Proceedings of 2024 WRC Symposium
on Advanced Robotics and Automation (WRC SARA)
Beijing, China, August 23, 2024

2024 WRC Symposium on Advanced Robotics and Automation (WRC SARA) | 979-8-3315-0610-0/24/$31.00 ©2024 IEEE | DOI: 10.1109/WRCSARA64167.2024.10685657

Design and control of a novel powered wearable knee exoskeleton for lower limb rehabilitation

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***Abstract*— Lower limb rehabilitation is a crucial aspect of recovery following various injuries and musculoskeletal disorders. Traditional techniques, while effective, can be time-consuming and require intensive therapist involvement. Powered wearable knee exoskeletons offer a promising solution by providing targeted torque assistance during rehabilitation exercises. This paper presents the design and control of a novel powered wearable knee exoskeleton specifically tailored for lower limb rehabilitation applications. The exoskeleton prioritizes user comfort, safety, and ease of use. We delve into the mechanical design considerations for achieving lightweight construction, comfortable wearability, and suitable range of motion. Subsequently, we elaborate on the control algorithm employed on the knee exoskeleton, focusing on its ability to improve rehabilitation efficiency.**

I. INTRODUCTION

Lower limb rehabilitation is a critical aspect of recovery following a wide range of injuries and musculoskeletal disorders, including stroke, ligament tears, and osteoarthritis [1]. These conditions, which can affect individuals of all ages, can significantly limit mobility, reduce quality of life, and impose a substantial burden on healthcare systems. Traditional rehabilitation techniques, such as physiotherapy and therapeutic exercises, while effective, often require intensive therapist involvement and can be time-consuming [2]. This can lead to challenges with patient adherence and limit the frequency and intensity of rehabilitation sessions.

Powered wearable knee exoskeletons offer a promising solution to address these limitations and enhance lower limb rehabilitation. These innovative assistive devices can be worn by the user and provide targeted torque assistance at the knee joint. This assistance can facilitate repetitive movement practice during rehabilitation exercises, improve gait patterns for individuals with impaired mobility, and reduce stress on the injured or weakened muscles, allowing for longer training sessions. Developing effective knee exoskeletons requires careful consideration of both mechanical design and control strategies.

The design of the knee exoskeleton needs to strike a delicate balance between functionality, user comfort, and ease of use. Ideally, the knee exoskeleton should be: (1) lightweight and compact, minimizing additional weight on the user's limb and allowing for a wider range of natural leg movement patterns.(2) adjustable and customizable, adaptable to fit a variety of leg sizes and morphologies to ensure proper user alignment, able to adjust range of motion and assistive torque levels. (3) safe and reliable, prioritizing user safety by incorporating features such as emergency stop mechanisms, joint range-of-motion limitations. Singh *et al.* [3] developed an assistive four-bar linkage knee exoskeleton using the shape synthesis methodology to obtain new shapes of thigh and shank attachments. Kim *et al.* [4] developed a robot exoskeleton for knee joint rehabilitation therapy. The robot was designed with a kinematic structure to make its joints to have the same range of motion (ROM) as that of human knee joints. Sarani *et al.* [5] proposed a novel mechanical design for a knee rehabilitation portable exoskeleton. The design was based on a lever mechanism with an adjustable pivot which moves along a straight line in the radial direction without any linear guide. The design feasibility of the robot was only investigated via simulation.

The control system of the knee exoskeleton plays a critical role in determining the effectiveness and safety of the rehabilitation intervention. It needs to be able to: (1) accurately interpret the user's desired movement to provide appropriate assistive torque at the right time. (2) dynamically adjust the level of assistance based on the user's progress, rehabilitation goals, and real-time feedback from sensors. (3) smoothly and responsively provide torques to facilitate human-robot interaction. Beyl *et al.* [6] proposed a trajectory-based controller by means of proxy-based sliding mode control as well as a torque controller minimizing the interaction during unassisted walking for a powered knee exoskeleton. Ahmed *et al.* [7] implemented three control strategies based on MPC, LQR and PID that were applied to knee joint of lower limb exoskeleton model for passive exercise. The performance of these controllers were analyzed in simulation instead of settled on real hardware.

At present, products such as Japan's HAL [8] and Israel's Rewalk [9] represent the advanced level of lower limb rehabilitation exoskeletons. Although they perform well in rehabilitation training, the large equipment weight is an important factor restricting their usage. The equipment weights of HAL and ReWalk are both 23kg, which restrict the movement of the wearer and increases unnecessary physical exertion. On the other hand, the comprehensive use of multi-dimensional sensors has become a trend in the research and development of lower limb rehabilitation exoskeleton systems. ReWalk's perception system only relies on the inertial sensor, while

This work was supported by Emerging Interdisciplinary Platform for Medicine and Engineering in Sports (EIPMES) and the Science and Technology School-strengthen Project of CUPES, No.1552230XX.

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HAL uses a combination of electromyographic sensors and plantar pressure sensors.

This paper presents the design and control of a novel powered wearable knee exoskeleton specifically tailored for lower limb rehabilitation. The device is integrated with several types of sensors. We will delve into the mechanical design considerations for achieving lightweight construction and comfortable wearability. The actuation system, joint structure, and weight minimization of the designed device will be introduced in Sec. II in detail. Subsequently, we will elaborate on the control algorithm employed by the knee exoskeleton in Sec. III, focusing on its ability to enhancing the rehabilitation efficiency. We also present the results of preliminary evaluations and discuss the future work for this knee exoskeleton technology in lower limb rehabilitation. Finally, in Sec. IV, the paper will conclude by summarizing the key features of the developed knee exoskeleton.

II. KNEE EXOSKELETON SYSTEM DESIGN

Instead of constructing a self-contained version directly, we have developed a power-tethered knee exoskeleton system. This facilitates laboratory-based hardware design and controller testing. As shown in Fig. 1, the general exoskeleton hardware system designed in this work includes a knee exoskeleton device, a host computer, a DC power supply, and communication equipment. We will describe these components in detail in the following.

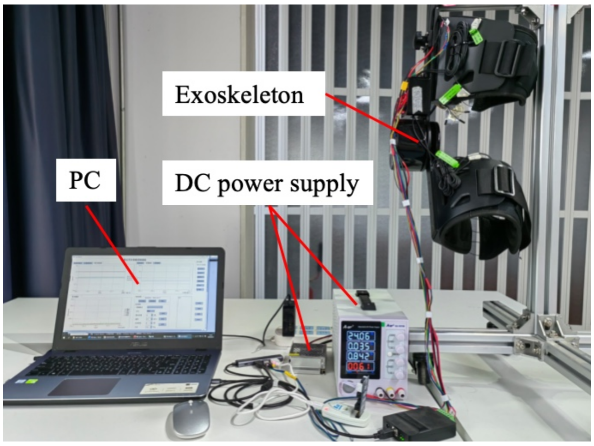


Fig. 1: The general structure of the knee exoskeleton system.

A. Hardware Design

The knee joint exoskeleton structure proposed in this study aims to ensure functionality, user comfort and minimize interference with human movement. The key design rationales for the knee exoskeleton are:

- **Miniaturization and lightweight construction:** A compact and lightweight exoskeleton design enhances wearer comfort and minimizes restrictions on natural movement patterns.
- **Prioritizing flexion-extension movement:** Knee flexion-extension movements are essential for daily activities and are disproportionately affected by injuries. Therefore, the exoskeleton focuses on assisting these movements.
- **Fixed-axis rotation joint configuration:** A fixed-axis rotation joint structure promotes simplicity and compactness, aligning with the exoskeleton's design goals.

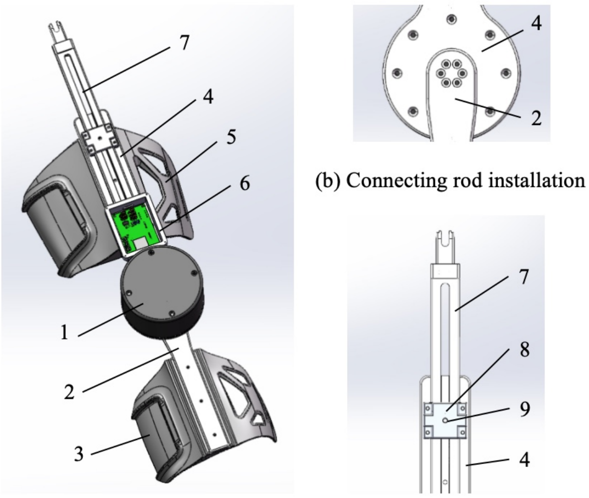
Following these design principles, we developed a compact, lightweight, user-friendly knee exoskeleton prototype. It is mainly composed of mechanical structure, joint unit, control board and sensor modules.

1) Mechanical structure: The knee joint has the movement forms of flexion, extension, inversion, eversion, internal rotation, and external rotation. The range of flexion and extension is 3˜4 times that of other movement forms [10]. Moreover, the number and level of activation of leg muscles during flexion and extension are higher. Knee injuries have a greater impact on flexion and extension [11]. Therefore, in this work, the movement form of the exoskeleton is selected as flexion and extension of knee joint in the coronal plane. Fixed-axis rotation exoskeleton joint configuration is considered to ensure the simplicity and compactness of the structure.

The exoskeleton mechanics consists of a thigh cuff, a calf cuff, and a connecting rod, as displayed in Fig. 2. The output shaft of the DC motor coincides with the knee joint flexion and extension axis, on which the calf connecting rod is installed. The calf binding assembly is installed at the end of the calf connecting rod by bolts. The thigh connecting rod is fixedly connected to the motor stator, on which the thigh binding part, the control chip box and the extension rod are installed. The extension rod can move in the slot on the thigh connecting rod. An adjustable screw is used to lock the relative position of them. This design makes the exoskeleton compact to minimize additional weight on the wearer's limb and allow for a natural range of motion during exercises. A prototype was manufactured according to the above design model, as shown in Fig. 3.

Apart from being compact, the knee exoskeleton is supposed to be lightweight as well. This can be achieved by using lightweight materials like high-strength aluminum alloys or carbon fiber composites. In fact, the supporting structural parts such as the thigh connecting rod, calf connecting rod and extension rod are all made of 7075 aviation aluminum alloy with a tensile strength of 524 MPa and a density of 2.78 g/cm^3. The average width of the three rods is 31.3 mm and the average thickness is 7 mm. Weight-reducing holes are opened at non-important force-bearing locations to reduce the weight of the equipment. The material of the thigh binding, calf binding and control chip box is glass fiber nylon with a density of 1.36 g/cm^3. Through the systematically lightweight design, the total weight of the knee rehabilitation exoskeleton is 1.85kg, while the average weight of a single leg for adult men and women is 15.3kg and 12.6kg respectively [12]. The load introduced by the exoskeleton is equivalent to 12.1% ˜ 14.7% of the weight of a single leg.

The exoskeleton should be adaptable to fit a variety of leg sizes and morphologies. Adjustable straps and padding



(a) Knee exoskeleton (left leg)

(b) Connecting rod installation

(c) Extension rod mechanism

1. Motor 2. Calf connecting rod 3. Calf binding
4. Thigh connecting rod 5. Thigh binding 6. Control chip box
7. Extension rod 8. Pressing piece 9. Locking bolt

Fig. 2: 3D Model of Knee Exoskeleton.

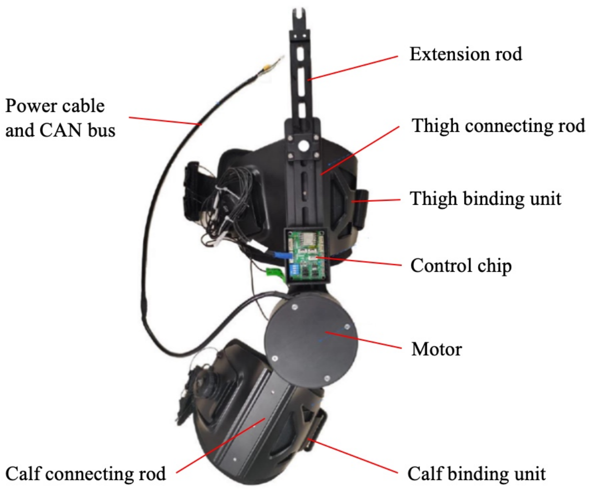


Fig. 3: Knee exoskeleton prototype

can ensure proper user alignment. The range of motion and assistive torque levels should also be adjustable to tailor the intervention to the specific needs and progress of each patient. Therefore, the binding is equipped with locking parts. Both turnbuckles and Velcro are adopted to achieve ease of wearing and adjustability.

Considering that patients with knee injuries have high requirements for the comfort of wearing equipment during the rehabilitation process, and there are often wearing scenes that require direct skin contact, an ergonomically designed 3D flexible cushion is used as a close-fitting wearable part, and the straps are designed with a mesh to increase breathability.

The knee exoskeleton proposed in this paper is mainly applicable to low-speed and low-intensity rehabilitation exercises. We further tested the mechanical structure performance under various stress conditions, including walking at a constant speed for 10 minutes on a treadmill, jogging for 5 minutes. The results show that the exoskeleton has no interference with the human body during exercise, and the mechanical structure is stable and reliable.

2) Joint unit: A DC motor located in the thigh cuff provides torque assistance at the knee joint. The motor specifications are listed in Table I Encoder is integrated into the joint unit to measure knee joint position and velocity.

TABLE I: Motor Specifications

Item	Specification
Rated Voltage	24VDC
Continuous Current	4.6A
Rated Speed	100rpm
Maximum Speed	110rpm
Rated Torque	6Nm
Maximum Torque	12Nm
Reduction Ratio	24.33:1
Weight	380g

3) Control board: The AEXO_SAS A22082 main control board, designed based on the STM32 system board, is the main device for sensor data processing and motor control of the exoskeleton system. Fig. 4 displays the layout of the control board and how it is connected to other components.

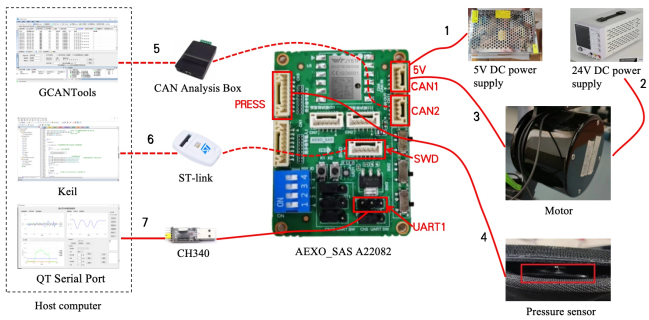
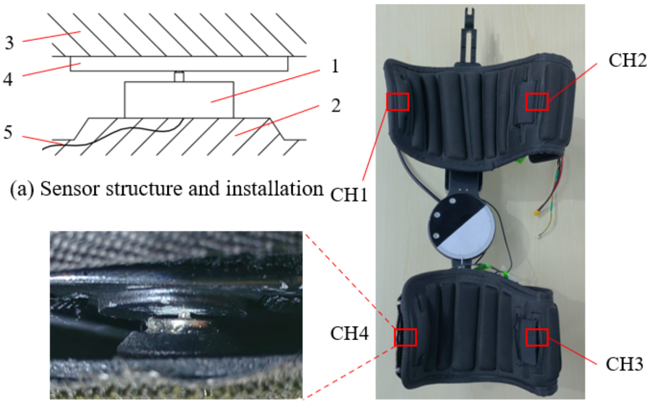


Fig. 4: The layout of the control board.

AEXO_SAS A22082 uses a 5V DC power supply while the the motor is powered through a 24V DC power supply. The load cells are connected to and powered by the control board. The collected voltage signal is amplified 330 times through the operational amplifier circuit. The motor is connected to the CAN bus through which motor states and control commands can be transmitted. The CH340 module is used to perform serial port communication between the host computer and the main control board. The other components are for debugging.

4) Sensor modules: Diverse sensor modules are necessary for the human-robot interaction, rehabilitation program personalization and objective monitoring of patient progress. Different types of sensors are integrated in the designed knee exoskeleton device. The encoder and Hall sensor integrated into the motor can collect data of motor position, rotation speed and current of motor output shaft (output torque).

The main control board integrates a micro 9-axis IMU sensor to obtain the three-axis angular velocity, acceleration and Euler angle. Moreover, four miniature pressure sensors are embedded inside the wearable cushion to measure the interaction force between the exoskeleton and the patient's leg. They located on the front and back of the thigh, the front and back of the calf. The distribution of the four pressure sensors are displayed in Fig. 5.



1.Pressure sensor 2. Binding parts 3. 3D cushion 4. Insulation sheet
5. Signal cable CH1. Pressure sensor on the front of the thigh
CH2. Back of the thigh CH3. Back of the calf CH4. Front of the calf

Fig. 5: Distribution of the load cells.

After sensors installed, there is a pre-pressure between the binding and the cushion, which causes the output of the pressure sensor to be non-zero when no-loading. Therefore, the force sensors need to be calibrated. The voltage generated by the strain gauge inside the sensor is linearly related to the loaded weight. Therefore, the weight W_M measured by the sensor and the actual weight W_L have the relationship as follows:

$$W_M = aW_L + b \tag{1}$$

where a and b are the calibration coefficients. Arranging the weights of 1000 g, 500 g, 200 g, 50 g, and 50 g into 24 loads:

$$L = (l_1, l_2, ..., l_{24}) \tag{2}$$

Taking the average of 950 weighing data for each load, and get the measured weight sequence:

$$M = (m_1, m_2, ..., m_{24}) \tag{3}$$

We get 24 pairs of loading weight and measured weight. The calibration coefficients a and b will be obtained by performing linear fitting algorithm on this 24 points. We calibrated all the four channels of force sensors. The fitting curves are shown in Fig. 6. The calibration results are shown in Table II.

B. Software Design

The software should not only receive the data from the sensor in real time, but also accurately control the movements of the exoskeleton. In addition, the exoskeleton system used

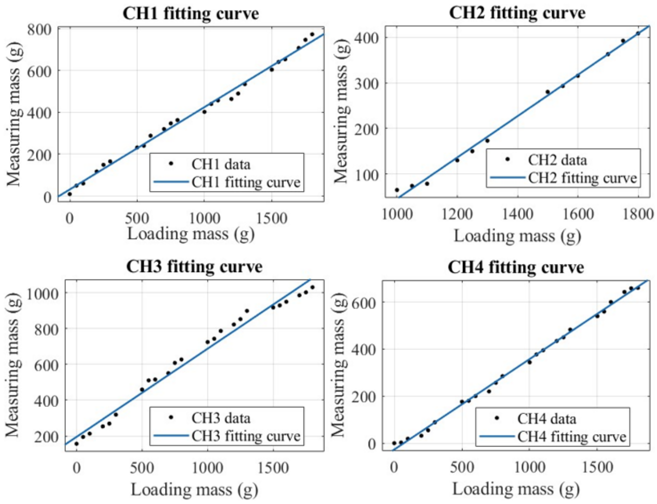


Fig. 6: Curve fitting for force sensor calibration.

TABLE II: Calibration coefficients of pressure sensors

	CH1	CH2	CH3	CH4
a	0.3935	0.4537	0.4923	0.3843
b	31.43	-408.4	195	-26.86

for rehabilitation treatment requires a convenient human-computer interface for rehabilitation therapists. The human-computer interface should be easily modified and optimized based on the therapist's feedback. In this work, the software designed for the knee exoskeleton system hierarchically includes three parts: the underlying drive controller, the middle signal processing program and the high-level human-computer interface.

The underlying driver implemented a three-loop servo motor controller based on PID feedback control. The middle signal processing program was implemented and run on the AEXO_SAS A22082 main board. It is in charge of collecting sensor data and transmitting the data to the host computer in real time. The frequencies of force, IMU and motor information are 67 Hz, 25 Hz and 50 Hz, respectively. In the middle program, we incorporated joint range-of-motion limitation as a soft stop mechanism to ensure safety of the wearer. Namely, when the flexion and extension range of the human knee joint comes to the predefined thresholds, the joint motor is disabled.

The high-level human-computer interface was implemented based on the Qt 5.12, as shown in Fig. 7. There are three dynamic curve display areas: for IMU data curves (with three tabs displaying acceleration / angular velocity / Euler angle data respectively), for pressure data curve (four channels shown in the same tab), for motor data curve (with three tabs displaying motor speed / position / torque). The therapist can monitor the status of the patient through this interface.

The therapist is supposed to intuitively operate the knee skeleton. For this aim, we designed the motor setting component including motor activation, motor deactivation and standby. The knee joint can be controlled under three modes:

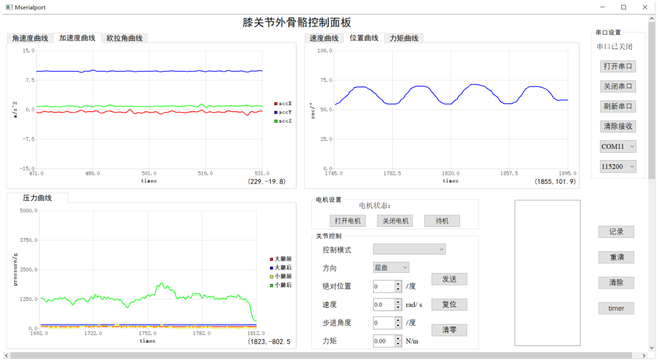


Fig. 7: The high-level human-computer interface.

speed mode, position mode and torque mode. In speed mode, the therapist can assign a specific velocity and motion direction (flexion or extension). In position mode, the therapist can specify a desired angle to reach. In torque mode, the assistive torque (with direction) command is directly sent to the motor.

III. KNEE EXOSKELETON CONTROL

A. Control Strategy

The control system of the knee exoskeleton plays a critical role in determining the effectiveness of the rehabilitation intervention. Normally, a good control system should be able to accurately interpret the user's desired movement through various sensors and provide appropriate assistive torque at the right time. Adaptive assistance is also necessary to dynamically adjust the level of assistance based on the user's progress and rehabilitation goals for personalized rehabilitation programs. In this work, we developed a simple control strategy called "therapist teach and replay" that focuses on improving rehabilitation efficiency.

In the rehabilitation treatment for knee injuries, repetitive knee flexion and extension is an important rehabilitation action. It requires the rehabilitation therapist to use his experience to bend the patient's calf back and forth at a specific speed. Due to individual differences in patients, the speed and limit position of the reciprocating movement are also different. If the recorded trajectory can be accurately replayed using the exoskeleton system, this will not only help ensure the consistency of rehabilitation movements, but also help the rehabilitation therapist save unnecessary physical exertion. For this aim, we implemented the "therapist teach and replay" controller: the trajectory was recorded as a therapist manually guided the subject's legs to achieve a desired gait pattern. The recorded kinematics was then replayed to generate a participant-specific stepping trajectory by using a PID controller.

Assuming the recording frequency is f_n, the recorded position sequence is:

$$P_n = (x_1, x_2, ..., x_n) \tag{4}$$

Down-sampling the position sequence P_n:

$$f_m = 0.2 f_n \tag{5}$$

Then the position sequence obtained is:

$$P_m = (y_1, y_2, ..., y_m) \tag{6}$$

This is equivalent to taking a value from P_n every 5 points. So the elements in P_m and P_n have the following relationship:

$$y_m = x_{5m-4} \tag{7}$$

In P_m, the difference between two adjacent positions divided by the interval time $(1/f_m)$ equals the average velocity ω:

$$\omega_m = (y_{i+1} - y_i) f_m \tag{8}$$

Then the sampling velocity sequence is obtained:

$$V_m = (\omega_1, \omega_2, ..., \omega_m) \tag{9}$$

Finally, the obtained P_m and V_m are sent to control the knee joint at the frequency of f_m under the position mode.

B. Experiment

In order to implement the "therapist teach and replay" controller, we added "Record" and "Replay" buttons in the human-computer interface. By clicking the "Record" button, the position and speed data of the motor can be recorded. By clicking the "Replay" button, the knee exoskeleton can drive the human leg to move following the recorded trajectory. In the experiment, we manually bent the knee joint according to the rhythm of "flexion-extension-flexion-extension-flexion". We did the experiment under two conditions for comparison: no-load condition and under load condition (with human leg wearing the knee exoskeleton). Fig. 7 shows the subject wearing the device.

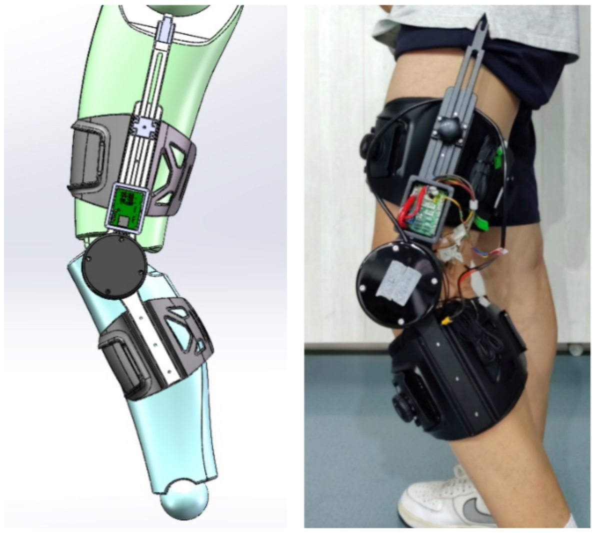


Fig. 8: Knee exoskeleton wear demonstration.

C. Results and Discussion

Fig. 9 displays the comparison between the replay results and the recorded data when no-loading. It can be seen from the figure that the recorded trajectory and the replay trajectory overlap well. The difference between recorded position and replay position at the same time is taken as the position error. The absolute value of the average position error is 1.35°. Similarly, the recorded and replayed speed curves overlap well, and the average speed error is 0.09 rad/s.

Fig. 10 displays the comparison between the replay results and the recorded data under load condition. The average position error is 1.13°, and the average speed error is 0.08 rad/s. It can be seen that under interference (human body load), the exoskeleton system is still able to accurately reproduce the prescribed trajectory.

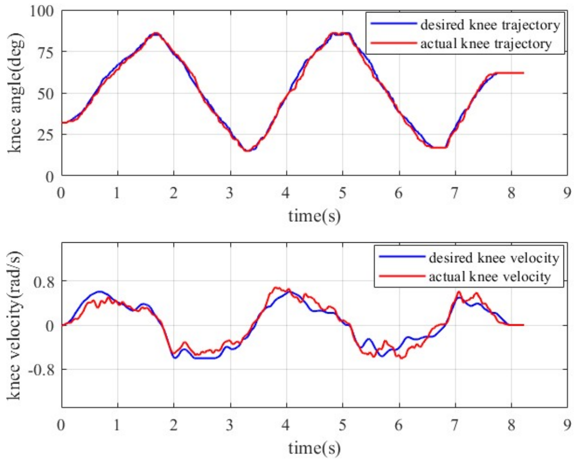


Fig. 9: Controller performance under no-load condition.

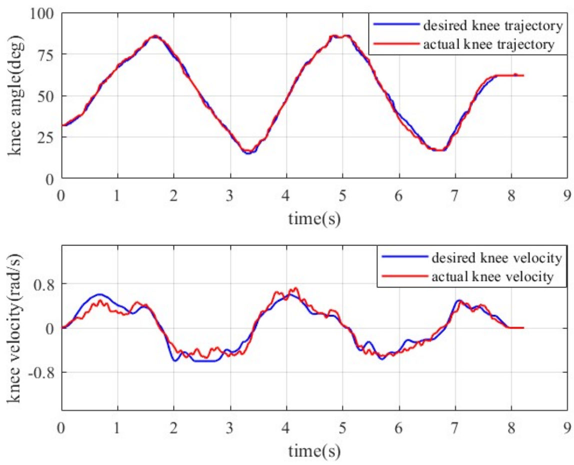


Fig. 10: Controller performance under load condition.

Although the "therapist teach and replay" controller is able to execute the therapist's experience with high accuracy, it can only generate a fixed reference trajectory. In the future, we will implement smooth torque controller based on impedance control to allow flexible adaptation and to facilitate natural human-robot interaction. Moreover, we will take advantage of the real-time feedback from encoders, force sensors, and potentially EMG sensors (measuring muscle activity) to continuously monitor user intent and adjust the control parameters to optimize assistance and ensure user safety.

IV. CONCLUSIONS

In this paper, we designed a novel powered knee exoskeleton system. This exoskeleton can meet the patient's rehabilitation requirements for lightweight, comfortable and intelligent wearable devices. Based on ergonomic binding design, it can enhance the comfort of the wearer. In addition, the collection of multi-dimensional sensor data helps to accurately describe the wearer's movement state, which is crucial for rehabilitation therapists to evaluate the patient's recovery progress. The control algorithm designed based on the exoskeleton hardware system can not only display the movement data collected by the sensor in real time, but also accurately control the movement of the exoskeleton. The designed human-computer interaction interface reduces the difficulty for rehabilitation therapists to operate the exoskeleton system. Finally, the feasibility of the exoskeleton system was verified in the trajectory record-and-replay experiment.

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