting up the objective function, we co-simulated the model in Adams and the algorithm in MATLAB for optimization. Test points were added to each revolute joint during the experiment, and torsion springs with different attributes and their motion curves were tested at the test points. After hundreds of parameter iterations, we obtained the convergence model diagram of each revolute joint, as shown in Figure 10. In addition, the minimum angle convergence diagram and the optimized model attitude diagram are shown in Figure 11.

By comparing the optimization results, we can intuitively see the influence of multiple groups of experiments on the 'S-shaped' curve through the test results. The minimum convergence angle diagram shows this after hundreds of iterations. The result shows that the total error of the seven revolute joints in the horizontal direction is only 3.104 degrees. This shows that the deviation rate of the overall SRE mechanism is less than 2.5% close to 120 degrees in the test, which shows the superiority of this optimization result.

6.2. Results and Analysis

From the figure, the end effector motion during the experimental test fell within the expected range, with no significant deviation from the leg's skeletal rotation angle of an average adult in a gait cycle. Additionally, we conducted a detailed comparison of the motion curves before and after optimization, as illustrated in Figure 12.

This figure presents a difference analysis of the objective function after performing both the actual simulation and theoretical analysis of the mechanism in three cases. Normally, when the end effector of the SRE mechanism is aligned with the same angle as the patient's knee, the rotational angle of each joint should be the same as the others. However, due to various factors, including gravity, load, and spring torque, deviations occur in the center of mass of the seven revolute joints. The red curve shows the difference between the mechanism joints and the objective function without optimization, the blue curve represents the difference between an arbitrary reference value selected during optimization, and the pink curve corresponds to the difference compared to the optimal solution.

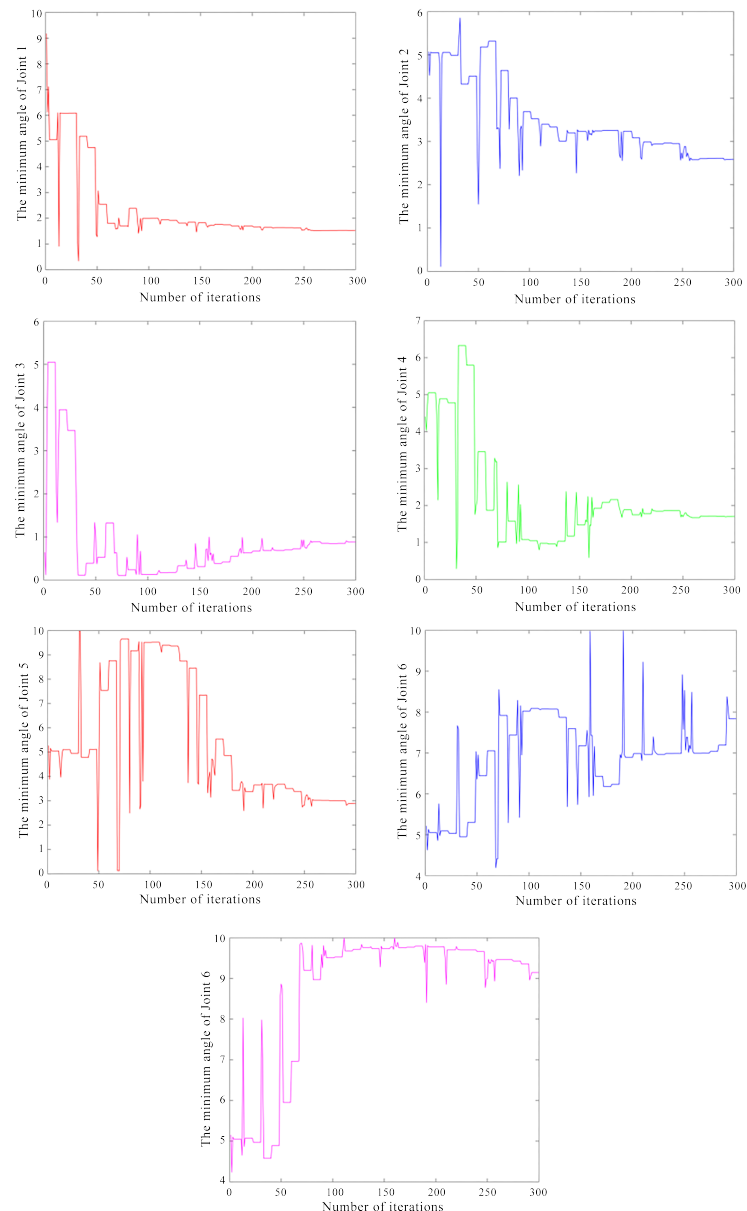


Figure 10. Convergence of minimum angles of 7 separate revolute joints after 300 iterations.

The specific parameters for the seven revolute joints were 1.526 radians, 2.586 radians, 0.880 radians, 1.701 radians, 2.888 radians, 7.838 radians and 9.148 radians, respectively. After optimization, the simulation in the target state revealed a minimum value after 3.433 s, with only one objective function: a 0.0535-radian difference from the objective function. This optimization significantly improves the accuracy of the mechanism, validating the feasibility of the optimization method used.

$$Deviation = |\theta_{\text{actual}} - \theta_{\text{target}}| \quad (25)$$

$$MotionAccuracy = 100\% - Deviation \quad (26)$$

Furthermore, the optimized SRE model achieved an accuracy of 97.5% (as calculation criteria, the deviation equals the absolute value of the difference between the actual angle and the target angle, and the accuracy of motion is calculated by subtracting the deviation ratio from 100%), which underscores the robustness and precision of the system. These

results validate the feasibility of the proposed mechanism for practical use, especially in the rehabilitation of stroke patients. This result suggests that the mechanism can not only adapt to the unique needs of individual patients, improving the potential for improved functional recovery, but can also effectively replicate natural leg movements while reducing the risk of secondary injuries, thus providing meaningful assistance in the rehabilitation process.

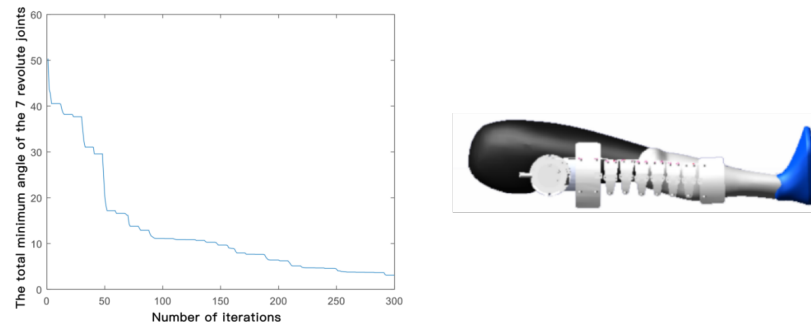


Figure 11. Convergence of minimum angles of SRE mechanism and optimized model.

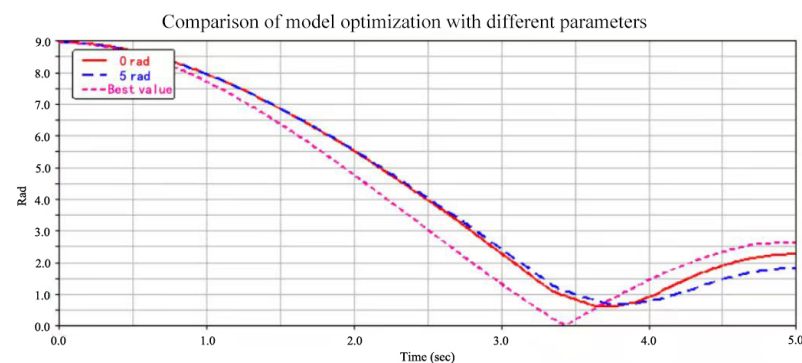


Figure 12. Comparison of model optimization with different parameters.

The design and kinematic simulations of the SRE model, coupled with the Adams and MATLAB co-simulation results, demonstrate that the SRE mechanism effectively facilitates flexible and coordinated movements of the lower limb. The optimized SRE improves trajectory fitting performance against experimentally captured human knee motion in a simulation environment; quantitative calculation of the instantaneous rotation center and deviation of the human–machine center will be realized via physical prototype tests in future research. This effectively addresses the challenge of imperfect human–machine coordination in the field of rehabilitation robotics.

6.3. Experimental Tests

By analyzing the optimized results and data from the human lower extremity, we evaluated the stress distribution within the continuum and derived equations to calculate the optimal parameters for the double-torsion helical spring.

Moreover, Figure 13 shows a diagram that illustrates the continuum model of the knee joint, offering a clear overview of the structure and components of the mechanism.

During the rotation, the force formula can be obtained from the final segment as follows at any time:

$$\begin{cases} F_{T,y} - F_{se,y} - m_e g - F_{load} = m_e a_y \\ F_{T,x} - F_{se,x} = m_e a_x \\ \tan \theta = \frac{F_{T,y}}{F_{T,x}} \end{cases} \quad (27)$$

Then, the moment formula of the final segment can also be found as follows:

$$F_T l_{0,e} - M_e - m_e g l_{G,e} - F_{load} l_{load,e} = 0 \quad (28)$$

Here, $F_{T,x}$ and $F_{T,y}$ are the torsion forces, which come from the motor, and their directions are along the x and y axis. $F_{S_e,x}$ and $F_{S_e,y}$ are the forces that come from the end torsion spring, and their direction is also along the x and y axis. $m_e a_x$ and $m_e a_y$ are the combination of the mass of the end segment and the acceleration velocity along with the axes x and y . Next, $l_{0,e}$ means the distance between the base and the end segment, $l_{G,e}$ means the distance between the center of mass and the center of rotational joint, and $l_{load,e}$ means the distance from the point of loading force to the center of rotational joint. Meanwhile, the rotational angle of the end torsion spring is θ_e , and g is the acceleration of gravity in the global environment.

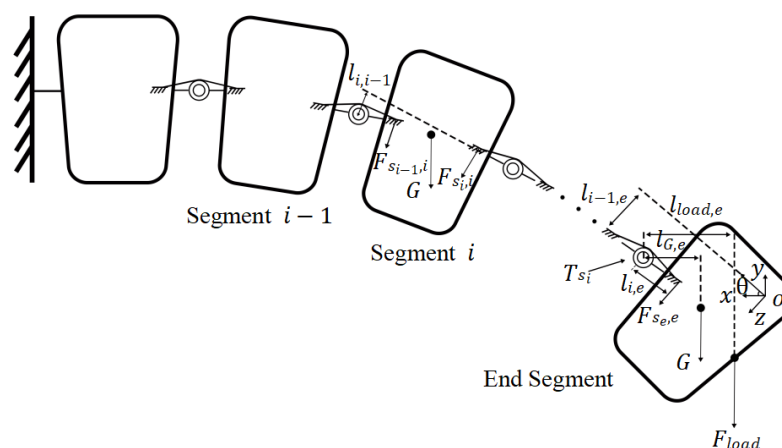


Figure 13. Diagram of the testing model of the knee joint.

In the same way, the force formula can be obtained from the arbitrary segment i as follows at any time while revolute:

$$\begin{cases} F_{i,y} - F_{i-1,y} - m_{i-1}g = m_{i-1}a_y \\ F_{i,x} - F_{i-1,x} = m_{i-1}a_x \\ \tan \theta_{i,i-1} = \frac{F_{i,y}}{F_{i,x}} \end{cases} \quad (29)$$

Next, the moment formula of the arbitrarily segment i could be found:

$$M_i = M_{i-1} - l_{i,i-1} \quad (30)$$

where, $F_{i,x}$ and $F_{i,y}$ are the forces that come from the i_{th} torsion spring on the i_{th} segment, along with the x and y axes. F_{i-1} is the force that comes from the $(i-1)_{th}$ torsion spring on the i_{th} segment in this formula. The rotational angle $\theta_{i,i-1}$ defines the angle between the $(i-1)_{th}$ torsion spring and the i_{th} torsion spring during this time.

Finally, the overall force equation of the whole mechanism can be calculated as

$$\begin{cases} F_{T,y} - \sum F_{s,y} - \sum mg - F_{load} = \sum m_{a,y} \\ F_{T,x} - \sum F_{s,x} = \sum m_{a,x} \\ \tan \theta = \frac{F_{T,y}}{F_{T,x}} \end{cases} \quad (31)$$

The overall moment equation of the whole mechanism can be obtained as

$$F_T l_{0,e} = M_1 + \dots + M_e + m_e g l_{0,e} + \dots + m_1 g l_{0,1} \quad (32)$$

where $\sum F_{s,x}$ and $\sum F_{s,y}$ are the total spring forces along with the directions of the x and y axes, and $\sum mg$ is the gravity of the segments. M_i indicates the moment of the i_{th} torsion spring, while M_e indicates the moment of the end segment. $l_{0,i}$ is the distance between the base and the forcing point of the i_{th} torsion spring. Normally, the mass of each torsion spring is the same value.

The content provided allows for the theoretical computation of the required torque that various locations on a torsion spring need to bear. To produce appropriate torsion springs, the dimensions of the spring would be determined based on the established principles of torsion spring design and the actual manufacturing capabilities of the factory. It is important to consider any necessary modifications or adjustments that may be required to ensure that the final product meets the desired specifications. This attention to detail guarantees that our manufactured torsion springs will be of the highest quality and meet all necessary specifications.

The force resulting from the torsion spring is

$$F = \frac{(Ed^4)\varphi}{3670 \times nDL} \quad (33)$$

where E is the modulus of elasticity, d is the diameter of the torsion spring, n is the number of circles, and D is the mean diameter, which is the average value of the outside diameter D_o and the inside diameter D_i . L is the length of the feet, and the angle φ here means the passive rotational angle of the total torsion spring. The contents of the parameters are shown in the Figure 14.

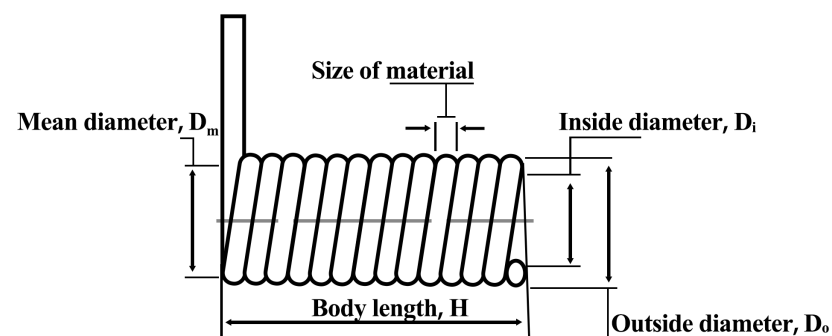


Figure 14. The structural features of the torsion spring.

Taking into account the loading force, motor output and mechanism dimensions, we determined the key parameters for the torsion springs, as shown in Table 4, including material, foot length, and diameters. The design equations account for the material properties, the diameter of the wire, and the number of coils to ensure optimal performance and reliability. This approach led to the successful design of a torsion spring that met all the performance criteria.

Table 4. Torsion spring parameters.

Material	Wire Diameter	Mean Diameter	Coil(s)	Stiffness
65 Mn	2 mm	10 mm	8	1.2 Nm/deg

Understanding the stress distribution and optimizing the design parameters are crucial to achieving superior spring performance and durability in various applications, such as automotive, aerospace, and medical devices. The detailed test model is shown in Figure 15.



Figure 15. Different states of the testing model.

This device offers two recovery modes: knee extension during seated rehabilitation and knee flexion during standing, aligned with the patient's recovery stage. To facilitate movement of the knee joint during rehabilitation, the design includes two channels for steel wire ropes. The torsion spring, pre-curved to mimic a bent knee, uses a clutch to extend the upper rope, straightening the knee for seated rehabilitation. In the second phase, the spring posture is adjusted to straighten the mechanism and the clutch contracts the lower rope to allow knee flexion, aiding later-stage rehabilitation when the patient can walk independently.

7. Discussion

7.1. Advantages

Compared with previously reported self-alignment knee exoskeletons, the proposed mechanism exhibits three distinct improvements. First, it realizes three-dimensional passive self-alignment to accommodate the spatial offset of the knee rotation center. Second, the hybrid rigid–flexible structure effectively suppresses non-ideal S-shaped deformation during movement. Third, the PSO-based parameter optimization method improves motion accuracy by adjusting joint mechanical parameters. These enhancements address the issues of misalignment and structural deformation commonly encountered in existing designs.

7.2. Limitations

This study has several limitations that should be noted. First, all results are derived solely from numerical simulation analysis, and no human-subject experiments or clinical trials were performed. Therefore, the mechanical performance and rehabilitation effectiveness of the SRE were not verified under real physiological conditions, which limits the reliability and generalizability of the current findings. Second, the motion capture database is constructed from a small group of young, healthy participants with relatively homogeneous physical characteristics. The data do not fully represent the kinematic variability of different age groups, body types, or pathological populations, which may affect the adaptability of the proposed mechanism in practical applications. The lack of a multi-joint kinematic analysis limits a comprehensive evaluation of the general motion compatibility of the exoskeleton. Third, current research focuses only on the kinematic characteristics of the knee joint in isolation, without considering the coordinated motion of the hip and ankle joints during natural human movement.

7.3. Future Work

Future work will focus on addressing these limitations. First, human-subject trials with healthy participants will be conducted to verify the mechanical performance and motion compatibility of the SRE under real physiological conditions. Second, further validation experiments will be carried out in stroke patients to evaluate the practical rehabilitation effects and clinical applicability of the mechanism. Third, multi-joint kinematic analysis will be integrated to explore the coordinated motion characteristics of the lower limb, improving the overall adaptability of the exoskeleton. In addition, human–robot interaction and operational safety require further analysis to ensure stable rehabilitation performance. Intelligent adaptive control strategies will also be developed to improve the safety and effectiveness of the rehabilitation training.

8. Conclusions

This paper introduces a novel mechanism design, kinematic analysis, and motion optimization scheme for a novel rehabilitation exoskeleton. The exoskeleton uniquely integrates active and passive clutch drive mechanisms, achieving exceptional rigidity, substantial loading capacity, and smooth motion characteristics through a synergistic blend of rigid and flexible components. Subsequently, the particle swarm optimization algorithm (PSO) was used to meticulously optimize the torsion spring parameters at the revolute joints. Under simulated working conditions, this optimization effectively improved the trajectory-fitting performance between the SRE mechanism and the measured human knee motion, and optimized the passive rotation center tracking effect; further quantitative verification with physical prototypes will be carried out in follow-up research. The mechanism also features modularity and customization, enabling future adaptable adjustments to torsion spring parameters and component quantities according to individual patient requirements. In addition, the integration of sophisticated motion principles and responsive feedback mechanisms elevates the entire rehabilitation process.

In summary, this study offers a simulation-verified design scheme for knee rehabilitation exoskeletons, with clinical validation scheduled for future research. Although exoskeleton technology remains in its early development stage and faces challenges in technical implementation and clinical application, the effectiveness of the system promises substantial improvement through subsequent research. Future work will incorporate advanced sensors, refined control logic, and even AI algorithms, combined with expanded clinical trial cases.

Author Contributions: Conceptualization, M.G.; methodology, D.J. and F.Z.; software, H.W. and M.G.; validation, M.G., D.J. and Y.W.; formal analysis, M.G.; investigation, W.Z. and F.Z.; resources, Y.W.; data curation, M.G.; writing—original draft preparation, M.G.; writing—review and editing, W.F.; visualization, W.F.; supervision, H.W.; project administration, H.W. and Y.W.; funding acquisition, H.W. All authors have read and agreed to the published version of the manuscript.

Funding: This research was funded by the Basic Research Project of China North Industries Group Corporation (Grant No. K23F048) and the National Natural Science Foundation of China (NSFC) (Grant No. 52405006).

Institutional Review Board Statement: Not applicable.

Informed Consent Statement: Not applicable.

Data Availability Statement: The data that support the findings of this study are available from the corresponding authors upon reasonable request.

Acknowledgments: During the preparation of this study, the authors used a motion capture system to obtain rotational data of the normal human lower limb during movement. The authors have reviewed and edited the output and take full responsibility for the content of this publication.

Conflicts of Interest: Author Wentao Feng was employed by the company Beijing North Vehicle Group Corporation. The remaining authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

Abbreviations

The following abbreviations are used in this manuscript:

HRI	Human-Robot Interaction
PSO	Particle Swarm Optimization
SRE	Segmented Redundant Exoskeleton
ROM	Range of Motion

References

1. Xiao, F.; Li, G.; Jiang, D.; Xie, Y.; Yun, J.; Liu, Y.; Huang, L.; Fang, Z. An effective and unified method to derive the inverse kinematics formulas of general six-DOF manipulator with simple geometry. *Mech. Mach. Theory* **2021**, *159*, 104265. [\[CrossRef\]](#)
2. Yang, C.; Kang, R.; Branson, D.T.; Chen, L.; Dai, J.S. Kinematics and statics of eccentric soft bending actuators with external payloads. *Mech. Mach. Theory* **2019**, *139*, 526–541. [\[CrossRef\]](#)
3. Obregón-Flores, J.; Arechavaleta, G.; Becerra, H.M.; Morales-Díaz, A. Predefined-time robust hierarchical inverse dynamics on torque-controlled redundant manipulators. *IEEE Trans. Robot.* **2021**, *37*, 962–978.
4. Alabdulkarim, S.; Nussbaum, M.A. Influences of different exoskeleton designs and tool mass on physical demands and performance in a simulated overhead drilling task. *Appl. Ergon.* **2019**, *74*, 55–66. [\[CrossRef\]](#) [\[PubMed\]](#)
5. Amiri, M.S.; Ramli, R.; Ibrahim, M.F. Hybrid design of PID controller for four DoF lower limb exoskeleton. *Appl. Math. Model.* **2019**, *72*, 17–27. [\[CrossRef\]](#)
6. Baltrusch, S.; Van Dieën, J.; Koopman, A.S.; Näf, M.; Rodriguez-Guerrero, C.; Babič, J.; Houdijk, H. SPEXOR passive spinal exoskeleton decreases metabolic cost during symmetric repetitive lifting. *Eur. J. Appl. Physiol.* **2020**, *120*, 401–412. [\[PubMed\]](#)
7. Bär, M.; Steinhilber, B.; Rieger, M.A.; Luger, T. The influence of using exoskeletons during occupational tasks on acute physical stress and strain compared to no exoskeleton—A systematic review and meta-analysis. *Appl. Ergon.* **2021**, *94*, 103385. [\[PubMed\]](#)
8. Yang, C.; Geng, S.; Walker, I.; Branson, D.T.; Liu, J.; Dai, J.S.; Kang, R. Geometric constraint-based modeling and analysis of a novel continuum robot with shape memory alloy initiated variable stiffness. *Int. J. Robot. Res.* **2020**, *39*, 1620–1634. [\[CrossRef\]](#)
9. Cao, W.; Chen, C.; Hu, H.; Fang, K.; Wu, X. Effect of hip assistance modes on metabolic cost of walking with a soft exoskeleton. *IEEE Trans. Autom. Sci. Eng.* **2020**, *18*, 426–436. [\[CrossRef\]](#)
10. Wei, G.; Dai, J.S.; Wang, S.; Luo, H. Kinematic analysis and prototype of a metamorphic anthropomorphic hand with a reconfigurable palm. *Int. J. Humanoid Robot.* **2011**, *8*, 459–479. [\[CrossRef\]](#)
11. Cha, J.S.; Monfared, S.; Stefanidis, D.; Nussbaum, M.A.; Yu, D. Supporting surgical teams: Identifying needs and barriers for exoskeleton implementation in the operating room. *Hum. Factors* **2020**, *62*, 377–390. [\[PubMed\]](#)
12. Chen, L.; Wang, H.; Zhang, Y.; Li, M.; Zhao, J.; Liu, X.; Yang, S.; Zhou, K.; Wu, Q.; Gao, T. Design of a Geriatric Lower Limb Exoskeleton. *J. Geriatr. Rehabil.* **2024**, *8*, 45–62.
13. de Vries, A.; Murphy, M.; Könnemann, R.; Kingma, I.; de Looze, M. The amount of support provided by a passive arm support exoskeleton in a range of elevated arm postures. *IIEE Trans. Occup. Ergon. Hum. Factors* **2019**, *7*, 311–321. [\[CrossRef\]](#)
14. Febrer-Nafria, M.; Pallarès-López, R.; Fregly, B.J.; Font-Llagunes, J.M. Comparison of different optimal control formulations for generating dynamically consistent crutch walking simulations using a torque-driven model. *Mech. Mach. Theory* **2020**, *154*, 104031. [\[CrossRef\]](#)
15. Guanziroli, E.; Cazzaniga, M.; Colombo, L.; Basilico, S.; Legnani, G.; Molteni, F. Assistive powered exoskeleton for complete spinal cord injury: Correlations between walking ability and exoskeleton control. *Eur. J. Phys. Rehabil. Med.* **2018**, *55*, 209–216. [\[PubMed\]](#)
16. Hayes, S.C.; White, M.; White, H.S.F.; Vanicek, N. A biomechanical comparison of powered robotic exoskeleton gait with normal and slow walking: An investigation with able-bodied individuals. *Clin. Biomech.* **2020**, *80*, 105133. [\[CrossRef\]](#)
17. Kashef, S.R.; Amini, S.; Akbarzadeh, A. Robotic hand: A review on linkage-driven finger mechanisms of prosthetic hands and evaluation of the performance criteria. *Mech. Mach. Theory* **2020**, *145*, 103677. [\[CrossRef\]](#)

18. Kim, S.; Moore, A.; Srinivasan, D.; Akanmu, A.; Barr, A.; Harris-Adamson, C.; Rempel, D.M.; Nussbaum, M.A. Potential of exoskeleton technologies to enhance safety, health, and performance in construction: Industry perspectives and future research directions. *IIEE Trans. Occup. Ergon. Hum. Factors* **2019**, *7*, 185–191. [\[CrossRef\]](#)
19. Li, Z.; Huang, B.; Ye, Z.; Deng, M.; Yang, C. Physical human–robot interaction of a robotic exoskeleton by admittance control. *IEEE Trans. Ind. Electron.* **2018**, *65*, 9614–9624. [\[CrossRef\]](#)
20. Wei, G.; Chen, Y.; Dai, J.S. Synthesis, mobility, and multifurcation of deployable polyhedral mechanisms with radially reciprocating motion. *J. Mech. Des.* **2014**, *136*, 091003. [\[CrossRef\]](#)
21. Li, Z.; Zuo, W.; Li, S. Zeroing dynamics method for motion control of industrial upper-limb exoskeleton system with minimal potential energy modulation. *Measurement* **2020**, *163*, 107964. [\[CrossRef\]](#)
22. McCain, E.M.; Dick, T.J.; Giest, T.N.; Nuckols, R.W.; Lewek, M.D.; Saul, K.R.; Sawicki, G.S. Mechanics and energetics of post-stroke walking aided by a powered ankle exoskeleton with speed-adaptive myoelectric control. *J. Neuroeng. Rehabil.* **2019**, *16*, 57. [\[PubMed\]](#)
23. Nikkhah, A.; Bradley, C.; Ahmadian, A.S. Design, dynamic modeling, control and implementation of hydraulic artificial muscles in an antagonistic pair configuration. *Mech. Mach. Theory* **2020**, *153*, 104007. [\[CrossRef\]](#)
24. Park, Y.; Jo, I.; Lee, J.; Bae, J. WeHAPTIC: A wearable haptic interface for accurate position tracking and interactive force control. *Mech. Mach. Theory* **2020**, *153*, 104005. [\[CrossRef\]](#)
25. Sarkisian, S.V.; Ishmael, M.K.; Lenzi, T. Self-aligning mechanism improves comfort and performance with a powered knee exoskeleton. *IEEE Trans. Neural Syst. Rehabil. Eng.* **2021**, *29*, 629–640. [\[CrossRef\]](#) [\[PubMed\]](#)
26. Shafiei, M.; Behzadipour, S. Adding backlash to the connection elements can improve the performance of a robotic exoskeleton. *Mech. Mach. Theory* **2020**, *152*, 103937. [\[CrossRef\]](#)
27. Qin, Y.; Zhang, K.; Li, J.; Dai, J.S. Modelling and analysis of a rigid–compliant parallel mechanism. *Robot. Comput.-Integr. Manuf.* **2013**, *29*, 33–40. [\[CrossRef\]](#)
28. Wang, T.; Spyrakos-Papastavridis, E.; Dai, J.S. Design and analysis of a novel reconfigurable ankle rehabilitation exoskeleton capable of matching the mobile biological joint center in real-time. *J. Mech. Robot.* **2023**, *15*, 011011.
29. Lai, J.; Song, A.; Lu, Y.; Wu, P.; Zhang, W.; Li, Y.; Wang, J.; Wu, T.; Wei, X.; Wang, H. Design, Modeling and Evaluation of Lattice Structure Pneumatic Actuators (LSPAs) for Soft Wrist Rehabilitation Robot. *IEEE Trans. Biomed. Eng.* **2025**, *73*, 579–590. [\[CrossRef\]](#)
30. Li, G.; Cheng, L.; Gao, Z.; Xia, X.; Jiang, J. Development of an Untethered Adaptive Thumb Exoskeleton for Delicate Rehabilitation Assistance. *IEEE Trans. Neural Syst. Rehabil. Eng.* **2022**, *38*, 3514–3529. [\[CrossRef\]](#)
31. Xie, C.; Lyu, Y.; Li, G.; Tong, R.K.; Xia, H.; Song, R.; Li, Z. A Cable-Driven Upper Limb Rehabilitation Robot With Muscle-Synergy-Based Myoelectric Controller. *IEEE Trans. Robot.* **2024**, *40*, 3199–3210.
32. Best, T.K.; Welker, C.G.; Rouse, E.J.; Gregg, R.D. Data-Driven Variable Impedance Control of a Powered Knee–Ankle Prosthesis for Adaptive Speed and Incline Walking. *IEEE Trans. Robot.* **2023**, *39*, 2151–2169. [\[PubMed\]](#)
33. Williams, J.R.; Cuddeback, C.F.; Fang, S.; Lerner, Z.F. Design and Validation of a Wearable Ankle Push-off Device in Cerebral Palsy: Is Spring Resistance as Effective as Motorized Resistance? *IEEE Trans. Biomed. Eng.* **2025**, *72*, 1124–1133. [\[CrossRef\]](#)
34. Dell’Eva, F.; Guanziroli, E.; Camerini, V.; Gandolla, M.; Brignole, L.; Maludrotto, S.; Gruppioni, E.; Ferrigno, G.; Molteni, F.; Ambrosini, E.; et al. A multifaceted hybrid ES-robotic device for gait training in individuals with neurological disorders. *Nat. Commun.* **2025**, *16*, 8510. [\[CrossRef\]](#) [\[PubMed\]](#)
35. Cha, J.M.; Hong, J.; Yoo, J.; Rha, D.W. Wearable Robots for Rehabilitation and Assistance of Gait: A Narrative Review. *Ann. Rehabil. Med.* **2025**, *49*, 187–195. [\[CrossRef\]](#) [\[PubMed\]](#)
36. Wang, Y.; Li, H.; Zhang, Q.; Liu, J.; Zhao, K.; Chen, M.; Zhou, L.; Sun, W.; Hu, X.; Xu, Z.; et al. Extended State Observer-Based Sliding Mode Control for Hip Exoskeleton Without Torque Sensors. *IEEE Trans. Neural Syst. Rehabil. Eng.* **2025**, *33*, 1245–1254.
37. Zhang, H.; Li, M.; Wang, T.; Liu, S.; Zhao, Y.; Chen, J.; Zhou, W.; Wu, L.; Xu, B.; Gao, K.; et al. Orthopedic Rehabilitation Exoskeleton. *Clin. Biomech.* **2023**, *38*, 215–228.
38. Li, G.; Wang, A.; Zhang, B.; Liu, C.; Zhao, D.; Chen, E.; Zhou, F.; Wu, G.; Xu, H.; Yang, I.; et al. Development and Validation of a Self-Aligning Knee Exoskeleton with Hip Rotation Capability. *IEEE Trans. Neural Syst. Rehabil. Eng.* **2024**, *32*, 472–481. [\[CrossRef\]](#) [\[PubMed\]](#)
39. Kim, J.; Park, S.; Lee, M.; Choi, H.; Jung, W.; Seo, Y.; Han, T.; Ahn, J.; Bae, K.; Yoon, S.; et al. Passive vs. Active Exoskeletons. *IEEE Rev. Biomed. Eng.* **2022**, *15*, 89–106.
40. Yang, X.; Guo, S.; Wang, P.; Wu, Y.; Niu, L.; Liu, D. Design and Optimization Analysis of an Adaptive Knee Exoskeleton. *Chin. J. Mech. Eng.* **2024**, *37*, 104. [\[CrossRef\]](#)
41. Liu, Z.; Han, J.; Han, J.; Zhang, J. Design and Evaluation of a Lightweight, Ligaments-Inspired Knee Exoskeleton for Walking Assistance. *IEEE Robot. Autom. Lett.* **2024**, *9*, 8491–8498. [\[CrossRef\]](#)
42. Kittisares, S.; Ide, T.; Nabae, H.; Suzumori, K. Ergonomic dual four-bar linkage knee exoskeleton for stair ascent assistance. *Front. Robot. AI* **2023**, *10*, 1285520. [\[CrossRef\]](#) [\[PubMed\]](#)

43. Song, H.Y.; Chung, H.J.; Jeon, K.W.; Kim, T.H.; Kim, J.; Yoo, J.Y.; Ryu, J.K. Optimal Placement of Foot Pressure Sensors for Lower-Limb Exoskeletons Based on Multi-Objective Particle Swarm Optimization Algorithm. In *2024 International Conference on Artificial Intelligence in Information and Communication (ICAIIIC)*; IEEE: Piscataway, NJ, USA, 2025.
44. Liu, T.; Liu, K.; Luo, W.; Kou, J.; Zhan, H.; Yu, G.; Guo, Q.; Shi, Y. Motion Gait Recognition of Lower Limb Exoskeleton Based on Particle Swarm Optimization-Based Extreme Learning Machine Algorithm. *Actuators* **2025**, *14*, 120. [[CrossRef](#)]
45. Li, C.; Wang, M.; Zhang, T.; Liu, H.; Zhao, R.; Chen, Y.; Zhou, J.; Wu, S.; Xu, L.; Yang, K.; et al. Real-time adaptive super twisting algorithm based on PSO algorithm: Application for an exoskeleton robot. *Robotica* **2024**, *42*, 2110–2126. [[CrossRef](#)]
46. Wang, Y.; Liu, H.; Zhang, S.; Li, M.; Zhao, Q.; Chen, J.; Zhou, X.; Wu, T.; Xu, R.; Yang, L.; et al. Rehabilitation exoskeleton torque control based on PSO-RBFNN optimization. *PLoS ONE* **2024**, *19*, e0285453. [[PubMed](#)]
47. Gautam, S.M.; Singla, E.; Singla, A. Optimal Kinematic Synthesis and Prototype Development of Knee Exoskeleton. *arXiv* **2024**, arXiv:2409.02635.

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.