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Research Article

Development and Evaluation of a Rehabilitation Wheelchair with Multiposture Transformation and Smart Control

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Stroke and other neurological disorders have an effect on mobility which has a significant impact on independence and quality of life. The core rehabilitation requirements for patients with lower limb motor dysfunction are gait training, restand, and mobility. In this work, we introduce a newly developed multifunctional wheelchair that we call “ReChair” and evaluated its performance preliminarily. ReChair seamlessly integrates the mobility, gait training, and multiposture transformation. ReChair driving and multiposture transformation are done using the voice, button, and mobile terminal control. This work describes the functional requirements, mechanical structure, and control system and the overall evaluation of ReChair including the kinematic simulation of the multiposture transformation and passive lower limb rehabilitation training to quantitatively verify the motion capability of ReChair, the voice control system evaluation that shows how the voice recognition system is suitable for home environment, the sensorless speed detection test results that indicate how the wheel speeds measured by sensorless method are appropriate for travelling control, and the passive and balance training test results that show how the lower limb rehabilitation training in daily life by ReChair is convenient. Finally, the experimental results show that ReChair meets the patients’ requirements and has practical significance. It is cost-effective, easy to use, and supports multiple control modes to adapt to different rehabilitation phases.

1. Introduction

The aging population trend is accelerating in the world. According to “The World Population Prospects 2019: Highlights,” released by the United Nations, one in six people in the world will be over the age of 65 (16%) by 2050, compared with one in 11 in 2019 (9%). The share of the population aged 65 years or more is projected to double between 2019 and 2050 in most countries [1]. Stroke, spinal cord injury (SCI), and other neurological disorders are often related to aging [2] and have an effect on various functions necessary for daily life, especially mobility which has a significant impact on independence and quality of life. The relationship points out that a larger number of seniors are expected to be affected by lower limb motor dysfunction. Additionally, stroke or SCI patients who stay in bed for a long time are at risk for secondary problems including pressure ulcers, obesity, diabetes mellitus, osteoporosis, and other chronic health conditions that increase the risk of mortality [3].

The good news is that lower limb motor function can be improved through rehabilitation training. An effective training method is to conduct high-intensity repetitive tasks [4]. Conventional lower limb rehabilitation training relies on manual exercises, in which the physiotherapist and the patient must be in a one-to-one interaction. However, as the amount of lower limb motor dysfunction patients is increasing, the number of physiotherapists is far from being sufficient [5]. This shortage leads to limited therapy sessions for the patients and physical exertion of the physiotherapist. In order to alleviate the physical burden on physiotherapists and maximize independent function of stroke or SCI patients, numerous lower limb rehabilitation and assistive devices have been proposed. Lower limb rehabilitation devices are typically divided into body weight support training devices (BWSTDs) and lower limb exoskeletons (LLE) [6]. The basic components of a BWSTD are a body weight support system and the LLE linked to a treadmill or movable footplates. The significant size and cost of BWSTDs usually result in limited applications. Additionally, they are all installed in specialized facilities such as hospitals and clinical centers.

To provide daily assistive systems and substitute the residual neurological deficits that prevent the return to upright bipedal ambulation, LLEs are applied in the course of daily rehabilitation [7–9]. LLEs are usually able to carry the person wearing them, providing walking assistive force and keeping the body balance. There are several wearable LLEs currently in development and early clinical evaluation. The LLE ReWalk developed by Argo Medical Technologies provides powered hip and knee motion to enable individuals with SCI to stand upright, walk, turn, and climb stairs up and down. The ankles are passive and use a double-action orthosis with limited motion and spring-assisted dorsiflexion. The walking control is realized through subtle trunk motion and changes in the center of gravity. It is the first exoskeleton to receive FDA clearance for personal and rehabilitation use in the United States [10]. Researchers from Vanderbilt University proposed a powered LLE that provides assistive torques at both hip and knee joints for SCI individuals. A built-in carbon fiber ankle-foot orthosis (AFO) was used to provide ankle stability and transmit the weight of the LLE to the ground. The developed LLE was called Indego in the following research studies [11, 12]. The system developed by Ekso Bionics consists of electric motors to power the movement of hip and knee joints, passive spring-loaded ankle joints, footplates on which the user stands, and a backpack that houses a computer, battery supply, and wired controller [3]. It is necessary for Rewalk, Indego, and Ekso to use supplementary walking aids, such as crutches, to balance and change direction. REX (Rex Bionics PLC, London, UK) is the only commercialized self-stabilizing LLE that requires no supplemental upper body support to balance [13].

The large size and significant cost of existing BWSTDs limit their daily life applications. On the other hand, the level of safety, endurance, and wear comfort leads to limited use of LLEs [14]. The core rehabilitation requirements of patients with lower limb motor dysfunction are gait training, multiposture transformation, and mobility. BWSTDs and LLEs are not irreplaceable means to meet the above requirements. The wheelchairs are still the most used devices for patients with lower limb motor dysfunction. Various research studies about different aspects of powered wheelchairs have been conducted [15–21]. With the development of human-computer interface hardware and sensor processing algorithms, the smart wheelchairs give people with lower limb motor dysfunction not only mobility but also the necessary rehabilitation and daily living help [22–24]. Doyoung and colleagues proposed a wheelchair integrated lower limb rehabilitation system [25] that is composed of an electrical wheelchair, a lifter, and an LLE. The gait training and transportation of the patients are combined by an LLE that is based on a wheelchair. This provides a new train of thought for rehabilitation of patients with lower limb motor dysfunction. However, the gait training of this system relies on the LLE, and this is complex and difficult to use in daily life. Therefore, a mobile device which has the multiposture transformation function to avoid pressure ulcers and that combines the gait training system in daily life would be beneficial to the patients with lower limb motor dysfunction.

We have presented a technical note and gave a general introduction of the voice-controlled wheelchair in previous work [26]. In this work, we present the development of a multifunctional wheelchair (ReChair) that seamlessly integrates the mobility, gait training, and multiposture transformation in daily life conditions detailly. Compared with BWSTDs of lower limb rehabilitation in hospitals, ReChair provides the benefits of reduced efforts of the nursing staff, being time saving and the ability to be used at home. Compared with LLEs, ReChair is easier and safer in daily use. Firstly, we introduce the main design concepts and technical characteristics of ReChair. Secondly, the detailed specifications of each module including mechanical structure, kinematic simulation, and control system are described. Thirdly, we present the experimental tasks and evaluation of ReChair from different perspectives. The system is discussed and compared with previous related work. Lastly, a brief conclusion and the plan for future studies are given in the final part.

2. Materials and Methods

2.1. Functional Requirements of ReChair. ReChair is the combination of the lower limb rehabilitation training device and powered wheelchair. Four main functional requirements should be satisfied. First of all, ReChair has the basic functions of a powered wheelchair. When the lower limb rehabilitation training is not conducted, ReChair is used as an ordinary powered wheelchair both indoors and outdoors. The electric drive and multiposture transformation function are necessary. These functions are assistive to lower limb rehabilitation training, such as balance control training during standing posture and passive training during sitting or lying position. Then, multiple lower limb training modes are needed for different rehabilitation procedures. The training modes mainly include active and passive training. Active training is realized using the following method. During standing posture, the patient moves the center of gravity left and right. Pressure sensors on the footplate collect information about the center of gravity to control the bird's left and right movements in the game. When the bird eats the fallen gem, the patient scores. Passive training is carried out during sitting or standing posture to drive lower limb to simulate walking motion or planned trajectory. Finally, multiple control methods are combined in ReChair. The voice, button, and mobile terminal (mobile phone or smart tablet) control are used for driving and multiposture transformation of ReChair. Multiple control methods meet different requirements of the patients. Patients and nursing staff can choose the appropriate way to operate the ReChair in a convenient way.

2.2. Mechanical Design. ReChair is designed based on the idea of modularization. The system design is shown in Figure 1. The basic components of ReChair are frame module, linear drive module, lower limb rehabilitation training module, cushion module, armrest module, and back module.

In order to ensure stability, the frame uses the rear-wheel drive method. Two drive motors (Motion Technology Electric & Machinery Co., Ltd. Taiwan) whose rated voltage is 24 V and power is 320 W are attached to the left and right of the frame for the wheelchair movement. Two 12-Volt batteries for the whole power system are installed in the back of the frame. This arrangement is useful for the backward movement of the center of gravity, which prevents rollover and rear-tipping of the wheelchair. The lower limb rehabilitation training module is connected to the seat module by the connection between the shank and the seat. The linear actuator is used to drive the back and forth motion of the shank. The lower limb passive training is realized by the cooperation of the back/forth and up/down motors of the shank. Force sensors are installed in the footplate for balance training. The front antitilt wheel under the footplate is used as driving safety assistance. The linear drive module mainly includes eight motors for multiposture transformation and lower limb rehabilitation training. The wheelchair can be used for standing, sitting, and lying positions' transformation. Passive training activated by the shank or thigh is also realized by corresponding motors. When the thigh up/down motor moves, it can drive the thigh joint to rotate around its fixed rotation center, so as to lift the patient's thigh and drive the patient to perform lower limb rehabilitation training. The thigh bondage is used to secure the patient's thigh and prevent the patient from slipping while in standing posture. The seat cushion and the wheelchair frame are connected by fixed hinges. When the wheelchair is in the sitting position, the slot on the cushion can be stuck downwards on the side bar of the wheelchair frame so that the frame can support the cushion. The height and the front and rear position of the arm pad are adjustable. The front/back and left/right distance of the controller from the arm pad can also be adjusted to suit different human bodies. The left and right handrails can be rolled backward and removed if needed. These adjustable designs help the patients to get in and out of the wheelchair easily.

Since the wheelchair is mainly used in patients with lower limb dysfunction, the safety and stability of the mechanism are very important. The motion mechanism of the ReChair is briefly shown as Figure 2. The stand-up posture was realized by the linear displacement of the motor and the clockwise rotation of the cushion and back modular. The thigh up and down motions were driven by linear motor and the rotation of the cushion. The linear motor retracts and back modular rotates to realize the transformation from sit to back lie. The linear motor drives the lower limb rehabilitation training modular to realize the shank up and down. The shank back and forth was achieved by the linear displacement of the linear motor parallel to the shank. In order to prevent accidents, the uniqueness of the mechanism motion must be verified. The conditions that the mechanism has definite motion are mechanism DOF is greater than zero and the number of driving link is equal to the mechanism DOF [27]. The planar motion mechanism has three DOFs when the component is not connected with other mechanism. The lower pair introduces two constraints, and the higher pair brings one.

The calculating formula of mechanism DOF is described as follows:

$$F = 3n - 2p_l - p_h, \tag{1}$$

where F is the mechanism DOF, n is the number of moving link, p_l is the number of lower pair, and p_h is the number of higher pair. We calculate the mechanism DOF of each motion mechanism from Figure 3. All the mechanism DOF is equal to one and the same with the number of driving link. The uniqueness of the motion mechanism of the ReChair is verified.

$$\begin{aligned}
F_{3(a)} &= 3 \times 3 - 2 \times 4 - 0 = 1,\\
F_{3(b)} &= 3 \times 3 - 2 \times 4 - 0 = 1,\\
F_{3(c)} &= 3 \times 3 - 2 \times 4 - 0 = 1,\\
F_{3(d)} &= 3 \times 3 - 2 \times 4 - 0 = 1,\\
F_{3(e)} &= 3 \times 1 - 2 \times 1 - 0 = 1.
\end{aligned} \tag{2}$$

2.3. Kinematic Simulation. The motion modes of ReChair include sit to stand, sit to lie, and lower limb rehabilitation training during sitting. In order to demonstrate that ReChair can realize multiposture transformation and passive lower extremity training in a stable way, the COSMOS motion module of SolidWorks is used to carry out kinematic simulation of the rehabilitation wheelchair. The position of monitoring points and sit to stand procedure is shown in Figure 3(a). When the wheelchair is changing from sitting to standing posture, the driving links are stand-up motor, lie motor, and shank back/forth motor. Before standing, let the front antitilt wheel touch the ground to ensure the stability of the mechanism and prevent its center of gravity from moving forward and causing the unexpected situation or discomfort given the ground impact. The horizontal angular displacement of seat, back, and leg support is shown in Figure 3(b). The whole simulation process is smooth, and there is no interference in the motion of the mechanism. When the displacement of the standing motor reaches the limit position, the wheelchair is exactly at the desired posture. The design requirements are achieved, and the motor selection and structural design are reasonable.

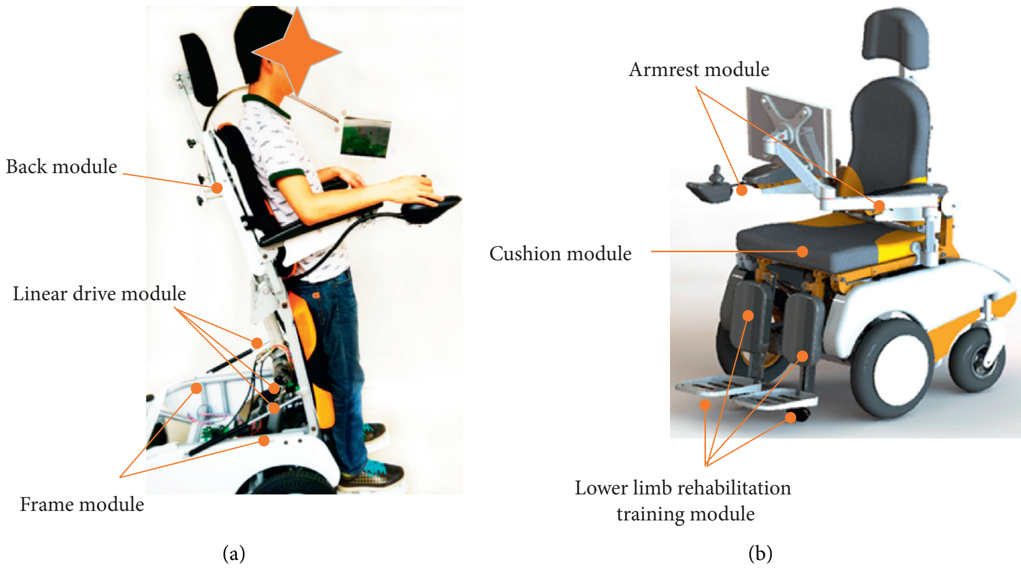


Figure 1: System design of ReChair.

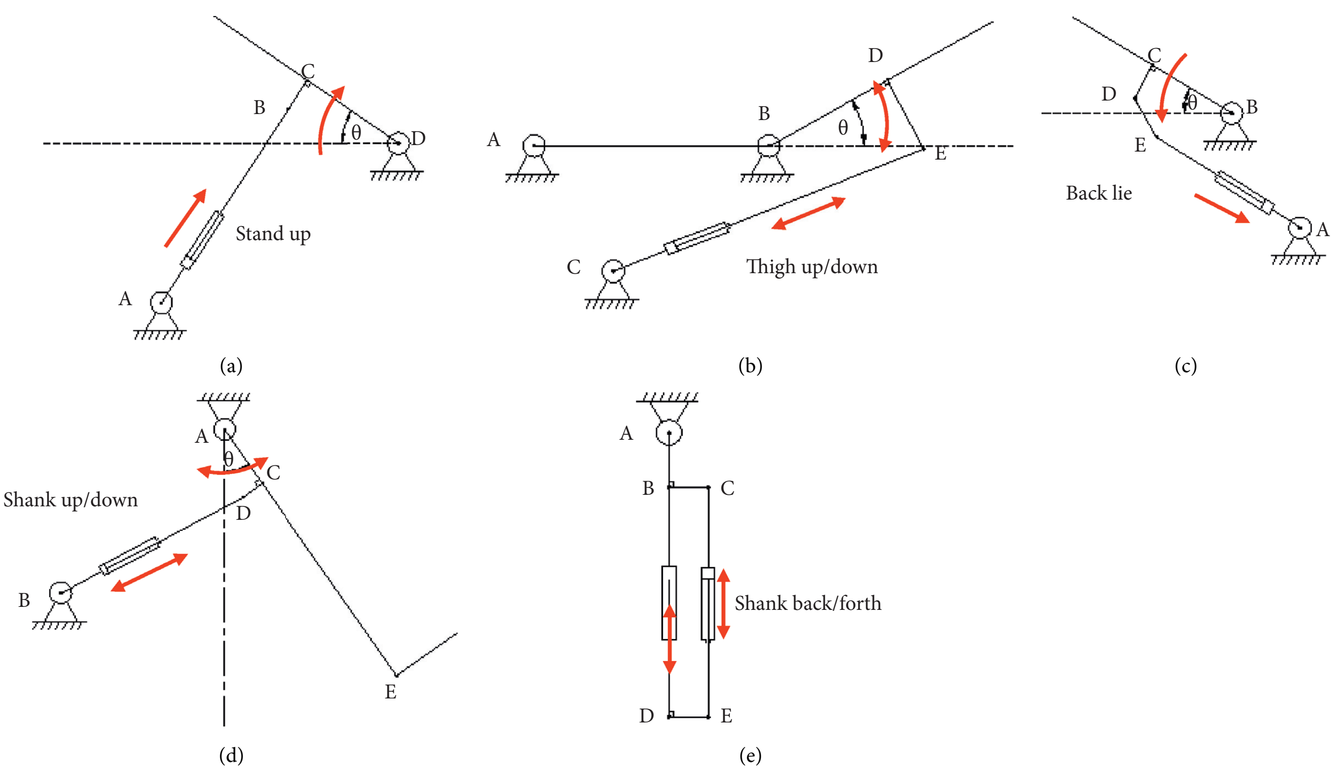
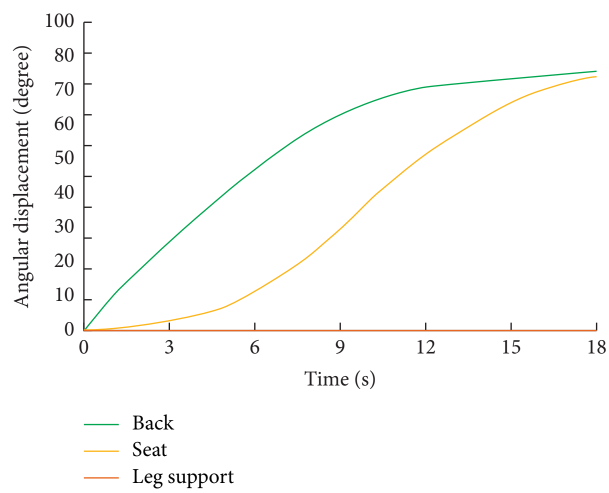


Figure 2: Motion mechanism: (a) stand up; (b) thigh up/down; (c) back lie; (d) shank up/down; (e) shank back/forth.

The sit to lie procedure and horizontal angular displacement of seat, back, and leg support are shown in Figure 4. The same simulation method is used with sit to stand posture transformation. The driving links are lie motor and shank up/down motor. The trajectories of the key components are smooth. When the displacement of the lie motor reaches the limit position, the back is exactly at the horizontal position and no hypokinesis or interference will occur. The sit to stand and sit to lie simulation results show that the multiposture transformation can be realized with ReChair.

The lower limb rehabilitation training procedure during sitting and the trajectory of the unique footplate are shown in Figure 5. The whole procedure is divided into two process: the up and forward movement of the footplate and the down and back movement of the footplate. The footplates are made

(a)

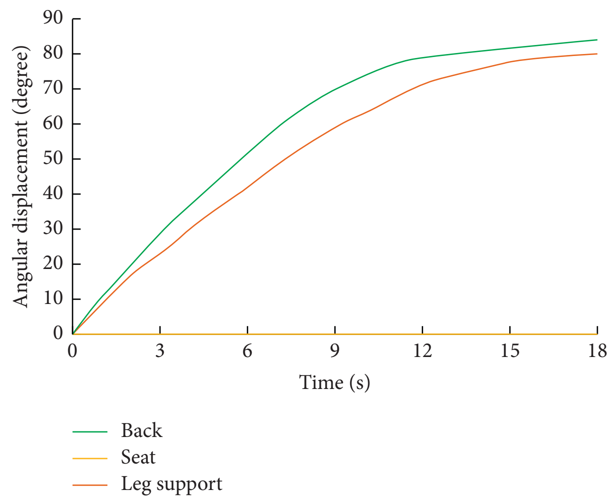


(b)

FIGURE 3: Sit to stand simulation. (a) Sit to stand procedure. A, B, C, D, E, and F are monitoring points. (b) The horizontal angular displacement of seat, back, and leg support.

(a)

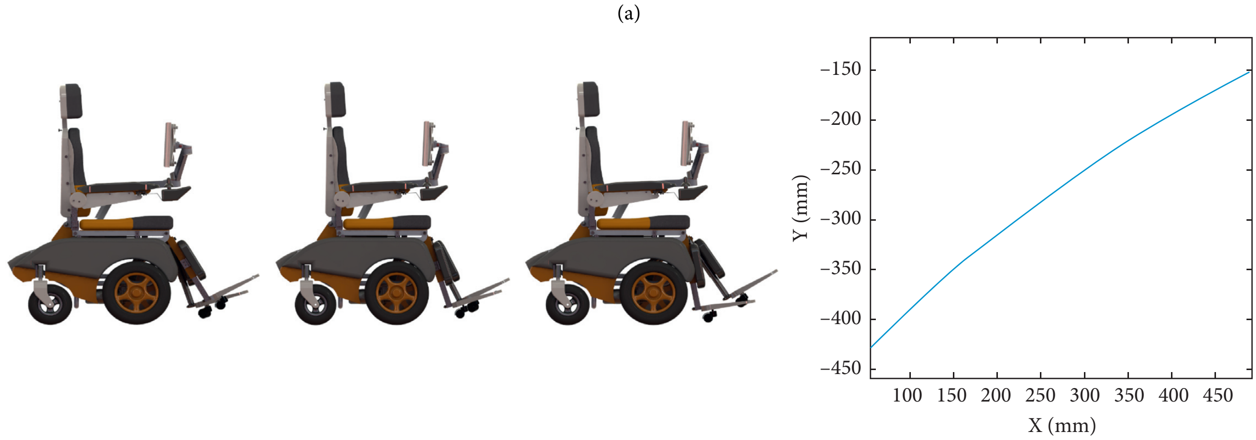
FIGURE 4: Continued.



(b)

Figure 4: Sit to lie simulation. (a) Sit to lie procedure. (b) The horizontal angular displacement of seat, back, and leg support.

(a)



(b)

Figure 5: The lower limb rehabilitation training procedure during sit and trajectory of the unique footplate. (a) Footplate up and forward procedure and trajectory. (b) Footplate down and back procedure and trajectory.

of aluminum alloy in this work. The laminated composite structures have shown good performance in dynamic conditions with random vibrations. Saman et al. proposed an accurate finite element model for N-layer MR-laminated beams using a layerwise theory [28] and conducted experimental works on dynamic behavior of laminated composite beam incorporated with magneto-rheological fluid under random excitation [29]. In view of the excellent characteristics of laminated composite structure, we will optimize the footplate by laminated composite materials in the following work. The footplates of both sides move in opposite directions during the same process. The passive training is similar to circular reciprocating motion. When the speed of the motion is changed, the motion trajectory of the footplate is different as well. Thus, the trajectory planning of the passive lower limb rehabilitation training is realized. The training time and intensity can be set by the nursing personnel or the patients themselves.

2.4. Control System Implementation. Commercially available controllers of the powered wheelchair include hand joystick, sip-n-puff, chin joystick, and head joystick [30]. The input methods including voice, touch, computer vision, accelerometer, and EEG become popular in the smart wheelchair recently [31]. Mousa et al. proposed a smart controller of a wheel chair mobile robot using particle swarm optimization proportional controller (PSO-P) to tune the proportional controller's gains for each axis [32], Rabhi et al. presented a new smart joystick for the electric wheelchair with position controller [33], and Ruzaij et al. described the design and implementation of three modes of operation for a voice controller in the wheelchair. The sound-dependent mode with the dynamic time warping algorithm, the sound-dependent mode with the hidden Markov model algorithm, and the sound-independent mode with the text to sound-independent algorithm were compared [34]. The control system of ReChair uses the idea of modularization and is shown in Figure 6. It is divided into four modules that are the main controller, travelling controller, posture controller, and signal processor.

The main controller is the core part of the ReChair control system. The instructions from the upper computer (mobile phone or smart tablet) and control panel (joystick or button) are received and processed by the main controller. Then, corresponding tasks and data are distributed to each subcontroller. The mode switch of each submodule is adjusted by the main controller to ensure normal function of the control system. Important data or information are fed back to the upper computer via Bluetooth and network communication system. The core chip control circuit is based on STM32F103RCT6 (STMicroelectronics, Switzerland). The STM32F103 incorporates the high-performance ARM® Cortex®-M3 32-bit RISC core operating at a 72 MHz frequency, high-speed embedded memories, and an extensive range of enhanced I/O and peripherals connected to two APB buses. It can satisfy the current and speed control of each motor, the communication between each module, and the acquisition and processing of various sensor signals. The power supplies of the main control module are 5 V and 3.3 V. The battery provides 24 V dc power supply, so voltage regulator chip is used to reduce the voltage to 5 V and 3.3 V. The 5 V DC power supply is obtained through DC-DC chip MP2303 (Monolithic Power Systems, American) and LDO chip LM1117-5.5 V (National Instruments, American). The 3.3 V DC power supply is generated by the 5V power supply through LM1117-3.3 V.

The wireless module in the main controller is responsible for the transmission of data communication and instruction. The chip SI4432 (Silicon Labs, American) is used to design the wireless module. Considering the practical application requirements, the 433 Mhz frequency band is chosen as the operating frequency of the wireless communication module. The module uses SPI communication interface, which is connected to the SPI2 of the main control chip on the main board.

The voice recognition module is embedded in the main controller to achieve convenient interaction between the user and the machine. The sitting, standing, and lying posture transformation can be implemented using voice control. Voice recognition can be divided into speaker-dependent and speaker-independent recognition [35]. Speaker-dependent recognition refers to the recognition of a specific person's voice and is used in the voice module of this system to ensure safety, while speaker-independent recognition refers to the recognition of anyone's voice. The core component used to design the voice recognition module is the NLP-5X (Sensory, American) voice recognition processor. To prevent loss of information during power outages, the EEPROM (electrically erasable programmable read only memory) is used to store speaker-independent templates. The EEPROM chip used for the speech template storage module is 24LC128 (MICROCHIP, America). To obtain the speech template, the voice recognition module collects and processes the speech signals through the microphone and sends them to the processor NLP-5X which then perform software coding of the recorded voice. The processed signal communicates with the MCU (Microcontroller Unit) module and informs it to make corresponding commands. The function of the loudspeaker is to make a sound cue during training and usage.

The joystick is a commonly used human-computer interface tool. Moving the joystick into different positions indicates different output voltage signals, which can easily be captured by electronic components. Thus, the input position signals of the operator can be collected. Small 9000SERIES joystick (APEM, France) with two axis is used in ReChair. It has two potentiometers that can output two mutually perpendicular voltage signals. These two channel voltage signals are connected to the main controller for 12-bit high-precision ADC conversion. The main controller sends the two analog voltage signals and the user-defined wheelchair speed level to the travelling controller through a CAN bus. First, the travelling controller normalizes the position coordinates of the joystick and the wheelchair, and then it calculates the PWM (pulse width modulation) duty cycle of the travelling control signals and the direction of the left and right motors. The rotation direction and speed of the two motors are used to realize the forward, backward, and steering of the wheelchair.

The multiposture transformation module is mainly composed of posture controller, posture driver, and linear actuator. The HT-KC35 (GEMING, China) linear actuator is used as power source for the multiposture transformation. This module is also equipped with three limit switches, two of them are installed at the positive and negative sides of the

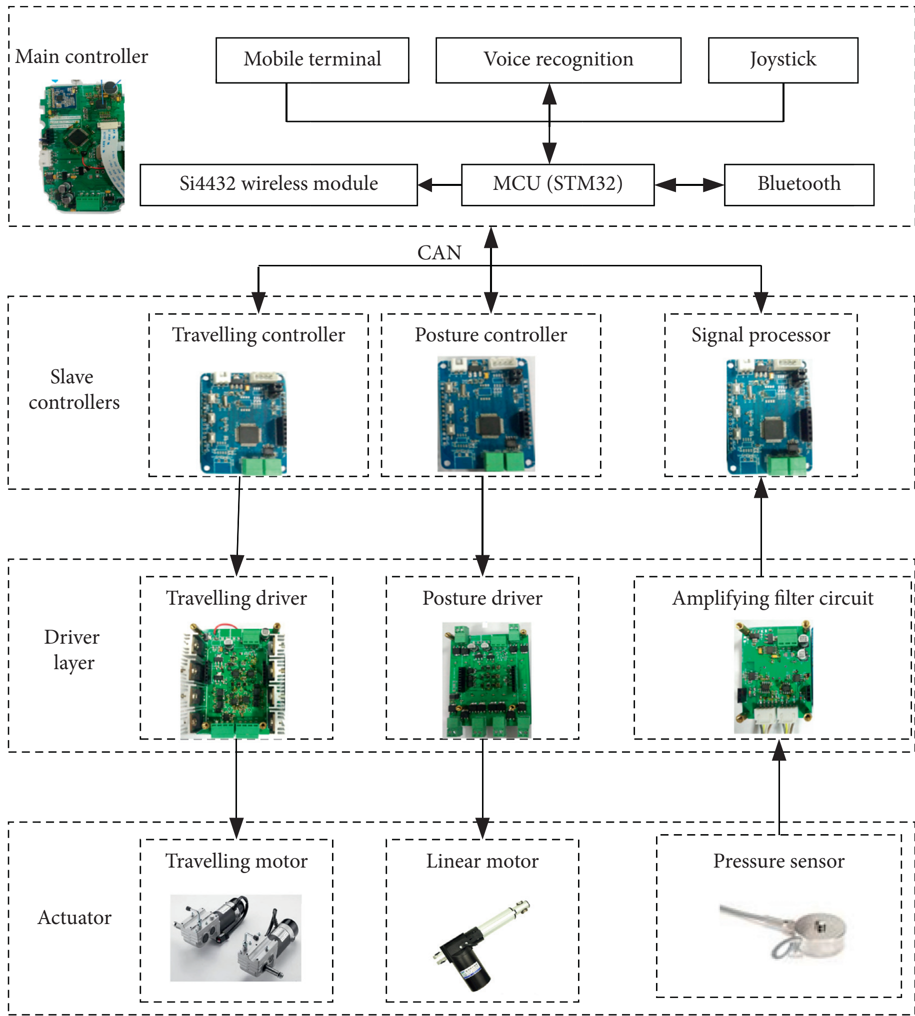


Figure 6: The control system of ReChair.

junction of the back and seat and the third one is installed at the junction of the seat and frame. Although the linear actuator automatically stops when it runs to stroke limit, the controller does not disconnect the control signal then. On the one hand, security risks are hard to avoid, but on the other hand, thermal loss will occur when the MOS tube is alternately switched on and off for a long time. The limit switch does not only ensure safety and save energy but it also informs the controller whether the current posture transformation process is completed.

The main function of the sensor module is the signal acquisition and the processing of the pressure sensors, which is mainly composed of the signal processor, an amplifying filter circuit, and pressure sensors. Two JHBM-100 kg (HBM, Germany) pressure sensors are installed on the left and right footplates of the intelligent rehabilitation wheelchair. In the standing mode, patients can change their center of gravity to control the left and right movements of the game object in order to complete the training game. The pressure sensors detect the change of the center of gravity and convert pressure signals into electrical ones. The deviation side pressure of the center of gravity will increase, resulting in an increase in the output voltage. Lower pressure leads to smaller output voltage of the pressure sensor. The output voltages of the pressure sensors on both sides are amplified by a differential amplification circuit. Then, the signals are used as input to the signal processor after ADC. In the game, the bird will move to the side where the output voltage increases, namely, to the shift side of center of gravity. The greater the voltage difference is, the faster the bird moves.

3. Results and Discussion

3.1. Voice Control System Evaluation. ReChair provides a voice interactive interface so that the posture transformation and lower limb rehabilitation training can be controlled by voice. Seven voice commands (1: sit up, 2: lie down, 3: stand up, 4: sit down, 5: passive training, 6: balance training, and 7: stop) are used as control signals. The user says the control command and the wheelchair controller execute the corresponding action as a success, while the others are recorded as failure. The voice control system is tested by one male and one female users in different environments. First, voice command training is conducted in quiet environment (20–30 DB). Each command training is conducted twice and stored. Next, two subjects repeat the seven commands in Mandarin, English, and Cantonese 20 times each in quiet environment (20–30 DB) as well. Finally, the above-mentioned process is repeated in a relatively quiet environment (30–40 DB) and a noisy environment (50–60 DB). The success recognition times and rates as well as the mean values are as shown in Table 1. The results indicate that the average success recognition rate can reach 91% to 96% in a quiet environment (20–30 DB), no matter which language is used or what the sex of the subject is. The average success recognition rate is 81% to 84% in a relatively quiet environment (30–40 DB). The range of 20–40 DB is close to home environment. The success recognition rate shows that the voice control system is suitable for control in home environment. While the recognition rate decreases to 49% to 59% during noisy environment (50–60 DB), the results indicate that the voice control system is not suitable for noisy environment. There is little difference in the success rate of different languages during the same environment. Language difference has little effect on the voice control system.

3.2. Sensorless Speed Detection Method Test of Travelling Motor. The speed closed-loop control method is generally used to improve the static and dynamic performance of the DC motor speed control system. The premise of the speed closed loop is the speed detection. Commonly used speed detection methods rely on centrifugal, optical digital, and flash tachometers. Although the accuracy of the sensor method is high, they need to be in a close friction with the rotating part of the motor in the travelling process or fixed in a special position. The installation of the speed detection device increases the structural complexity and the mechanical assembly difficulty of the wheelchair. This work proposes a sensorless speed detection method. When the brush DC motor is working, the DC provided by the H-bridge drive loop is introduced into the armature commutator, and the pulsating current is generated in the drive loop. The pulsation frequency is related to the speed of the DC motor. As long as the frequency information of the pulsating current is collected, the actual speed can be estimated. The pulsating current is collected by the current transformer in series to the circuit to output the pulsating waveform of voltage. The speed of the brush motor is also related to its own related parameters, such as the number of commutator segment, the pairs of magnetic poles, and the parity of the number of the commutator segment [36]. The relation is expressed as

$$f_n = \frac{ckpn}{60}, \tag{3}$$

where f_n indicates the current pulsation frequency, n is the speed of the motor, k is the number of the commutator segment, p indicates the pairs of magnetic poles, and c is the coefficient related to the parity of k such that $c = 1$ when k is an even number and $c = 2$ when k is an odd number. The number of the commutator segment, the pairs of magnetic poles, and the coefficient are all constant values for a specific brush DC motor. Therefore, the speed of the brush DC motor n is proportional to the current pulsation frequency.

$$n = \frac{60 f_n}{ckp}. \tag{4}$$

Considering the reduction ratio i, the wheel speed detected by the sensorless method n_w is

$$n_w = \frac{n}{i} = \frac{60 f_n}{ickp}. \tag{5}$$

To verify the sensorless speed detection method, two travelling motors of the electric wheelchair are selected as experimental objects. The speed measured by the sensorless speed detection circuit is transmitted to the PC using the Bluetooth module of the travelling controller (Figure 7).

For the travelling motor, the values are $p = 8$, $k = 59$, $c = 2$, and $i = 32$. The laser tachometer commonly used in the market is used to measure the actual wheel speed. PWM is used to control the motor speed, and the driving signal frequency is set to 20 KHz. Thirteen sets of duty ratio are given to travelling motors, and the frequency of commutating current is measured. The wheel speeds measured by the sensorless speed detection method and laser tachometer are shown in Table 2. When the PWM is below 15%, such as 10% and 5%, the current is so small that the motor torque cannot overcome the internal friction torque of the motor. The wheelchair stays still, and the speed is zero. When the PWM is enough to drive the wheelchair, the results indicate that the wheel speeds measured by the sensorless method are close to the laser tachometer displayed values. The procreant reasons of plausible error are the interference of the electrical spark generated by the motor on the measured waveform and the influence of distance with the laser tachometer on measuring accuracy. The higher the running speed of brush DC motor, the greater the measurement error. This work investigated the speed detection of the wheelchair-driving motor, whose maximum speed is limited and not suitable for high-speed operations, which meet the requirement.

3.3. Lower Limb Rehabilitation Training Test. The study was approved by our institutional review board, and all subjects have signed informed consent. ReChair includes two training modes: passive training with constraint trajectory and balance training combined with games. Two persons including a healthy person and a patient are involved in the lower limb rehabilitation training test (Figure 8).

Before passive training, the rehabilitation doctor logs into the personal center on the tablet application and makes a rehabilitation training plan. The plan includes the following parameters: training time, maximum range of motion, and training speed. The training plan is synchronized with the wheelchair, and users can then carry out lower limb rehabilitation training of pedaling gait on the wheelchair. The experimental procedure runs as follows:

Table 1: Experimental results of the voice control system.

Subject	Env. (DB)	Commands language		1	2	3	4	5	6	7	Mean
Male	20–30	Man.	Suc. times	19	18	20	19	20	18	19	19
			Suc. rate	95%	90%	100%	95%	100%	90%	95%	95%
		Eng.	Suc. times	18	20	19	20	18	19	19	19
			Suc. rate	90%	100%	95%	100%	90%	95%	95%	95%
		Can.	Suc. times	20	19	19	20	18	19	19	19
			Suc. rate	100%	95%	95%	100%	90%	95%	95%	95%
	30–40	Man.	Suc. times	18	16	15	17	16	17	17	17
			Suc. rate	90%	80%	75%	85%	80%	85%	85%	83%
		Eng.	Suc. times	17	17	16	16	15	17	18	17
			Suc. rate	85%	85%	80%	80%	75%	85%	90%	83%
		Can.	Suc. times	16	16	18	15	16	16	17	16
			Suc. rate	80%	80%	90%	75%	80%	80%	85%	81%
	50–60	Man.	Suc. times	10	12	10	10	11	10	10	10
			Suc. rate	50%	60%	50%	50%	55%	50%	50%	52%
		Eng.	Suc. times	9	11	10	9	10	10	10	10
			Suc. rate	45%	55%	50%	45%	50%	50%	50%	49%
		Can.	Suc. times	11	11	10	9	10	10	11	10
			Suc. rate	55%	55%	50%	45%	50%	50%	55%	51%
Female	20–30	Man.	Suc. times	20	20	19	19	19	19	19	19
			Suc. rate	100%	100%	95%	95%	95%	95%	95%	96%
		Eng.	Suc. times	18	18	18	19	19	18	18	18
			Suc. rate	90%	90%	90%	95%	95%	90%	90%	91%
		Can.	Suc. times	18	17	19	19	19	18	18	18
			Suc. rate	90%	85%	95%	95%	95%	90%	90%	91%
	30–40	Man.	Suc. times	16	16	18	17	17	17	17	17
			Suc. rate	80%	80%	90%	85%	85%	85%	85%	84%
		Eng.	Suc. tmes	16	16	16	17	17	16	16	16
			Suc. rate	80%	80%	80%	85%	85%	80%	80%	81%
		Can.	Suc. times	17	16	16	17	17	16	16	16
			Suc. rate	85%	80%	80%	85%	85%	80%	80%	82%
	50–60	Man.	Suc. times	12	13	11	11	11	12	12	12
			Suc. rate	60%	65%	55%	55%	55%	60%	60%	59%
		Eng.	Suc. times	13	12	12	12	9	10	10	11
			Suc. rate	65%	60%	60%	60%	45%	50%	50%	56%
		Can.	Suc. times	11	10	10	12	11	11	11	11
			Suc. .rate	55%	50%	50%	60%	55%	55%	55%	54%

Man.: Mandarin; Eng.: English; Can.: Cantonese; Suc.: success; 1: sit up; 2: lie down; 3: stand up; 4: sit down; 5: passive training; 6: balance training; 7: stop.

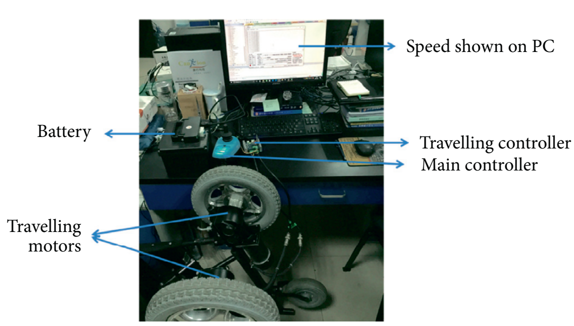


Figure 7: Sensorless speed detection method test of the travelling motor.

(1) Adjust the wheelchair to the sitting position and set the level of the motion range. There are six levels of motion range: the first test is set to level 1 which is the least range of motion, while level 6 is the largest one but is still within the normal motion range. The training time is set to 5 minutes, and the training speed is fixed at moderate level.

(2) Start the training and measure the maximum angles of θ_1 and θ_2.

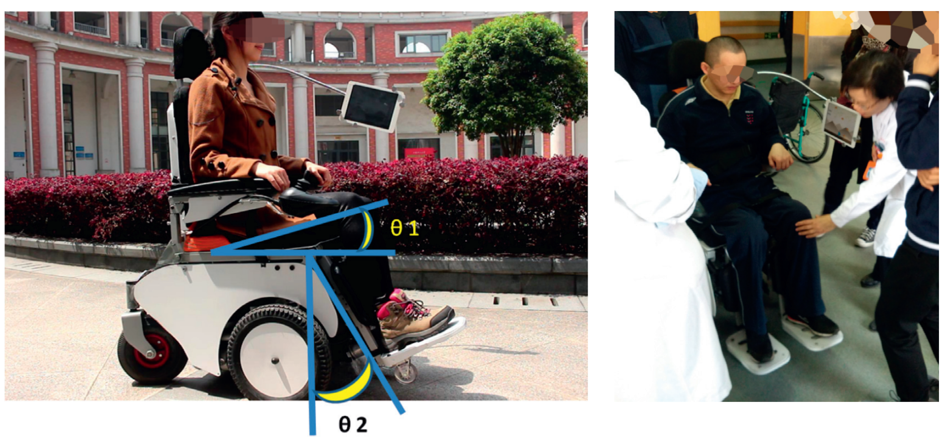
(3) Take a 10-minute break. Increase the range of motion level by one and repeat steps (1) and (2).

(4) Stop the experiment when the subject feels slightly sweatful.

The results are shown in Table 3. The safety must be placed first during rehabilitation training. If the angles range of θ_1 and θ_2 were too low or too large, the training may cause secondary injury to the patients. For θ_1, the maximum thigh up angle was set to be 20°. For θ_2, the maximum shank up angle was set to be 58°. The maximum angle had room for improvement in mechanical design. However, it was a safe threshold for rehabilitation training. The healthy person can complete passive training with all motion levels while the patient cannot complete level 6. Although the motion

TABLE 2: The wheel speeds measured by the sensorless speed detection method and laser tachometer.

PWM (%)	f_n (Hz)		Estimation speed (rad/min)		Laser tachometer display (rad/min)		Error (%)	
Wheel	Left	Right	Left	Right	Left	Right	Left	Right
5	0	0	0	0	0	0	\	
10	0	0	0	0	0	0		\
15	61	58	7.27	6.91	7.5	7.1	3.06	2.67
20	82	84	9.77	10.01	9.5	9.7	2.84	2.77
30	128	126	15.25	15.02	14.7	15.5	3.74	3.09
40	180	183	21.45	21.80	22.3	21.1	3.83	3.32
50	228	224	27.17	26.69	26.2	27.7	3.71	3.65
60	270	273	32.18	32.53	33.5	33.8	3.94	3.76
70	317	322	38.25	38.37	39.9	36.8	4.13	4.27
80	378	374	45.05	44.57	47.1	46.7	4.35	4.56
90	417	424	49.69	50.53	52.2	48.1	4.81	4.79
95	447	453	53.27	53.99	56.3	56.9	5.38	5.64
98	488	497	58.16	59.23	62.3	63.5	6.65	6.72



(a) (b)

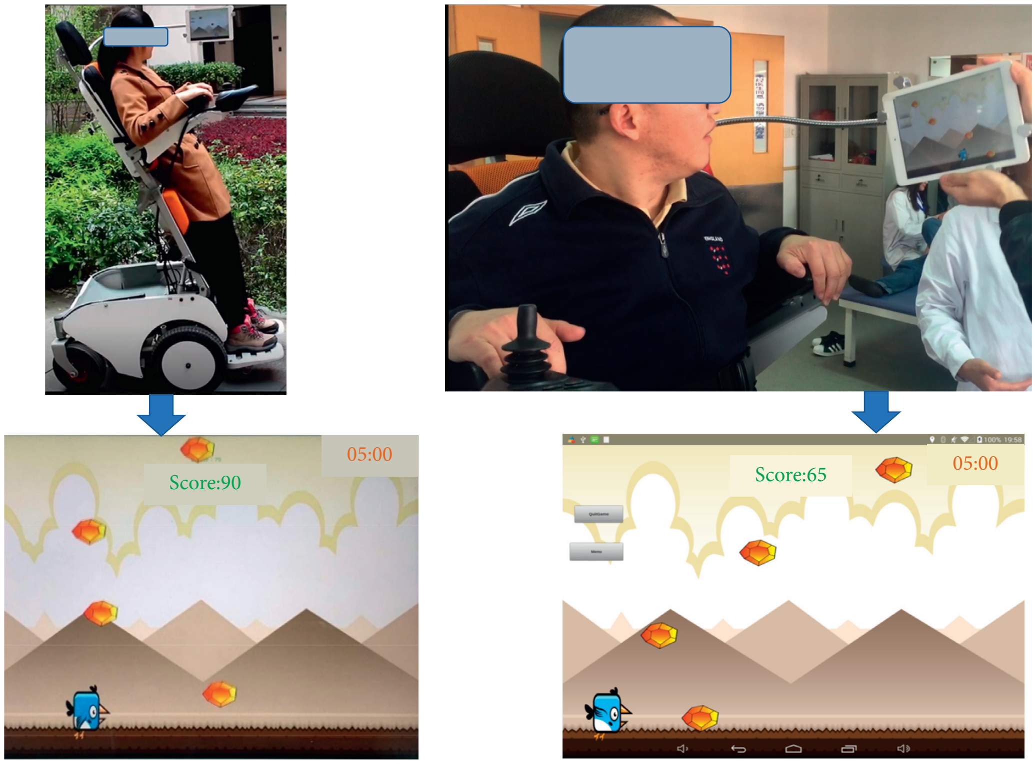
FIGURE 8: Passive lower limb rehabilitation training. (a) Normal person where θ_1 is the range of hip joint and θ_2 is the range of knee. (b) Patient.

TABLE 3: The passive training results of normal person and patient.

Subject	Motion level	Complete	Training time (minute)	θ_1(degree)	θ_2(degree)
Normal person	1	Yes	5	10	30
	2	Yes		12	35
	3	Yes		14	41
	4	Yes		16	45
	5	Yes		18	53
	6	Yes		20	58
Patient	1	Yes	5	11	32
	2	Yes		13	36
	3	Yes		15	43
	4	Yes		18	47
	5	Yes		20	55
	6	No		\	\

amplitude of level 6 is too large for the patient, experimental results show that the wheelchair can help users to do passive training, and the range of motion can be adjusted appropriately during the training process.

Balance training is conducted in standing posture while playing the game. The user changes the center of gravity to control the left and right movement of the bird. In the game, the bird will move to the side where the output voltage increases, namely, to the shift side of the center of gravity. The greater the voltage difference is, the faster the bird will move. The game interface is shown in Figure 9. The healthy person gets the high score indicating that the balance



(a) (b)

Figure 9: Balance training combined with game. (a) The score obtained by the normal person. (b) The score obtained by the patient.

training game is easy to operate, while the patient gets a lower score during the training time of five minutes. The result indicates that the balance control ability can still be significantly improved. The patient gets higher score with more training, and this point can be as a quantified guideline for rehabilitation. The game and pressure signal processing module are beneficial and could be an important part of rehabilitation training.

4. Discussion

In this work, we proposed a multifunctional wheelchair (ReChair) that seamlessly integrates the mobility, gait training, and multiposture transformation for daily life conditions. The voice, button, and mobile terminal (mobile phone or smart tablet) control are used for driving and multiposture transformation of ReChair. To evaluate the system, we conducted kinematic simulation and pilot test of ReChair. Results showed that the multiposture transformation by voice control, sensorless speed detection method of the travelling motor, and passive and active training are realized as expected.

Previous research presented lower limb rehabilitation training as mostly conducted by BWSTDs in hospitals. The process of moving a patient from the wheelchair to BWSTD is cumbersome and takes about twenty minutes. This step is skipped by ReChair as the lower limb rehabilitation training is conducted directly on the wheelchair. ReChair reduces the work intensity of the nursing staff and is time saving. ReChair can also be used at home. The low cost of ReChair compared with BWSTDs is more affordable to the patients. Additionally, ReChair reduces the patient's tension on gait rehabilitation machine. Compared with LLE for rewalk [3, 7–13], ReChair focuses on restand and remobility. The multiposture transformation avoids the complications caused by lying down for a long time. The mobility of ReChair meets the travel demand of the patient. It is easier and safer than LLE in daily use. Compared with the wheelchair integrating exercise/rehabilitation system in [25], ReChair includes alternative control modes that make it more widely applicable.

Some extensions and improvement such as smart home module can be integrated into ReChair. Electric control door, lights, and air conditioner in home environment can be combined with ReChair to form a local area network (LAN) system of things. It provides interactive interfaces such as voice recognition, touch terminal, and control panel for people with mobility disabilities so that they can control various electrical equipment at home wirelessly by ReChair. Future goals are to expand the effect of long time use of ReChair on noncomplete SCI and other neurological causes of gait dysfunction. The limitations of this study include the relatively small sample size, the limited range of lower limb motion, and the lack of postintervention follow-up.

5. Conclusions

Restand, mobility, and lower limb rehabilitation training are core requirements of patients with lower limb motor dysfunction. In this study, we proposed a multifunctional wheelchair ReChair integrating the mobility, gait training, and multiposture transformation in daily life conditions. The whole design concept and detailed specifications of each module were described. The kinematic simulation of multiposture transformation and passive lower limb rehabilitation training were conducted to quantitatively verify the motion capability of ReChair. The voice, button, and mobile terminal (mobile phone or smart tablet) control were used for driving and multiposture transformation of ReChair. The success recognition rate of voice control evaluation shows that the voice control system is suitable for home environment. The sensorless speed detection test results indicate that the wheel speeds measured by the sensorless method are close to the laser tachometer display. The passive and balance training results show that the lower limb rehabilitation training in daily life by ReChair is convenient. In future work, we plan to add a smart home module to ReChair and recruit paraplegic patients for long time rehabilitation training to verify the effectiveness of ReChair.

Data Availability

The data used to support the findings of this study are available from the corresponding author upon request.

Conflicts of Interest

The authors declare that they have no conflicts of interest regarding the publication of this paper.

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