

Kinematic and Rehabilitation Performance Analysis of a Reconfigurable Parallel Mechanism for Staged Ankle Rehabilitation

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1. Introduction

The ankle joint is one of the most frequently loaded and active joints in the human body and plays an important role in walking, running, jumping, and postural adjustment. Ankle rehabilitation has clear staged characteristics [1]. Early rehabilitation usually focuses on low-risk and easily controlled dorsiflexion/plantarflexion training. As tissue tolerance and neuromuscular control improve, eversion/inversion and multi-axis compound training can be gradually introduced. Therefore, an ankle rehabilitation robot should not only provide sufficient motion output, but also regulate the number and type of DOFs according to the rehabilitation stage.

Many studies have been carried out on parallel mechanisms for ankle rehabilitation. Girone et al. proposed the Rutgers Ankle based on a Stewart platform [2]. Bi Z. M. proposed an ankle rehabilitation robot based on a 3-UCU/S spherical parallel mechanism [3]. Yoon et al. developed a reconfigurable ankle rehabilitation robot to cover multiple training modes [4]. In China, Zou Y. et al. designed a 3-RRS spherical parallel ankle rehabilitation robot [5]. Jia W. H. et al. proposed a reconfigurable parallel ankle rehabilitation robot with two rehabilitation modes [6]. Zhang L. et al. proposed a dual-mode reconfigurable parallel ankle rehabilitation mechanism that can switch between 1T2R and 2T1R motion modes to satisfy the different motion requirements of ankle mobility training and proprioceptive training, respectively [7].

Although existing ankle rehabilitation mechanisms can realize single-axis or multi-axis motions, different mechanisms are generally designed for specific DOF configurations or limited training modes. When rehabilitation progresses from restricted single-axis training in the early stage to multi-axis coordinated training in the middle and late stages, multiple dedicated devices with different DOFs can separately satisfy the training requirements of each stage [8]. However, this increases the number of rehabilitation devices and the required treatment space. In addition, when patients are transferred between different devices, the foot must be fixed again and the anatomical ankle center must be realigned with the mechanism rotation center. Existing reconfigurable ankle rehabilitation mechanisms can reduce these problems to some extent, but most existing designs change their training modes by adding or removing functional modules, rearranging driving units, or switching among a limited number of configurations. Therefore, there remains further scope for developing a mechanism that can

successively realize one-, two-, and three-DOF staged rehabilitation training through progressive changes in constraints while keeping the main driving structure unchanged.

To address the above problems, a reconfigurable parallel mechanism for staged ankle rehabilitation is proposed in this paper. Unlike approaches that change the mechanism DOFs by replacing functional modules or rearranging multiple driving limbs, the proposed mechanism always uses three UPS limbs as the driving and load-bearing units, together with an independent central reconfigurable constraint chain as the DOF regulation unit. By changing the relative positions and locking states of the rotating axes inside the central joint, the joint can be made equivalent to R, U, and S joints, respectively. Thus, the moving platform can successively obtain 1R, 2R, and 3R pure rotational modes while the three UPS driving limbs and the main moving and fixed platforms remain unchanged. In essence, the rotational constraints imposed on the moving platform by the central constraint chain are progressively reduced, thereby gradually releasing the motion DOFs from low to high. Modes 1, 2, and 3 therefore correspond to high-constraint single-axis training in the early stage, restricted two-axis compound training in the middle stage, and three-axis coordinated training in the late stage, respectively. Therefore, the structural innovation of the proposed mechanism is not simply an increase in the number of training modes, but the establishment of a correspondence among the central constraint state, motion DOFs, and rehabilitation stages within a unified mechanism.

Furthermore, compared with a multi-DOF mechanism in which all DOFs remain open and non-target motions are suppressed entirely through active control, the proposed mechanism directly restricts the corresponding rotational DOFs through the central reconfigurable constraint chain, so that the motion subspace required at each rehabilitation stage is determined by the structural constraints of the mechanism itself. This not only reduces the number of posture variables that must be simultaneously regulated by the control system and decreases the dependence on active stabilization in non-target directions, but also provides structural-level motion constraints. When control abnormalities or deviations in drive commands occur, the mechanically constrained non-target DOFs remain physically restricted, thereby reducing the risk of unintended multi-axis motion and secondary injury to the patient to a certain extent. Therefore, the proposed reconfiguration method has the potential advantages of reducing control complexity and improving the inherent safety of restricted rehabilitation training.

From the perspective of rehabilitation motion function, a stage-specific rehabilitation mechanism is mainly required to provide the DOFs, training directions, and ankle motion ranges needed at the corresponding rehabilitation stage. The proposed mechanism covers these different stage-specific motion requirements on a single mechanical platform through three reconfigurable modes. Mode 1 retains only dorsiflexion/plantarflexion and can be used for low-risk single-axis training in the early rehabilitation stage. Mode 2 releases a second rotational DOF and can be used for restricted two-axis compound training in the middle stage. Mode 3 further releases all three rotational DOFs for multi-axis coordinated training in the late stage. The mobility analysis verifies that the three modes form the corresponding motion subspaces. The ADAMS simulations further show that the mechanism can continuously generate dorsiflexion/plantarflexion, inversion/eversion, and adduction/abduction motions in the target directions, while the AnyBody human-machine coupling analysis shows that different training directions induce direction-selective responses of the corresponding target muscle groups. Therefore, from the perspectives of motion-output capability and training-function coverage, the proposed single mechanism has the potential to undertake the motion tasks required at different rehabilitation stages, thereby reducing the need to replace independent mechanical platforms and realign the patient's ankle during transitions between rehabilitation stages. It should be noted that the training-function coverage discussed in this study is mainly limited to kinematic output and simulated human responses. Since comparative clinical trials with different stage-specific rehabilitation devices have not yet been conducted, the present results cannot demonstrate that the proposed mechanism has exactly the same clinical rehabilitation efficiency as multiple independent dedicated devices. Its actual rehabilitation effectiveness still requires further validation through a physical prototype and clinical trials.

2. Ankle Joint Motion Under Normal Gait Based on AnyBody

The ankle can be approximately regarded as a joint system with three main rotational DOFs. Its typical motions include dorsiflexion/plantarflexion shown in Fig. 1, a, eversion/inversion shown in Fig. 1, b, and adduction/abduction shown in Fig. 1, c [9].

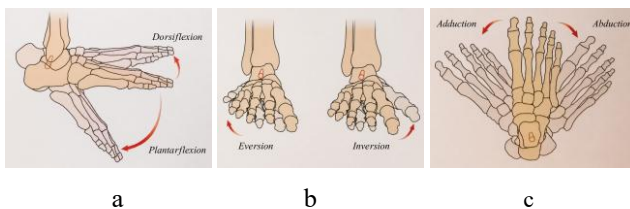


Fig. 1 Main motions of the human ankle joint: a – dorsiflexion/plantarflexion motion, b – eversion/inversion motion, c – adduction/abduction motion

Among these motions, dorsiflexion/plantarflexion mainly participates in foot clearance, toe lifting during the swing phase, and push-off during the stance phase. It is the motion with the largest amplitude and the most direct functional role in normal gait. Inversion/eversion is mainly used

to adjust the plantar contact posture and maintain lateral stability of the human body. Adduction/abduction is related to foot direction regulation and multi-axis compound posture control. Therefore, ankle rehabilitation training should not be limited to single-direction dorsiflexion/plantarflexion; multi-directional coordinated training should be introduced gradually according to the patient's rehabilitation stage.

A full-body motion-capture model based on the AnyMoCap framework in AnyBody Modeling System is used to analyze normal human gait, as shown in Fig. 2. In this study, normal gait refers specifically to the natural level-walking motion recorded in the GaitNormal_1 motion-capture trial rather than a gait pattern generated by prescribing ankle trajectories. The model is driven by C3D motion-capture data. Human joint angles are obtained through marker-trajectory matching and kinematic analysis, followed by inverse dynamic analysis. The initial height and body mass of the human model are 1.75 m and 66 kg, respectively. Frames 160–349 of the normal-gait trial are selected for analysis, and the gravitational acceleration is set to 9.81 m/s^2 . The right ankle-foot complex is selected as the analysis object, and the joint-angle variations over one complete gait cycle are extracted. The ankle-joint angle is used to represent dorsiflexion/plantarflexion in the sagittal plane, while the subtalar-joint angle is used to represent inversion/eversion in the coronal plane. For comparison under a unified time scale, the extracted complete gait cycle is normalized to 0%–100%, and the angle unit is converted from radians to degrees, yielding the ankle-angle variation curves shown in Fig. 3. Because the model is driven by measured C3D motion-capture data, walking speed, cadence, and step length are not independently prescribed input parameters but are inherent spatiotemporal characteristics of the GaitNormal_1 trial itself.

As shown in Fig. 3, during normal gait, ankle dorsiflexion/plantarflexion presents an obvious periodic variation. The angle range is approximately -15° to 16° , and the total range of motion is about 31° . The relatively large amplitude in this direction indicates that dorsiflexion/plantarflexion is the dominant functional motion of the ankle during walking. The subtalar inversion/eversion angle ranges from approximately -12° to 15° , with a total range of about 27° . Its variation trend is not synchronized with dorsiflexion/plantarflexion, indicating that the ankle-foot complex requires not only propulsion and swing control in the sagittal plane during normal gait, but also plantar posture regulation in the coronal plane.

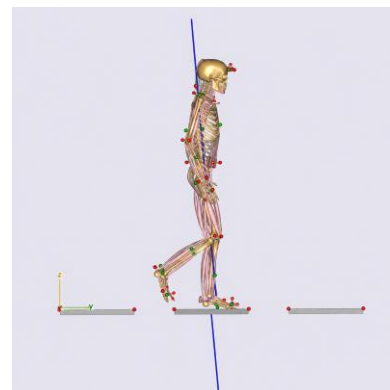


Fig. 2 AnyBody normal-gait simulation model

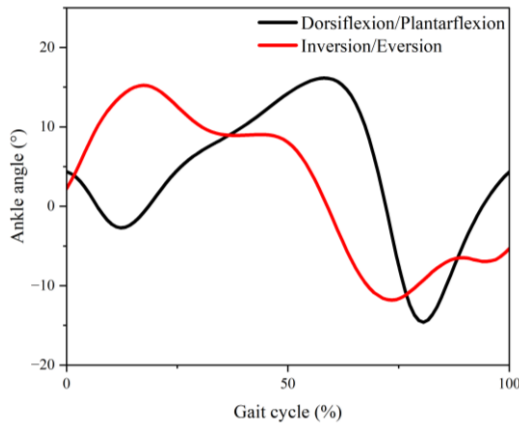


Fig. 3 Ankle-angle variation curves during a normal gait cycle

Comparison between the simulation results and the normal physiological range of ankle motion shows that the actual range used during normal gait is smaller than the maximum physiological range. Previous studies have shown that the maximum physiological range of ankle motion in the sagittal plane is usually 65° - 75° , including about 10° - 20° of dorsiflexion and 40° - 55° of plantarflexion, whereas the sagittal range required during normal walking is approximately 30° . The dorsiflexion/plantarflexion range obtained in this study is about 31° , which is basically consistent with the actual functional demand of the ankle during normal gait. The physiological range of inversion/eversion in the frontal plane is usually smaller than that in the sagittal plane. The inversion/eversion amplitude obtained in this study is lower than the maximum physiological range, but it can reflect the basic demand of foot posture adjustment during normal gait [10].

It should be noted that, in the AnyBody normal-gait model used in this paper, the ankle-foot complex is mainly described by the ankle joint and the subtalar joint, and adduction/abduction in the transverse plane is not output as an independent joint angle. Therefore, the normal-gait analysis in this study mainly extracts the two dominant motion curves of dorsiflexion/plantarflexion and inversion/eversion, without separately presenting an adduction/abduction simulation curve. Considering that multi-axis compound posture training is still required in the middle and late rehabilitation stages, the posture output capacity in the adduction/abduction direction is retained in the mechanism design, and its target output range is set to approximately $\pm 10^{\circ}$ to meet the needs of foot direction regulation and multi-directional coordinated training.

In summary, the normal gait range reflects the ankle angles actually used by healthy people during level walking and can be used as the basic functional recovery target of rehabilitation training. However, the design range of a rehabilitation mechanism should also consider the safety, training intensity, and compound motion requirements of patients at different recovery stages. In early rehabilitation, small-range and low-risk dorsiflexion/plantarflexion training should be the main task, corresponding to mode 1. In the middle stage, constrained compound training such as inversion/eversion can be introduced gradually, corresponding to mode 2. In the later stage, three-DOF compound posture training is required to restore foot direction regulation, proprioception, and gait stability, corresponding

to mode 3. Therefore, the proposed reconfigurable parallel ankle rehabilitation mechanism should have the ability to transition gradually from single DOF to two DOFs and three DOFs, so as to meet the needs of staged ankle rehabilitation.

3. Structure Design of the Reconfigurable Ankle Rehabilitation Mechanism

The proposed reconfigurable parallel ankle rehabilitation mechanism consists of a moving platform, a fixed platform, three UPS driving limbs, and a central reconfigurable constraint chain, as shown in Fig. 4. The moving platform is located at the upper part of the mechanism and is connected to the foot fixation unit. The fixed platform is located at the lower part of the mechanism and acts as the supporting base. The three UPS limbs are uniformly connected between the moving and fixed platforms. In each limb, the upper end is connected to the moving platform through a universal joint, the middle part is a prismatic joint, and the lower end is connected to the fixed platform through a spherical joint. This structure combines the load-bearing advantage of a parallel mechanism with compact spatial arrangement.

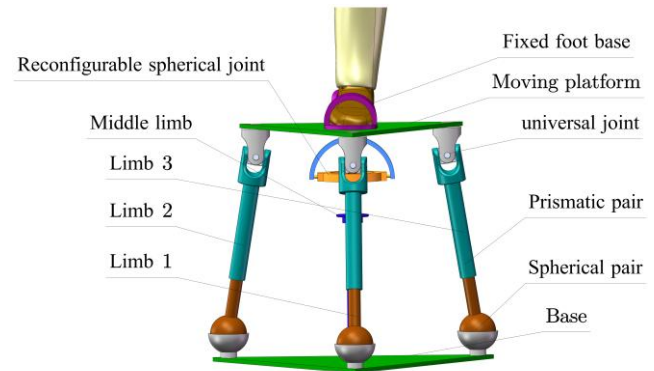


Fig. 4 System design scheme of the reconfigurable parallel mechanism

The core of the mechanism is the central reconfigurable constraint chain. A reconfigurable joint is arranged at the top of this chain, and its equivalent kinematic-pair form can be switched by changing the relative positions and locking states of the internal rotating axes [11]. When axis 1 and axis 2 coincide and axis 3 is locked, the joint is equivalent to a revolute joint, as shown in Fig. 5, a. When axis 1 and axis 2 are perpendicular and axis 3 is locked, the joint is equivalent to a universal joint, as shown in Fig. 5, b. When axis 1 and axis 2 are perpendicular and axis 3 can rotate, the joint is equivalent to a spherical joint, as shown in Fig. 5, c. With the change in the equivalent form of the central joint, the constraint conditions acting on the whole mechanism change accordingly, thereby realizing switching among different DOF modes [12].

To realize practical switching among different configurations of the central reconfigurable joint, axes 1 and 2 adopt a mechanical positioning method based on positioning holes and a positioning pin, as shown in Fig. 6. Multiple predetermined positioning holes are arranged on the corresponding components. By changing the relative position of the ring component, axes 1 and 2 can be adjusted to the relative orientation required by the target configuration. After

the prescribed position is reached, the positioning pin is inserted into the corresponding positioning hole to achieve mechanical positioning and reliable locking. This method avoids maintaining the configuration only through friction and ensures that the relative position of axes 1 and 2 does not change during rehabilitation training.

When axes 1 and 2 are adjusted to the coincident configuration and mechanically positioned, the central joint can be made equivalent to an R joint by locking axis 3. When axes 1 and 2 are adjusted to the orthogonal configuration, the central joint can be further made equivalent to a U or S joint depending on whether axis 3 is locked or released.

Unlike the mechanical positioning method used for axes 1 and 2, axis 3 employs an electromagnetic brake to lock or release its rotational DOF, as shown in Fig. 7. The electromagnetic brake is arranged coaxially with axis 3. By controlling the braking and releasing states of the brake, axis 3 can be switched between the locked and freely rotating states. When the electromagnetic brake is engaged, the rotation of axis 3 is restricted; when the brake is released, axis 3 can rotate about its own axis.

Therefore, in Mode 1, axes 1 and 2 are adjusted to the coincident configuration and axis 3 is locked by the electromagnetic brake, making the central joint equivalent to an R joint. In Mode 2, axes 1 and 2 are adjusted to the orthogonal configuration and axis 3 remains locked, making the central joint equivalent to a U joint. In Mode 3, axes 1 and 2 remain orthogonal while the electromagnetic brake of axis 3 is released, allowing rotation about all three axes and making the central joint equivalent to an S joint.

Thus, switching among the three training modes is

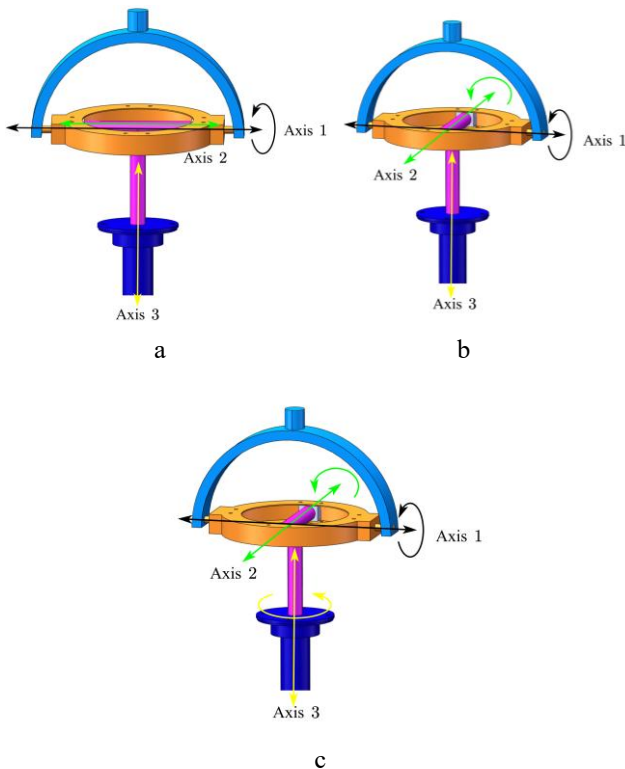


Fig. 5 Three sub-configurations of the reconfigurable motion unit: a – revolute-joint configuration, b – universal-joint configuration, c – spherical-joint configuration

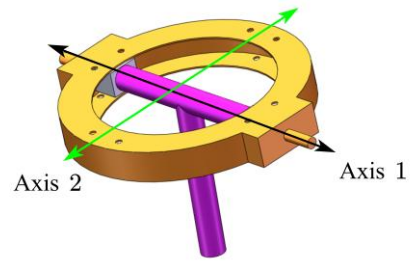


Fig. 6 Mechanical positioning and locking structure of axes 1 and 2

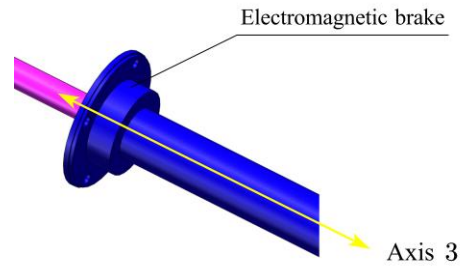


Fig. 7 Electromagnetic braking structure for locking axis 3

accomplished through two components: the relative configuration of axes 1 and 2 is determined by the positioning holes and mechanical locking element, while the rotational state of axis 3 is controlled by the electromagnetic brake. Mechanical positioning determines the basic configuration of the central joint, whereas the electromagnetic brake controls the locking and release of the third rotational DOF.

To avoid incorrect identification of the training mode, the current mode is jointly determined from the mechanical positioning state of axes 1 and 2 and the electromagnetic braking state of axis 3. The operator first checks the positioning-pin location to determine whether axes 1 and 2 are in the coincident or orthogonal configuration and then checks whether the electromagnetic brake of axis 3 is locked or released. When axes 1 and 2 are coincident and axis 3 is locked, the mechanism is in Mode 1. When axes 1 and 2 are orthogonal and axis 3 is locked, the mechanism is in Mode 2. When axes 1 and 2 are orthogonal and axis 3 is released, the mechanism is in Mode 3. Therefore, the current training mode can be confirmed before operation by checking both the mechanical positioning state of the positioning pin and the braking state of the electromagnetic brake.

According to the three sub-configurations of the central reconfigurable motion unit, the parallel mechanism is divided into three training modes, as listed in Table 1.

The main features of the mechanism are as follows: the three UPS limbs mainly perform driving and bearing

Table 1

Mode switching table

Training mode	Equivalent joint of the central limb	Freely rotatable axes	Rehabilitation training meaning
Mode 1	R joint	Axis 1	Early-stage low-risk training
Mode 2	U joint	Axes 1 and 2	Middle-stage restricted compound training
Mode 3	S joint	Axes 1, 2, and 3	Late-stage three-DOF compound training

functions and do not impose independent constraints on the moving platform; the central reconfigurable constraint chain acts as the DOF regulation unit and changes the number of constraints on the moving platform through the three equivalent kinematic-pair forms of R, U, and S. Therefore, the mechanism can switch among 1R, 2R, and 3R output modes without replacing the overall structure, thereby overcoming the limited adaptability of fixed-DOF mechanisms at different rehabilitation stages.

To ensure the safety of mode switching, the reconfiguration operation is not performed dynamically during rehabilitation motion. Before mode switching, the moving platform is first returned to the neutral position and the motion of the three UPS driving limbs is stopped so that the central joint is not subjected to significant dynamic loading. The mechanical locking of axes 1 and 2 is then released, the axes are adjusted to the target positioning-hole location, and the positioning pin is reinserted. Finally, axis 3 is locked or released by the electromagnetic brake according to the target mode. Rehabilitation training is restarted only after the mechanical positioning and braking states have been confirmed. Therefore, mode switching is not a continuous dynamic reconfiguration process during motion but is performed while the mechanism is stationary, thereby avoiding instantaneous impacts and unintended free motion caused by switching under load.

The three training modes are therefore not realized by three independent mechanisms but correspond to three motion subspaces formed by the same main mechanism under different constraint states. As rehabilitation progresses, the central constraint chain gradually reduces the rotational constraints imposed on the moving platform, allowing the mechanism to transition progressively from single-axis motion to multi-axis coordinated motion.

This DOF regulation method can also reduce the control complexity at different rehabilitation stages to a certain extent. In Mode 1, the central constraint chain structurally restricts two non-target rotational DOFs, leaving only one rotational DOF for dorsiflexion/plantarflexion. In Mode 2, two target rotational DOFs are retained while the third non-target rotational DOF remains constrained. In Mode 3, all three rotational DOFs are released. Therefore, during the restricted training of Modes 1 and 2, the control system does not need to rely entirely on active feedback control to suppress all non-target motions. Instead, it only needs to coordinate the three UPS driving limbs within the currently permitted motion subspace, thereby reducing the number of posture variables that must be simultaneously regulated and decreasing control complexity.

At the same time, the DOF constraints directly formed by the mechanical structure provide an additional safety measure in addition to active control. In Modes 1 and 2, the restricted non-target DOFs are physically limited through mechanical positioning/locking and braking constraints of the central joint. If a control-system abnormality, sensor-signal deviation, or unexpected drive command occurs, the constrained DOFs remain physically restricted and will not be released solely because of the control error. This can reduce unintended motion of the moving platform in non-target directions to a certain extent and consequently reduce the risk of secondary injury caused by unexpected multi-axis motion, particularly for patients in the early rehabilitation stage. It should be noted that these structural constraints serve as a supplement to active control safety

measures and cannot replace actuator limits, emergency-stop functions, or other control-system safety protections.

Throughout mode switching, the three UPS driving limbs, the foot-fixation platform, and the main structure of the mechanism remain unchanged. Only the configuration and locking state of the central reconfigurable constraint chain are changed. Thus, while satisfying the requirements of staged rehabilitation motion, the proposed mechanism also provides a structural basis for reducing control complexity and improving the structural-level safety of restricted rehabilitation training.

4. Mobility and Kinematic Analysis

4.1. Coordinate system establishment and mobility analysis

Taking the geometric center O of the moving platform as the origin, the moving coordinate system $O-XYZ$ is established. The Z axis is perpendicular to the moving platform and points downward, the X axis points to the universal joint A_1 , and the Y axis is determined by the right-hand rule. A fixed coordinate system $O_1-X_1Y_1Z_1$ is established at the geometric center of the fixed platform. The Z_1 axis is perpendicular to the fixed platform and points downward, and the X_1 and Y_1 axes have the same directions as the X and Y axes of the moving platform, as shown in Fig. 9. The fixed coordinate system is used as the reference frame, and the coordinates of points on the moving platform are mapped into this coordinate system through posture transformation for subsequent inverse kinematic solution.

For the i -th UPS limb, a local coordinate system $A_i-x_iy_iz_i$ ($i = 1, 2, 3$) is established at the center of the upper universal joint, where the x and y axes are along the two rotational axes of the universal joint and the z axis is along the centerline of the limb from top to bottom, pointing to the center of the lower spherical joint, as shown in Fig. 10, a. A local coordinate system $O-x_cy_cz_c$ is established at the center

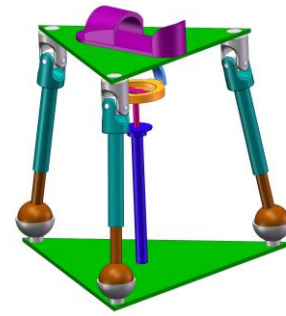


Fig. 8 Mechanism model

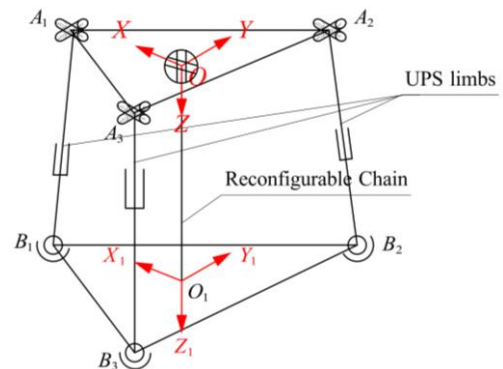


Fig. 9 Schematic diagram of the mechanism

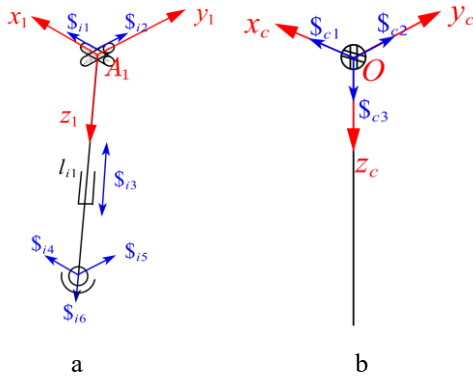


Fig. 10 Schematic diagrams of the limbs: a – UPS limb, b – central reconfigurable limb in mode 3

of the reconfigurable spherical joint at the top of the central reconfigurable constraint chain, as shown in Fig. 10, b.

Since mode 3 is the fully released three-DOF training mode and also the basis for the subsequent analysis, the mobility analysis is first carried out for mode 3.

Because the three UPS limbs have the same topology, only one limb is taken as an example. The screw system of the UPS limb is:

$$\begin{cases} \mathcal{S}_{i1} = (1 \ 0 \ 0; \ 0 \ 0 \ 0) \\ \mathcal{S}_{i2} = (0 \ 1 \ 0; \ 0 \ 0 \ 0) \\ \mathcal{S}_{i3} = (0 \ 0 \ 0; \ 0 \ 0 \ 1) \\ \mathcal{S}_{i4} = (1 \ 0 \ 0; \ 0 \ l_{i1} \ 0) \\ \mathcal{S}_{i5} = (0 \ 1 \ 0; \ -l_{i1} \ 0 \ 0) \\ \mathcal{S}_{i6} = (0 \ 0 \ 1; \ 0 \ 0 \ 0) \end{cases}; i = 1, 2, 3, \quad (1)$$

where the terms denote the motion screws of the joints in the limb, and the limb length, respectively. The first two terms correspond to the two rotational DOFs of the upper universal joint, the third term corresponds to the translational DOF of the middle prismatic joint, and the last three terms correspond to the three concurrent rotational DOFs decomposed from the lower spherical joint.

The screw system can be written as the following screw matrix:

$$T_1 = \begin{bmatrix} 1 & 0 & 0 & 1 & 0 & 0 \\ 0 & 1 & 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 0 & 0 & 1 \\ 0 & 0 & 0 & 0 & -l_1 & 0 \\ 0 & 0 & 0 & l_1 & 0 & 0 \\ 0 & 0 & 1 & 0 & 0 & 0 \end{bmatrix}. \quad (2)$$

From Eq. (2), the motion screw system of a single UPS limb is full rank and provides no independent constraint; therefore, it is essentially an unconstrained driving limb.

In mode 3, the central reconfigurable joint is equivalent to a spherical joint, and the motion screws of this limb are:

$$\begin{cases} \mathcal{S}_{c1} = (1 \ 0 \ 0; \ 0 \ 0 \ 0) \\ \mathcal{S}_{c2} = (0 \ 1 \ 0; \ 0 \ 0 \ 0), \ i = 3. \\ \mathcal{S}_{c3} = (0 \ 0 \ 1; \ 0 \ 0 \ 0) \end{cases} \quad (3)$$

The reciprocal screws of the reconfigurable limb motion screw system in the limb coordinate system $O-x_c y_c z_c$ are:

$$\begin{cases} \mathcal{S}_{c1}^r = (1 \ 0 \ 0; \ 0 \ 0 \ 0) \\ \mathcal{S}_{c2}^r = (0 \ 1 \ 0; \ 0 \ 0 \ 0), \ i = 3. \\ \mathcal{S}_{c3}^r = (0 \ 0 \ 1; \ 0 \ 0 \ 0) \end{cases} \quad (4)$$

From Eq. (4), the reciprocal screws of the reconfigurable limb are constraint force vectors along the three axes of its limb coordinate system, which limit the translations of the moving platform along these three axes.

According to screw theory, all three UPS limbs are unconstrained driving limbs. In mode 3, the central reconfigurable joint is equivalent to a spherical joint, and the constraint chain provides three independent constraint forces, eliminating the three translational DOFs of the moving platform. Therefore, the moving platform retains three rotational DOFs, and the mobility of the mechanism is $6 - 3 = 3$.

To describe the posture variables uniformly, α , β , and γ are defined as the rotation angles of the moving platform about the X_1 , Y_1 , and Z_1 axes of the fixed coordinate system, respectively. Here, α corresponds to eversion/inversion, β corresponds to dorsiflexion/plantarflexion, and γ corresponds to adduction/abduction. In mode 2, the central reconfigurable joint is equivalent to a universal joint. In addition to the above three translational constraints, one constraint couple is introduced so that $\gamma = 0$. The moving platform therefore retains only two rotational DOFs, and the mobility of the mechanism is $6 - 4 = 2$. This additional constraint suppresses rotation in the non-target direction and degenerates the three-dimensional posture domain of mode 3 into a two-dimensional posture subdomain, which is beneficial for direction stability in middle-stage restricted compound training.

In mode 1, the central reconfigurable joint is equivalent to a revolute joint. On the basis of the three translational constraints, two mutually independent constraint couples are added so that $\alpha = \gamma = 0$ and only the rotation about the target Y_1 axis is retained. The mobility of the mechanism is therefore $6 - 5 = 1$. Thus, mode 1 further degenerates from the two-dimensional posture subdomain of mode 2 into a one-dimensional rotational trajectory, reducing cross-axis coupling and control complexity in early rehabilitation training.

4.2. Inverse kinematic model

Because mode 3 corresponds to the fully unlocked three-DOF state of the mechanism, it can fully reflect the spatial posture variation of the moving platform around the center point and is the configuration with the highest mobility and the most complete kinematic relationship among the three training modes. Therefore, a unified inverse kinematic model is first established for mode 3.

In the moving coordinate system $O-XYZ$, the centers of the three universal joints on the moving platform are

set as follows, and their coordinates are expressed as:

$${}^m a_i = \begin{bmatrix} r_m \cos \phi_i \\ r_m \sin \phi_i \\ 0 \end{bmatrix}, \quad (5)$$

where r_m is the distribution radius of the moving platform, and the parameter is the polar angle of the i -th universal joint.

In the fixed coordinate system O_1 - X_1 Y_1 Z_1 , the centers of the three spherical joints on the fixed platform are set as follows, and their coordinates are expressed as:

$${}^s b_i = \begin{bmatrix} r_b \cos \varphi_i \\ r_b \sin \varphi_i \\ 0 \end{bmatrix}, \quad (6)$$

where r_b is the distribution radius of the fixed platform, and the parameter is the polar angle of the i -th spherical joint.

According to the posture angles defined in Section 4.1, the Z_1 - Y_1 - X_1 rotation sequence is adopted, and the posture matrix is:

$$\mathbf{C} = \mathbf{R}_{Z_1}(\gamma) \mathbf{R}_{Y_1}(\beta) \mathbf{R}_{X_1}(\alpha), \quad (7)$$

where

$$\begin{aligned} \mathbf{R}_{X_1}(\alpha) &= \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos \alpha & -\sin \alpha \\ 0 & \sin \alpha & \cos \alpha \end{bmatrix}, \\ \mathbf{R}_{Y_1}(\beta) &= \begin{bmatrix} \cos \beta & 0 & \sin \beta \\ 0 & 1 & 0 \\ -\sin \beta & 0 & \cos \beta \end{bmatrix}, \\ \mathbf{R}_{Z_1}(\gamma) &= \begin{bmatrix} \cos \gamma & -\sin \gamma & 0 \\ \sin \gamma & \cos \gamma & 0 \\ 0 & 0 & 1 \end{bmatrix}. \end{aligned} \quad (8)$$

The position of the moving-platform origin O in the fixed coordinate system is:

$${}^s p_o = \overline{O_1 O} = \begin{bmatrix} 0 \\ 0 \\ -h \end{bmatrix}, \quad (9)$$

where h is the length of the central limb.

Therefore, the position vector of the i -th universal-joint center A_i of the moving platform in the fixed coordinate system is:

$${}^s A_i = {}^s p_o + \mathbf{C}^m a_i. \quad (10)$$

The two ends of the i -th UPS limb are the upper universal-joint center A_i and the lower spherical-joint center B_i . Therefore, the vector of this limb in the fixed coordinate system can be expressed as:

$$l_i = {}^s b_i - {}^s A_i = {}^s b_i - {}^s p_o - \mathbf{C}^m a_i. \quad (11)$$

Let the driving length of the i -th limb be q_i . Then:

$$q_i = \|l_i\|, \quad i = 1, 2, 3. \quad (12)$$

Thus, the inverse solution equations of the three UPS limbs can be written uniformly as:

$$q_i = \|{}^s b_i - {}^s p_o - \mathbf{C}^m a_i\|, \quad i = 1, 2, 3. \quad (13)$$

The expression for the length of the i -th limb is obtained as:

$$q_i = \sqrt{{}^s r_b^2 + r_m^2 + h^2 - 2 {}^s b_i^T \mathbf{C}^m a_i + 2 {}^s p_o^T \mathbf{C}^m a_i}. \quad (14)$$

This is the inverse kinematic model of the mechanism in mode 3. The model reflects the mapping relationship between the lengths of the three driving limbs and the three-DOF posture of the moving platform, and it can be directly used for the subsequent workspace analysis. In mode 2, $\gamma = 0$ is imposed and α and β are used as the posture variables. In mode 1, $\alpha = \gamma = 0$ is further imposed and only β is retained. Substituting the corresponding constraints into the inverse solution of mode 3 gives the limb lengths for modes 2 and 1, without establishing independent equation sets.

4.3. Velocity Jacobian matrix analysis

According to the inverse kinematic analysis, the lengths of the three UPS driving limbs are functions of the posture angles of the moving platform, namely:

$$\begin{aligned} q_1 &= f_1(\alpha, \beta, \gamma), \\ q_2 &= f_2(\alpha, \beta, \gamma), \\ q_3 &= f_3(\alpha, \beta, \gamma). \end{aligned} \quad (15)$$

Since the limb length satisfies:

$$q_i^2 = l_i^T l_i. \quad (16)$$

Taking the derivative of the above equation with respect to time t gives:

$$q_i \dot{q}_i = g_{i1} \dot{\alpha} + g_{i2} \dot{\beta} + g_{i3} \dot{\gamma}, \quad (17)$$

where

$$\begin{aligned} g_{i1} &= \frac{1}{2} \frac{\partial q_i^2}{\partial \alpha}, \\ g_{i2} &= \frac{1}{2} \frac{\partial q_i^2}{\partial \beta}, \\ g_{i3} &= \frac{1}{2} \frac{\partial q_i^2}{\partial \gamma}. \end{aligned} \quad (18)$$

The velocity equations of the three limbs can be written uniformly in matrix form as:

$$J_t \dot{q} = G \dot{\theta}, \quad (19)$$

where

$$\dot{q} = \begin{bmatrix} \dot{q}_1 \\ \dot{q}_2 \\ \dot{q}_3 \end{bmatrix}, \quad (20)$$

$$\dot{\theta} = \begin{bmatrix} \dot{\alpha} \\ \dot{\beta} \\ \dot{\gamma} \end{bmatrix}, \quad (21)$$

$$J_t = \begin{bmatrix} q_1 & 0 & 0 \\ 0 & q_2 & 0 \\ 0 & 0 & q_3 \end{bmatrix}, \quad (22)$$

$$G = \begin{bmatrix} g_{11} & g_{12} & g_{13} \\ g_{21} & g_{22} & g_{23} \\ g_{31} & g_{32} & g_{33} \end{bmatrix}, \quad (23)$$

where the matrix is the velocity coefficient matrix of the driving limbs. Because the length variation of each driving limb corresponds only to its own extension velocity, and no direct velocity cross term exists among the limbs, the matrix is diagonal.

Further rearrangement gives the first-order velocity Jacobian matrix of the mechanism as:

$$\dot{q} = J\dot{\theta}, \quad (24)$$

where

$$J = J_t^{-1}G, \quad (25)$$

namely,

$$J = \begin{bmatrix} q_1 & 0 & 0 \\ 0 & q_2 & 0 \\ 0 & 0 & q_3 \end{bmatrix}^{-1} \begin{bmatrix} g_{11} & g_{12} & g_{13} \\ g_{21} & g_{22} & g_{23} \\ g_{31} & g_{32} & g_{33} \end{bmatrix}. \quad (26)$$

It can be seen that the velocity Jacobian matrix describes the mapping relationship between the posture angular velocity of the moving platform and the extension velocity of the three UPS driving limbs.

In mode 3, the mechanism has three rotational DOFs, and the velocity Jacobian matrix is a third-order square matrix. When

$$\det(J) \neq 0, \quad (27)$$

the mechanism has normal velocity transmission capability.

When

$$\det(J) = 0, \quad (28)$$

the mechanism is in a velocity singularity.

For mode 2, the central reconfigurable joint is equivalent to a universal joint, and the moving platform retains only two rotational DOFs about the X_1 and Y_1 axes. Therefore:

$$\gamma = 0, \dot{\gamma} = 0. \quad (29)$$

At this time, the velocity equation degenerates into:

$$\dot{q} = J_2 \begin{bmatrix} \dot{\alpha} \\ \dot{\beta} \end{bmatrix}. \quad (30)$$

For mode 1, the central reconfigurable joint is equivalent to a revolute joint, and the moving platform retains only dorsiflexion/plantarflexion about the Y_1 axis. Therefore:

$$\begin{aligned} \alpha &= 0, \dot{\alpha} = 0, \\ \gamma &= 0, \dot{\gamma} = 0. \end{aligned} \quad (31)$$

At this time, the velocity equation further degenerates into:

$$\dot{q} = J_1 \dot{\beta}. \quad (32)$$

In summary, the velocity Jacobian matrix in mode 3 can be used to analyze velocity transmission performance and singularity in three-DOF posture motion. Modes 2 and 1 can be obtained by constrained degeneration of the velocity Jacobian matrix of mode 3, corresponding to two-DOF restricted compound training and single-DOF dorsiflexion/plantarflexion training, respectively.

4.4. Workspace of the end-effector characteristic point

Based on the inverse kinematic model and considering the constraints of limb length and joint rotation angle, MATLAB is used to numerically search the workspace of the end-effector characteristic point in each mode. The geometric parameters of the mechanism are determined by considering human-foot dimensions, foot-fixation requirements, spatial arrangement of the driving limbs, and practical patient posture. According to adult male anthropometric dimensions, a representative foot length is approximately 251 mm, while larger foot lengths can reach approximately 279 mm. Therefore, in addition to providing sufficient space for the placement and fixation of an adult foot with a length of approximately 280 mm, the moving platform must reserve additional space around the foot for the fixation device, universal-joint installation, and motion clearance. Considering both the human-foot dimensions and the structural arrangement of the mechanism, the distribution radius of the three upper universal-joint centers is selected as $r_m = 317$ mm. The distribution radius of the fixed-platform joints is made appropriately larger than that of the moving platform so that the three UPS limbs have a certain outward inclination in the initial configuration, avoiding an approximately parallel and vertical arrangement and providing sufficient space for limb extension and swinging during posture changes. Therefore, $r_b = 400$ mm is selected. The axial distance h between the moving and fixed platforms is affected by both mechanism motion space and patient sitting posture. An excessively large h increases the overall mechanism height and the foot-training position, requiring a higher seat for the patient. An excessively small h reduces the effective space between the moving and fixed platforms and increases the initial inclination of the UPS limbs, thereby increasing

the risk of structural interference during large posture motions. In addition, the central reconfigurable constraint chain and the mechanical positioning and locking structures of axes 1 and 2 are located between the moving and fixed platforms, and rehabilitation personnel need to operate the positioning pin and corresponding positioning locations during mode switching. Therefore, sufficient manual operating space must also be reserved between the two platforms so that the locking element can be removed and inserted, the configuration can be adjusted, and the locking state can be checked after the mechanism has stopped. An excessively small h would restrict the operating space around the central reconfigurable mechanism and reduce the safety and convenience of switching among training modes. Considering the overall mechanism height, patient sitting posture, initial inclination of the UPS limbs, target motion range of the moving platform, structural clearance, and manual operating space of the central reconfigurable mechanism, $h = 483$ mm is selected. These parameters are the baseline dimensions of the current mechanism. For different patients, adaptation is mainly achieved by adjusting the foot-fixation unit, seat height, and relative human-machine position rather than changing the main geometric dimensions of the mechanism.

The search result for mode 3 is shown in Fig. 11. The workspace of the end-effector characteristic point is continuous overall and presents a relatively regular spherical-cap distribution, without obvious fracture or scattered points, indicating that the established model has good numerical continuity within the given search range. For mode 2, after the constraint is imposed, the feasible workspace is the intersection between the feasible domain of mode 3 and the corresponding constraint manifold, showing a continuous two-dimensional section. For mode 1, after an additional constraint is imposed, the workspace degenerates into a continuous one-dimensional trajectory. Since the workspaces of modes 2 and 1 can be obtained by constrained degeneration of mode 3, their workspace figures are not separately plotted. Therefore, the three modes share the same inverse-solution framework, while the workspace dimensions are 3, 2, and 1, corresponding to late-stage multi-axis compound training, middle-stage restricted compound training, and early-stage single-axis training, respectively.

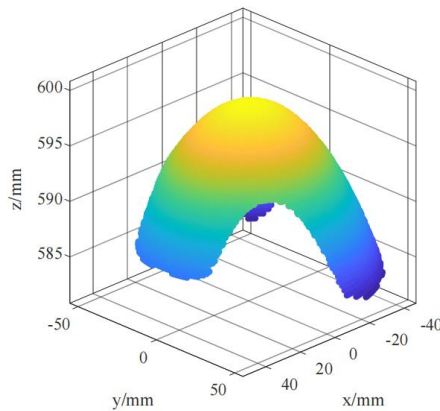


Fig. 11 Workspace of the end-effector characteristic point in mode 3

4.5. Dexterity analysis of the mechanism

As described above, the mechanism has three rotational DOFs in mode 3, and the posture angular velocity of the moving platform and the velocity of the driving limbs satisfy:

$$\dot{q} = J\dot{\theta}, \quad (33)$$

where $\dot{q}$ is the velocity vector of the three UPS driving limbs, $\dot{\theta}$ is the posture angular velocity vector of the moving platform, and J is the first-order velocity Jacobian matrix of the mechanism.

To further evaluate the velocity transmission performance and motion dexterity of the mechanism in the workspace, the condition number of the Jacobian matrix is adopted as the evaluation index. The condition number can be expressed as:

$$\kappa(J) = \|J\| \|J^{-1}\|. \quad (34)$$

According to singular value decomposition, the condition number can also be written as:

$$\kappa(J) = \frac{\sigma_{\max}}{\sigma_{\min}}, \quad (35)$$

where $\sigma_{\max}$ and $\sigma_{\min}$ are the maximum and minimum singular values of the Jacobian matrix, respectively. The closer the condition number is to 1, the more balanced the velocity transmission performance in different directions and the better the dexterity. A larger condition number indicates poorer velocity transmission performance.

To express the mechanism dexterity more intuitively, the reciprocal of the condition number is adopted as the local condition index (LCI), namely:

$$LCI = \frac{1}{\kappa(J)}. \quad (36)$$

Substituting Eq. (35) into Eq. (36) gives:

$$LCI = \frac{\sigma_{\min}}{\sigma_{\max}}. \quad (37)$$

Therefore, the value range of LCI is:

$$0 \leq LCI \leq 1. \quad (38)$$

According to Eq. (38), the closer the LCI is to 1, the better the velocity transmission performance and the higher the motion dexterity of the mechanism. The closer the LCI is to 0, the poorer the velocity transmission performance. Based on the inverse kinematic model and velocity Jacobian matrix established above, MATLAB is used to numerically calculate the LCI values at posture points within the specified posture range of mode 3. According to the normal gait range of the ankle joint and the training requirements of the mechanism, the posture search range in mode 3 is set as:

$$-25^\circ \leq \alpha \leq 25^\circ$$

$$-30^\circ \leq \beta \leq 30^\circ$$

$$-20^\circ \leq \gamma \leq 20^\circ$$

This range is larger than the ankle angle actually used during normal gait and is used to examine the velocity transmission performance of the mechanism within an expanded training posture range. The dexterity distribution of the mechanism is obtained as shown in Fig. 12.

As shown in Fig. 12, within the specified posture range of mode 3, the LCI values of the mechanism are continuously distributed overall, indicating that the established velocity Jacobian model has good numerical continuity within this range. From the global distribution results, the maximum LCI value of the mechanism is about 0.50 and the average value is about 0.21, showing that the mechanism has certain velocity transmission capability within the specified posture range.

From the three typical section results, the LCI distributions in the $\alpha = 0^\circ$ and $\beta = 0^\circ$ sections are relatively

similar. The surfaces vary continuously and the maximum value is about 0.40, indicating that the mechanism can maintain certain motion dexterity during compound posture changes involving dorsiflexion/plantarflexion, eversion/inversion, and adduction/abduction. However, in the $\gamma = 0^\circ$ section, the overall LCI value is relatively low and the maximum value is about 0.18, indicating that when the adduction/abduction angle is at the middle position, the velocity transmission performance of the mechanism is relatively weak in some postures.

In summary, the dexterity analysis based on the local condition index shows that the proposed mechanism has certain velocity transmission capability in mode 3, but the dexterity is relatively low in some posture regions. Therefore, in practical rehabilitation trajectory planning, the mechanism should avoid long-term operation in low-LCI regions, and posture ranges with higher dexterity should be preferentially selected as the main training workspace to improve motion stability and human-machine interaction safety.

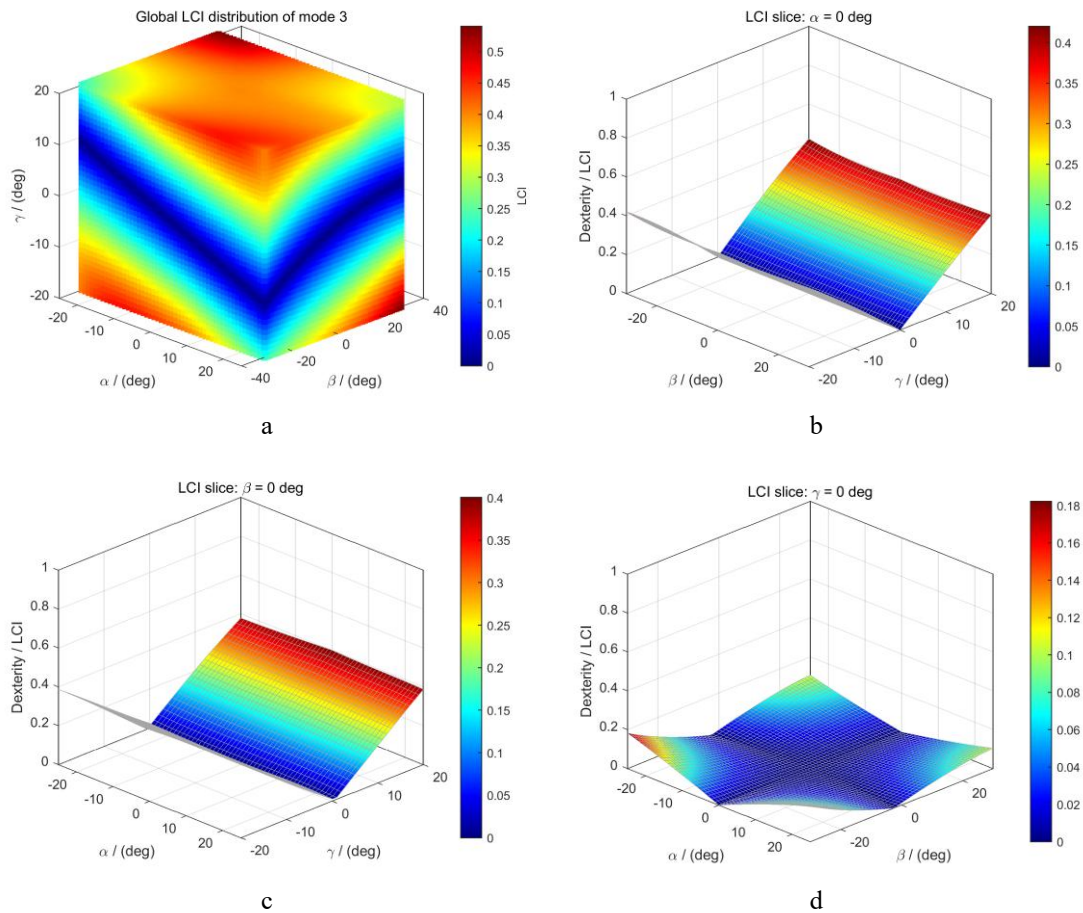


Fig. 12 Dexterity distribution of the mechanism in mode 3: a – global LCI distribution, b – dexterity surface at $\alpha = 0^\circ$, c – dexterity surface at $\beta = 0^\circ$, d – dexterity surface at $\gamma = 0^\circ$

5. Kinematic Simulation and Validation of the Mechanism

To verify the DOF analysis and inverse kinematic model of the mechanism, a virtual prototype validation method used for similar ankle rehabilitation mechanisms [13] is referenced. The three-dimensional model established in SolidWorks is imported into ADAMS to build the virtual prototype, as shown in Fig. 13. The simulation duration is

set to 5 s and the number of simulation steps is set to 500. The rotation angle, angular velocity, angular acceleration of the moving platform, and the displacement of the three UPS driving limbs are extracted to analyze the DOF, posture output direction, and continuity of the driving-limb motion under different configurations.

As shown in Figs. 14, a-c, under dorsiflexion/plantarflexion, eversion/inversion, and adduction/abduction conditions, the moving platform mainly rotates

about the Y_1 , X_1 , and Z_1 axes, respectively, and the rotation angles in non-target directions remain near 0° . The rotation ranges of the three motions are approximately -15° – 16° , -12° – 15° , and -10° – 10° , respectively, indicating that the mechanism can output posture in the predetermined directions. The simulation results of angular velocity and angular acceleration curves of the moving platform in Fig. 14, d–14, f show that the peak angular velocities of dorsiflexion/plantarflexion, eversion/inversion, and adduction/abduction are approximately $48.7^\circ/\text{s}$, $42.4^\circ/\text{s}$, and $31.4^\circ/\text{s}$, respectively, and the peak angular accelerations are approximately $153^\circ/\text{s}^2$, $133^\circ/\text{s}^2$, and $98.7^\circ/\text{s}^2$, respectively. All curves are continuous and smooth without obvious mutation. Therefore, the DOF, posture output, and limb motion law of the virtual prototype are consistent with the theoretical analysis, verifying the correctness of the mechanism configuration and inverse kinematic model from the kinematic perspective.

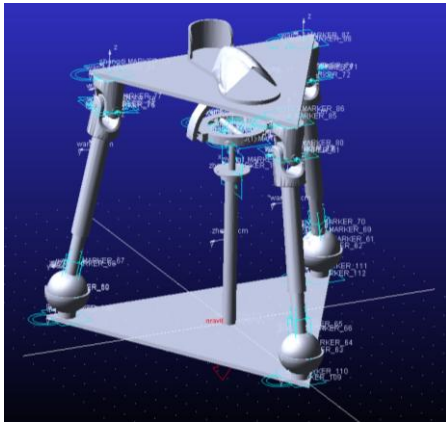


Fig. 13 ADAMS virtual prototype

As shown in Figs. 15, a–c, all driving limbs show continuous periodic extension and contraction. During dorsiflexion/plantarflexion, limb 1 has a larger variation amplitude, while limbs 2 and 3 have basically the same variation pattern. During eversion/inversion, limbs 2 and 3 extend and contract in opposite directions, while limb 1 changes slightly. During adduction/abduction, the displacement amplitudes of the three limbs are small, and the maximum value is about 3.2 mm. These patterns are consistent with the geometric characteristics of the three target motions.

It should be noted that the main purpose of the ADAMS simulation in this study is to verify the mechanism DOFs, posture-output directions, and motion laws of the driving limbs. Therefore, the current model mainly adopts ideal kinematic pairs for kinematic validation and does not systematically consider non-ideal factors such as joint friction, assembly clearance, structural flexibility, and active or passive resistance of the patient. In an actual mechanism, these factors do not change the theoretical DOFs determined by the mechanism constraints, but they increase the driving forces required by the UPS limbs and may lead to motion-tracking errors, local impacts, and energy losses. In passive rehabilitation training in particular, ankle-joint stiffness and muscle resistance directly affect actuator loading. Therefore, human–mechanism interaction forces should be further considered in the dynamic model for practical actuator selection and control-strategy design. The human muscle-response analysis in this study is mainly used to evaluate muscle-response patterns under different training directions,

and human resistance is not fed back in real time to the ADAMS mechanism model. Therefore, the present results mainly validate the kinematic feasibility of the mechanism, while dynamic coupling analysis including friction and human resistance, together with physical prototype experiments, will be carried out in future work.

6. Human–Machine Coupling Simulation Analysis

6.1. Human–machine coupling model establishment and evaluation indexes

To evaluate the correspondence between the training motions generated by the proposed mechanism and the responses of target ankle muscle groups, a human–machine coupling model is established in AnyBody Modeling System. The human model is an adult male model with a height of 175 cm and a weight of 70 kg [14], as shown in Fig. 16. The software can describe human segments, joints, muscles, and external constraints, and it solves muscle activity level, muscle force, and muscle–tendon length through inverse dynamics [15]. In this paper, maximum muscle activity, activity of the main muscles, main muscle force, and total muscle–tendon length are selected as evaluation indexes. Muscle activity is a normalized muscle recruitment index calculated by AnyBody inverse dynamics. A maximum muscle activity below 1.0 indicates that no obvious muscle-force overload occurs under the specified model parameters and loading conditions [16]. The activity of the main muscles and the muscle force are used to analyze the participation degree of target muscle groups. The total muscle–tendon length is used to judge the shortening or stretching state of the related muscle groups during training, thereby comprehensively evaluating training safety and the directionality of muscle-group stimulation.

To further verify the correspondence between mechanism output and ankle target-muscle response under different training directions, human–machine coupling simulations are carried out for six typical single-direction training motions: dorsiflexion, plantarflexion, inversion, eversion, adduction, and abduction. Due to space limitations, the complete curves of all six motions are not analyzed one by one. Instead, the maximum muscle activity, main responding muscles, and muscle–tendon length variation patterns of the six training motions are summarized in Table 2, and typical training motions are selected for graphical illustration.

In practical rehabilitation applications, the patient performs the training in a seated posture. The seat height is not directly determined by the mechanism parameter ($h = 483$ mm) but should be adjusted according to the patient's lower-limb dimensions. The adjustment principle is to provide sufficient thigh support, maintain the knee at a comfortable flexion angle of approximately 90° , keep the lower leg close to its natural vertical position, and align the anatomical ankle center of the training side as closely as possible with the equivalent rotation center of the mechanism. Because patients differ in tibial length, foot size, and sitting height, the seat height is determined according to the individual patient rather than being fixed at a uniform value. For the non-training-side lower limb, an independent fixed foot support is provided so that it does not participate in the moving-platform motion and to prevent pelvic-posture deviation or discomfort caused by leaving the contralateral foot unsupported during training. The height of the auxiliary

foot support can be adjusted according to the patient's lower-limb length. The driving limbs and the central constraint chain are arranged below and around the foot platform and do not occupy the main operating space around the

knee and lateral side of the lower leg. This allows rehabilitation personnel to assist the patient from the side during foot fixation, mode confirmation, and emergency release. For patients with limited lower-limb mobility, sitting

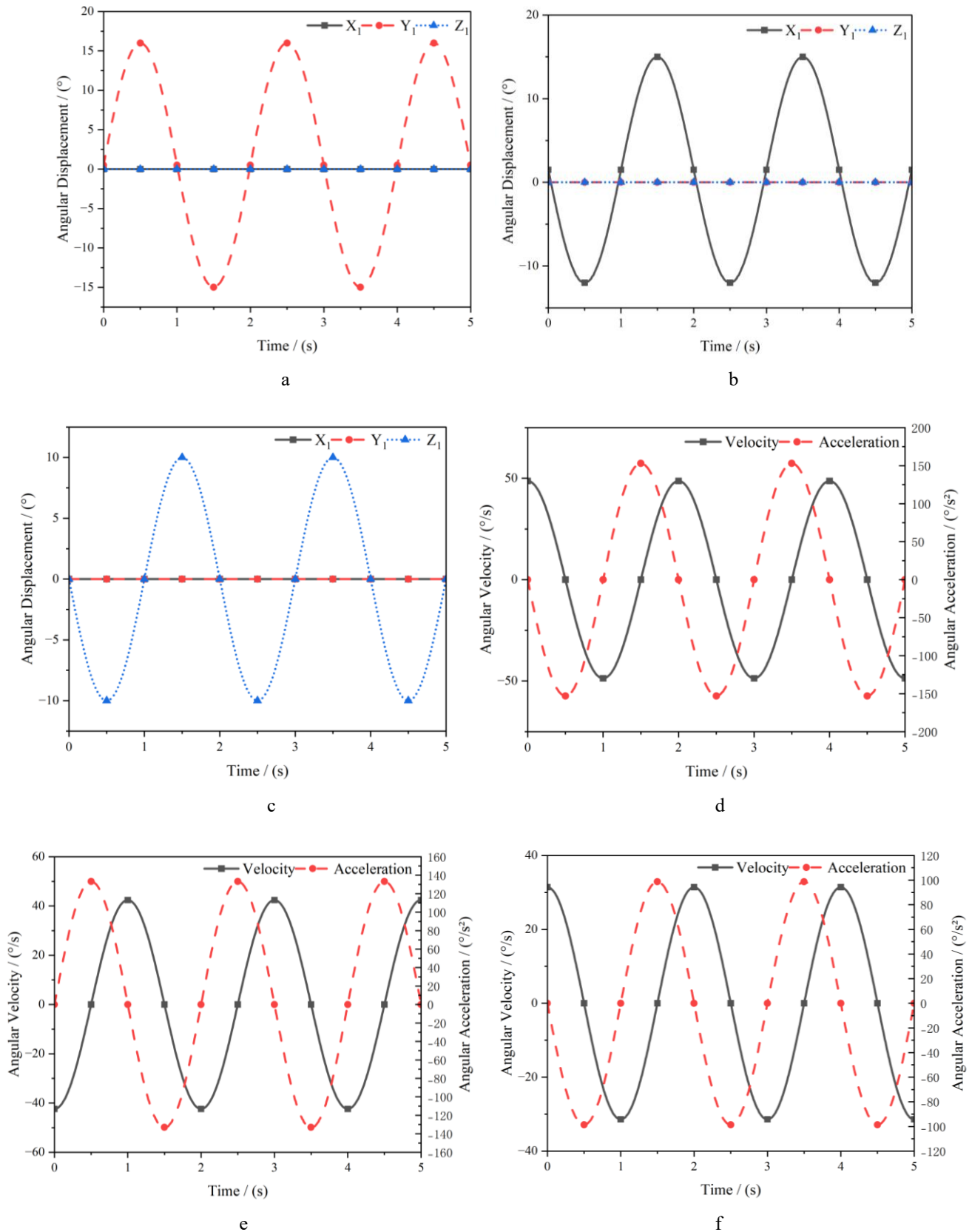
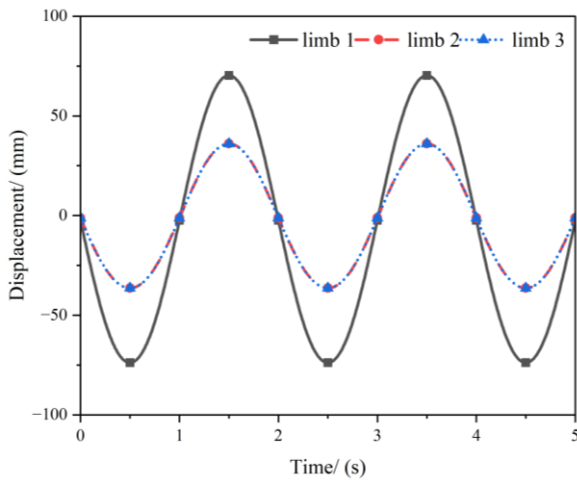
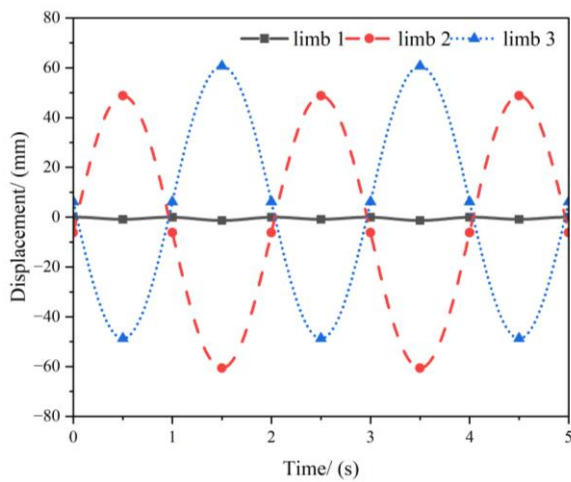


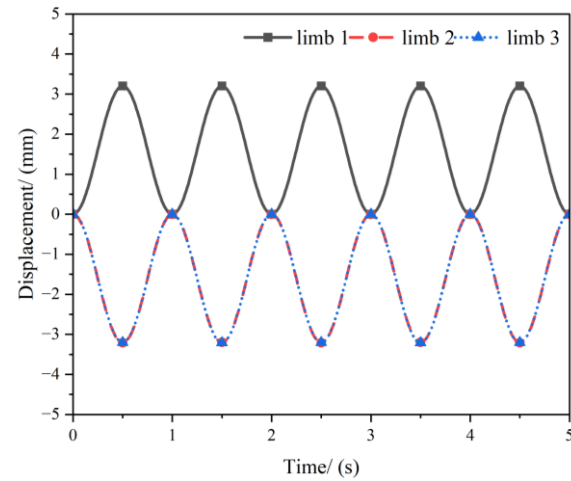
Fig. 14 Kinematic simulation results of typical motions: a – rotation angle during dorsiflexion/plantarflexion, b – rotation angle during eversion/inversion, c – rotation angle during adduction/abduction, d – angular velocity and acceleration during dorsiflexion/plantarflexion, e – angular velocity and acceleration during eversion/inversion, f – angular velocity and acceleration during adduction/abduction



a



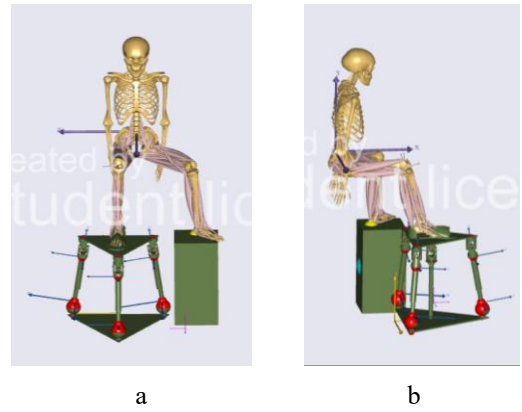
b



c

Fig. 15 Simulation results of typical motions: a – dorsiflexion/plantarflexion: limb displacement, b – eversion/inversion: limb displacement, c – adduction/abduction: limb displacement

posture adjustment and foot fixation can be completed before the mechanism is activated, thereby avoiding the need for the patient to stand on the device.



a

b

Fig. 16 Human-machine coupling model and rehabilitation posture: a – Front view, b – Side view

Table 2
Summary of human – machine coupling simulation results for six typical training motions

Training motion	Maximum muscle activity	Main responding muscles	Muscle-tendon length variation pattern
Dorsiflexion	0.234	Tibialis anterior, extensor digitorum longus, extensor hallucis longus	Anterior dorsiflexor group shortens; posterior plantarflexor group lengthens
Plantarflexion	0.112	Medial gastrocnemius, lateral gastrocnemius, lateral soleus	Posterior plantarflexor group shortens; anterior dorsiflexor group lengthens
Inversion	0.171	Tibialis anterior, tibialis posterior, flexor digitorum longus	Inversion-related muscle group shortens; lateral peroneal group lengthens
Eversion	0.150	Peroneus longus, peroneus brevis, peroneus tertius	Lateral peroneal group shortens; medial related muscle group lengthens
Adduction	0.125	Tibialis anterior, tibialis posterior, flexor digitorum longus	Medial and anterior related muscle groups shorten; lateral muscles lengthen
Abduction	0.238	Peroneus longus, peroneus brevis	Lateral related muscle group shortens; medial muscles lengthen

6.2. Summary of simulation results for six typical training motions

As shown in Table 2, the maximum muscle activity values under the six typical training motions are all below 1.0, indicating that no obvious muscle-force overload occurs under the given simulation conditions and that the training process of the mechanism has acceptable training safety. Meanwhile, the main responding muscle groups differ obviously in different training directions. Dorsiflexion and plantarflexion stimulate the anterior dorsiflexor group and posterior plantarflexor group of the lower leg, respectively. Inversion and eversion stimulate the medial related muscle group and the lateral peroneal muscle group, respectively. Adduction and abduction show differentiated participation of medial and lateral muscle groups during foot direction regulation. These results indicate that different posture

training motions output by the mechanism can induce direction-selective responses of target muscle groups in the human model.

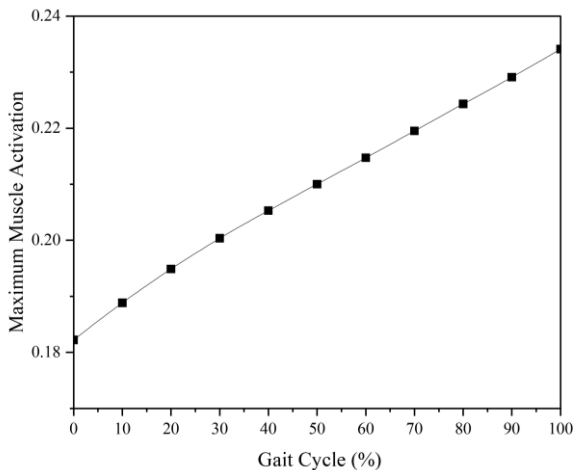
6.3. Biomechanical response analysis of typical training motions

To further illustrate the correspondence between mechanism output and target-muscle response, dorsiflexion and plantarflexion are selected as typical working conditions for graphical analysis. Dorsiflexion and plantarflexion represent early single-axis training motions in the sagittal plane of the ankle joint and can effectively reflect the muscle-group stimulation characteristics of the mechanism under different training directions.

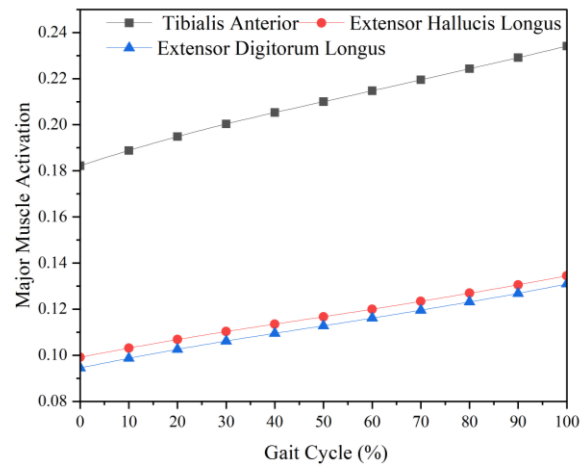
During ankle dorsiflexion training, as shown in Fig. 17, a, the maximum muscle activity increases from 0.182 to 0.234 and remains below 1.0, indicating that the muscle load induced by the mechanism is within a reasonable range and no obvious overload risk occurs. As shown in Fig. 17, b and Fig. 17, c, from the response of the main muscle groups, the activity and muscle force of the tibialis anterior change most significantly. Its activity is higher than those of the extensor hallucis longus and extensor digitorum longus, and its muscle force increases from about 133.6 N to 171.6 N, indicating that the tibialis anterior plays the main

force-generating role during dorsiflexion training, while the extensor hallucis longus and extensor digitorum longus mainly assist dorsiflexion. As shown in Fig. 17, d, the muscle–tendon length results show that the tibialis anterior and extensor digitorum longus generally shorten, whereas the lateral soleus, medial gastrocnemius, and lateral gastrocnemius lengthen. This indicates that, during dorsiflexion training, the anterior dorsiflexor group mainly contracts actively, while the posterior plantarflexor group is mainly passively stretched.

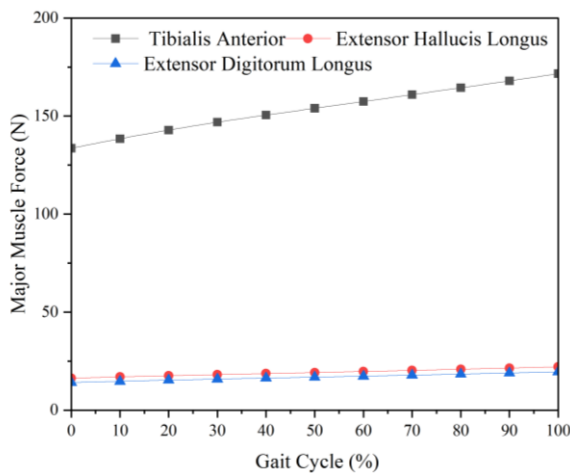
During ankle plantarflexion training, as shown in Fig. 18, a, the maximum muscle activity decreases from 0.112 to 0.086 and remains below 1.0, indicating that the muscle load induced by the mechanism is within a reasonable range and no obvious overload risk occurs. As shown in Fig. 18, b and Fig. 18, c, from the response of the main muscle groups, the activity and muscle force of the medial gastrocnemius change most significantly, followed by the lateral gastrocnemius and lateral soleus, indicating that the posterior plantarflexor group plays the main force-generating role during plantarflexion training. As shown in Fig. 18, d, the muscle–tendon length results show that the lateral soleus, medial gastrocnemius, and lateral gastrocnemius generally shorten, whereas the tibialis anterior and extensor digitorum longus lengthen. This indicates that, during plantarflexion training, the posterior plantarflexor group mainly



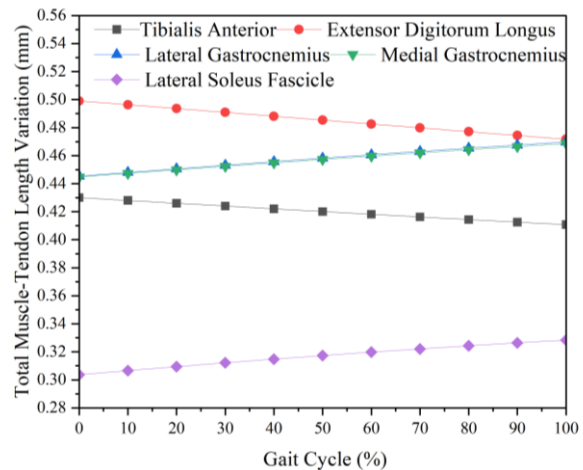
a



b



c



d

Fig. 17 Biomechanical responses during ankle dorsiflexion training: a – maximum muscle activity, b – activity of main muscles, c – main muscle force, d – total muscle–tendon length variation

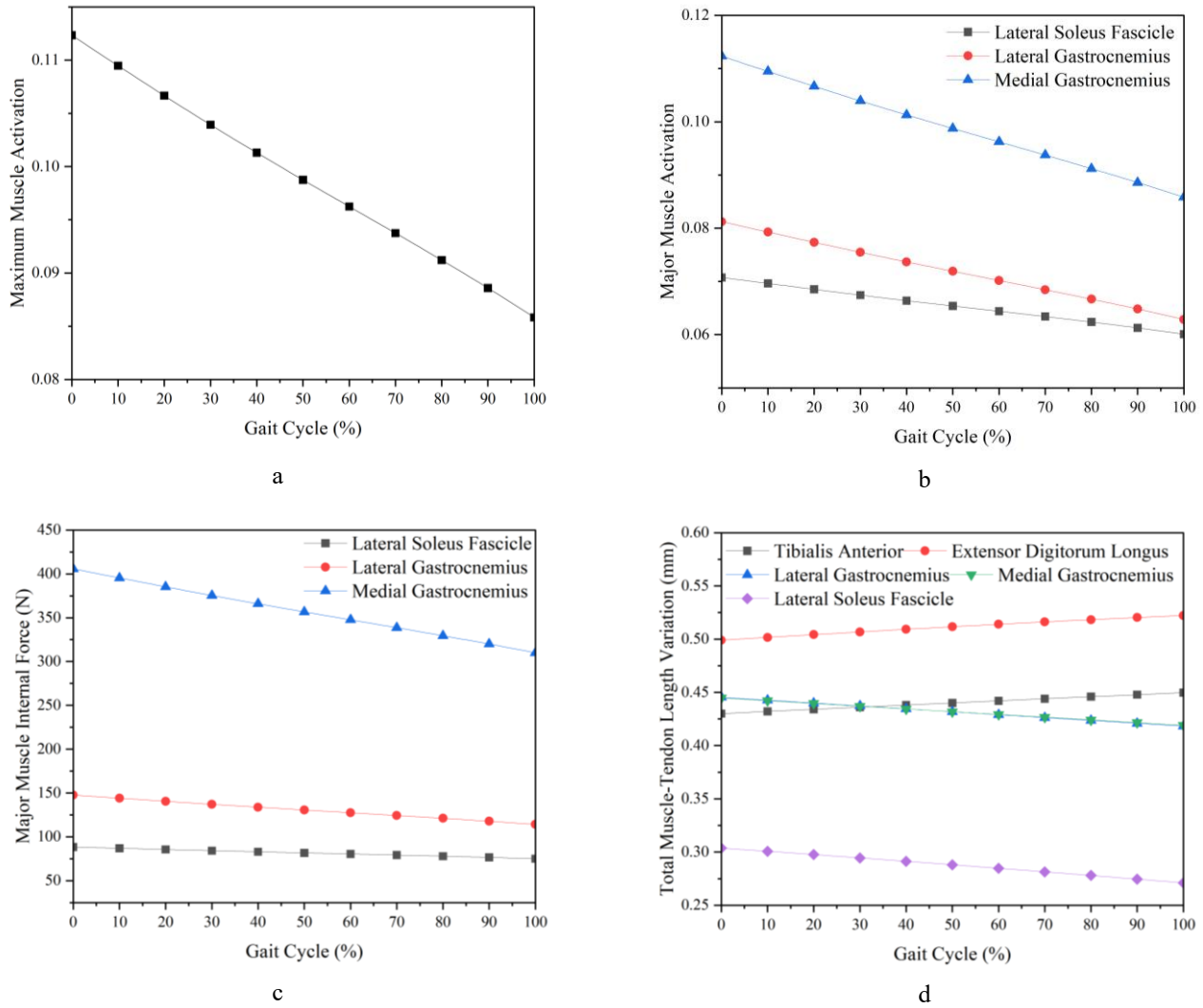


Fig. 18 Biomechanical responses during ankle plantarflexion training: a – maximum muscle activity, b – activity of main muscles, c – main muscle force, d – total muscle–tendon length variation

contracts actively, while the anterior dorsiflexor group is mainly passively stretched.

6.4. Comparative analysis

As shown by Table 2 and the simulation results of typical working conditions, ankle-related muscle responses under the six training directions show obvious direction selectivity. Dorsiflexion training mainly stimulates the anterior dorsiflexor group of the lower leg, including the tibialis anterior, extensor digitorum longus, and extensor hallucis longus. Plantarflexion training mainly stimulates the posterior plantarflexor group, including the gastrocnemius and soleus. During inversion training, the tibialis anterior, tibialis posterior, and flexor digitorum longus participate obviously, while the lateral peroneal muscle group mainly presents passive stretching. During eversion training, the peroneus longus, peroneus brevis, and peroneus tertius respond more obviously, while the medial related muscle groups mainly present passive stretching. Adduction and abduction training respectively induce the participation of medial and lateral ankle-foot related muscle groups.

From the perspective of maximum muscle activity, the maximum muscle activity values of all six training motions are below 1.0, indicating that no obvious muscle-force overload occurs under the given simulation conditions and

that the training process has acceptable training safety. From the variation of muscle–tendon length, the target muscle groups and the contralateral muscle groups show opposite trends in each training direction, indicating that the mechanism output posture can induce muscle-group coordination responses consistent with the functional anatomy of the ankle-foot complex. In summary, the proposed mechanism can realize direction-selective muscle-group stimulation under different training directions, verifying the feasibility of its application in staged ankle rehabilitation training from the perspective of human–machine coupling simulation.

6.5. Practical implementation and limitations

This study mainly verifies the feasibility of the proposed mechanism from the perspectives of mechanism configuration, kinematics, and human muscle responses, while clinical rehabilitation effectiveness under physical-prototype conditions has not yet been validated. The current theoretical model adopts ideal kinematic pairs and does not systematically consider friction, clearance, structural flexibility, or patient-specific joint resistance. Therefore, the present results mainly reflect the ideal kinematic performance of the mechanism. In practical applications, differences in patient body dimensions can be accommodated by adjusting the seat height, foot-fixation position, and the relative position

between the anatomical ankle center and the mechanism rotation center. Mode switching should be performed only after the mechanism has stopped and returned to the neutral position, with mechanical locking and mode confirmation used to avoid operating errors. Future work will establish a dynamic model incorporating actuator dynamics, joint friction, and human joint impedance, and an adjustable physical prototype and participant experiments will be developed to further evaluate actuator-force requirements, motion-tracking accuracy, ergonomic adaptability, and actual rehabilitation performance at different rehabilitation stages.

7. Conclusions

1. A parallel ankle rehabilitation mechanism composed of three UPS driving limbs and a central reconfigurable constraint chain is proposed. By switching the equivalent kinematic pair of the central joint, the mechanism can realize 1R, 2R, and 3R mode conversion.

2. Based on screw theory, inverse kinematics, and velocity Jacobian analysis, the role of the central reconfigurable constraint chain in regulating the DOFs of the mechanism is verified. The workspace of mode 3 is continuous and can meet the requirements of multi-axis compound training. The local condition index analysis shows that the mechanism has certain velocity transmission capability within the specified posture range, but the dexterity is low in some posture regions. In future work, the motion performance of the mechanism can be further improved by optimizing the hinge-point arrangement of the moving and fixed platforms.

3. The ADAMS virtual prototype simulation results show that the DOFs of the three configurations are 1, 2, and 3, respectively. The main rotation directions and limb extension laws of the three typical motions are consistent with the theoretical analysis, and all motion curves are continuous and smooth, verifying the correctness of the mechanism configuration and inverse kinematic model.

4. The AnyBody human-machine coupling simulation results show that the maximum muscle activity values of the six typical training motions are all below 1.0. The different training directions induce direction-selective responses of the corresponding target muscle groups, indicating that the proposed mechanism has acceptable training safety and muscle-group stimulation directionality.

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KINEMATIC AND REHABILITATION PERFORMANCE ANALYSIS OF A RECONFIGURABLE PARALLEL MECHANISM FOR STAGED ANKLE REHABILITATION

S u m m a r y

A reconfigurable parallel mechanism for staged ankle rehabilitation is proposed. The mechanism consists of three UPS driving limbs and a central reconfigurable constraint chain, which can switch among 1R, 2R, and 3R modes to match early, middle, and late rehabilitation training requirements. Normal-gait ankle motion is analyzed in

AnyBody, and the configuration, mobility, inverse kinematics, velocity Jacobian, workspace, and dexterity of the mechanism are investigated. ADAMS virtual prototype simulation verifies that the DOFs and limb motion laws of the three configurations are consistent with the theoretical analysis. Human-machine coupling simulation in AnyBody further shows that the maximum muscle activity values of six typical training motions are all below 1.0, and different training directions induce direction-selective responses of the target muscle groups. The results indicate that the proposed mechanism has certain training safety and adaptability for staged ankle rehabilitation.

Keywords: ankle rehabilitation, reconfigurable parallel mechanism, staged rehabilitation, kinematics, human-machine coupling simulation.

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