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Power Steering and Force Display Controls for a Cycling Wheelchair using Servo Brakes

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***Abstract*— In this study, we present a new type of wheelchair called the cycling wheelchair, in which the user moves the wheelchair by pedaling with his/her legs and changes direction using a steering handle. This wheelchair has great potential not only as an effective rehabilitation device but also as a mobility assistive device. However, the commercially available model has several hardware limitations that affect the steering operation. In this paper, we discuss these hardware limitations before presenting methods for improving steering maneuverability. Furthermore, we propose power steering and force display controls based on environmental information. From the perspective of safety, we employ servo brakes to control the cycling wheelchair, and the assistive functions are realized based on the feasible braking control region. The proposed control methods are experimentally applied to a newly developed cycling wheelchair with servo brakes, and the results illustrate the validity of the control methods.**

I. Introduction

Patients with impairment of their lower extremities find it difficult to walk and may therefore use mobility assistive devices such as wheelchairs. However, most standard wheelchairs are operated by the arms via a hand rim. Thus, muscular strength in the legs declines because of disuse, and patients may also experience discomfort from exerting their arms.

A new wheelchair has been developed for rehabilitating patients with severe impairment of the lower extremities [1]. This wheelchair is known as the "Cycling Wheelchair" as shown in Fig. 1, and it is based on a pedal-driven system similar to a bicycle. In [1], it was reported that by functional electrical stimulation (FES), patients could use their disabled legs to rotate the pedals of the cycling wheelchair. Following extensive clinical tests, a medical doctor observed that hemiplegic patients can move the cycling wheelchair without FES [2]; the healthy leg could apply high torque to one pedal of the system while the disabled leg did not interfere with the rotational pedaling motion. Furthermore, we asked many patients with a range of impairments of the lower extremities, including paraplegia, to use the cycling wheelchair. From these clinical tests, it was observed that many patients can pedal the cycling wheelchair if they have slight mobility in their legs.

The cycling wheelchair is now commercially available from TESS Co. Ltd., Japan [3] and has encouraged user participation in outdoor activities. Several patients with cycling wheelchairs enjoy shopping in supermarkets, travelling, and performing other such activities. By promoting enjoyment in performing outdoor activities, the cycling wheelchair provides these patients with an effective means of rehabilitation.

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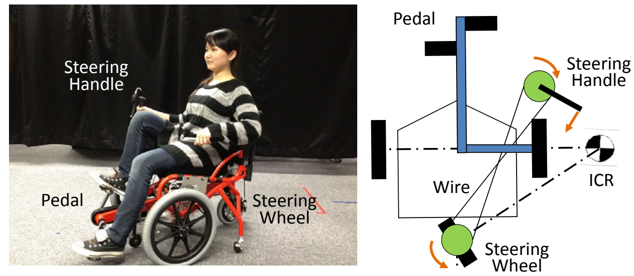

Fig. 1. Commercially Available Cycling Wheelchair [3]

However, patients with severe impairment of the lower extremities sometimes cannot maneuver the cycling wheelchair in outdoor environments because of barriers such as steep slopes, obstacles, and steps. Particular hazards include reaching excessive downhill speeds and falling down the steps. To minimize such risks, we have developed a method of braking control that employs servo brakes [4], and realized several assistive functions such as gravity compensation control and obstacle avoidance control.

A further limitation of the commercially available design of the cycling wheelchair involves the steering handle. Because of hardware limitations affecting the steering operation, users perceive right and left turns differently and require a high operating torque to change the direction of the cycling wheelchair. In addition, unlike a traditional hand rim wheelchair, the cycling wheelchair cannot easily be propelled from behind by a caregiver.

To overcome these limitations we propose a power steering control that can reduce the force required to steer the cycling wheelchair. Additionally, we propose a force display control that can guide the user in applying force to the steering handle by providing environment-specific information. Generally, an active actuator such as a servo motor attached to the steering handle realizes the power steering control. However, in this study, we focus on the wheel braking control of the cycling wheelchair. In our previous research [4], we implemented a braking-based assistive control. The braking system is an indispensable safety feature for real world wheelchair use. Thus, our design utilizes a brake system attached to the wheels to implement the power steering and

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force display controls without using servo motors. In general, the braking system cannot amplify human applied force. The proposed power steering control is realized by appropriately combining the braking force with a proportion of the human pedaling force. That is, the pedaling force, which is much greater than the force generated by the arms, compensates for the insufficient force of the arms for the steering operation.

In this paper, we first consider the hardware-based steering limitations of the commercially available cycling wheelchair, and then describe a new design of the cycling wheelchair with servo brakes. Next, we propose a method for power steering control using the servo brakes, and extend this method to realize a force display control that provides user guidance based on environmental information. Finally, we apply the proposed method to the cycling wheelchair and describe our experimental results, which illustrate the validity of our design.

II. Cycling Wheelchair with Servo Brakes

A. Limitations of the Commercially Available Cycling Wheelchair

The commercially available cycling wheelchair has a steering wheel as shown in Fig. 1. The user can change the angle of the steering wheel using the steering handle, which is connected to the steering wheel via a wire. The pedaling torque applied by the user is transmitted only to the right wheel; the left wheel is passive. The right, left, and steering wheels are in contact with the ground to give the cycling wheelchair 1 degree of freedom (DOF) around an instantaneous center of rotation. As the position of the instantaneous center of rotation is determined by the angle of the steering wheel, its angle changes the direction of the wheelchair. However, as the pedaling torque is transmitted to the right wheel alone, the user perceives right and left turns differently, with right turns requiring more pedaling torque.

Furthermore, the user has to apply substantial torque to the steering handle to change the angle of the steering wheel. This necessitates a large offset between the steering lever and the rotational point of the handle. However, if we increase the offset of the steering lever, the user cannot use the wheelchair in narrow spaces where obstacles such as walls restrict steering operation. In addition, if the user has an impairment of the upper extremities, he/she may not be able to move his/her arms sufficiently to operate the steering handle. As the result, the offset of the steering lever has to be small and the user must apply high steering torque.

On occasions when the user cannot apply sufficient driving force to the wheelchair, a caregiver may assist the user by applying force from the rear. However, as the cycling wheelchair has 1 DOF of motion determined by the angle of the steering wheel, the caregiver cannot change the direction of the cycling wheelchair directly by applying moment to it. That is, only the user can change the direction of the cycling wheelchair, and the caregiver has to lift up the steering wheel while turning the wheelchair if he/she wants to move it without the operation of the user.

Fig. 2. Modified Cycling Wheelchair with Servo Brakes

B. New Cycling Wheelchair

To overcome the above limitations, we have developed a new cycling wheelchair to which a differential gear, a steering caster, and servo brakes are attached. The differential gear is attached to the axis of the right and left wheels, and the pedaling torque is transmitted to each wheel equally. The differential gear improves maneuverability and enables the user to perceive right and left turns equally. Additionally, we have changed the steering wheel to a caster with offset. The standard wheel does not have an offset between the wheel axis and its pivot shaft, so that the motion around the pivot shaft is nonholonomic. Conversely, the caster has an offset and the motion around the pivot shaft is holonomic. Therefore, the cycling wheelchair can achieve translational motion in the forward direction and rotational motion around the center of the wheel axis without being disturbed by the motion of the steering caster. This motion characteristic is the same as that of a standard wheelchair, so that the caregiver can propel the cycling wheelchair from behind.

To implement power steering control and display the force to the steering handle based on environmental information, we have attached servo brakes to the right and left wheels. By applying braking torque to each wheel independently, we can apply moment to the cycling wheelchair to change its direction. This moment contributes to the assistive motion of the steering handle to realize the power steering and force display controls.

Power steering control is generally implemented using active actuators such as servo motors. When focusing solely on the directional control of the system, Cobot architecture, as proposed in [5], is a good solution. It enables servo motor control of the steering wheel angle. Conversely, the methods of motion control presented in our previous research employed servo brakes to implement multiple functions, such as velocity and obstacle avoidance controls, for a walking support system [6], an object handling system [7], and a cycling wheelchair [4]. The proposed braking control methods realized not only the velocity control but also the rotational control of the system with mobility. In addition,

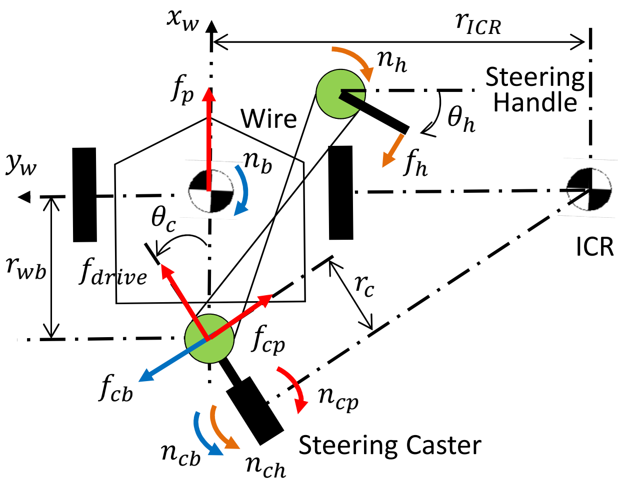


Fig. 3. Steering Control Using Braking Moment

these methods can provide greater user safety than an active actuator system while using little electricity.

We also previously proposed a direction control method for the cycling wheelchair [8]. In that study, we removed the steering wheel and controlled the motion of the right and left wheels to change the direction of the wheelchair, thus implementing hands-free operation. However, if the battery of this system fails, the wheel actuator cannot provide directional control. This drawback prevents the user to use the cycling wheelchair in an outdoor environment. Therefore, in the present study, we do not remove the steering wheel but simply change it from a standard wheel to a caster, so that the user can operate the cycling wheelchair manually in the case of battery failure.

III. Steering Assist Control

A. Power Steering Control

In this section, we consider a motion control method for assisting the steering handle using the servo brakes attached to the right and left wheels. Fig. 3 shows a schematic illustration of the cycling wheelchair as viewed from the top. A wheelchair coordinate system is defined as ${}^{w}\Sigma$, as shown in Fig. 3, where the x-axis is the forward direction of the cycling wheelchair. The cycling wheelchair is moved by human pedaling force f_p. The pedaling force is distributed equally to each wheel via the differential gear. This indicates that the user cannot apply moment to the wheelchair via the pedal.

When human pedaling force f_p is applied to the pivot shaft of the caster, we can divide the force f_p into two forces: f_{drive} and f_{cp}. f_{drive} drives the wheelchair along the heading direction of the caster. f_{cp} is the force applied perpendicular to the heading direction of the caster, and it contributes to the change in the angle of the caster. f_{cp} is derived as follows:

$$f_{cp} = f_p \sin\theta_c \ , \tag{1}$$

where θ_c is the caster angle. From f_{cp}, the moment applied around the axis of the caster wheel n_{cp} is expressed by the following equation:

$$n_{cp} = f_{cp} r_c \ , \tag{2}$$

where r_c is the offset of the caster as show in Fig. 3. n_{cp} is the restoring moment for synchronizing the angle of the steering wheel with the forward direction of the wheelchair. This moment is familiar as the caster angle of the front wheels of bicycles and automobiles.

When we can control the braking forces of right and left wheels independently using the servo brakes, we can apply moment n_b around the center of the wheel axis as shown in Fig. 3. From n_b, we can derive the force f_{cb} applied to the pivot shaft of the caster as follows:

$$f_{cb} = \frac{n_b \cos\theta_c}{r_{wb}} \ , \tag{3}$$

where r_{wb} is the distance between the center of the wheel axis and the pivot shaft of the caster as shown in Fig. 3. From f_{cb}, we can also derive the moment n_{cb} applied around the axis of the caster wheel as follows:

$$n_{cb} = f_{cb} r_c = \frac{n_b \cos\theta_c r_c}{r_{wb}} \ . \tag{4}$$

On applying the force f_h to the steering lever, moment n_h is applied to the rotational axis of the steering handle as shown in Fig. 3. As the steering handle is connected to the steering caster via a wire, the moment around the caster wheel caused by n_h is expressed as n_{ch}. If the direction of this moment is the same as that of moment n_{cb} generated by the servo brakes, the user can rotate the steering handle with less force than the standard steering force f_h. Consequently, power steering control can be implemented using the brake system.

Under the assumption that the steering handle is connected to the steering caster via a wire with an elastic coefficient k, the motion equations for the steering caster and the steering handle are expressed as follows:

$$m_c\ddot{\theta}_c + d_c\dot{\theta}_c - k(\theta_h - \theta_c) = n_{cb} - n_{cp} \tag{5}$$
$$m_h\ddot{\theta}_h - k(\theta_c - \theta_h) = n_h \ , \tag{6}$$

where m_c and m_h denote the inertia of the steering caster and the steering handle, respectively, and d_c denotes the damping coefficient of the steering caster. In this study, we ignore the damping coefficient of the steering handle because the steering handle is designed to rotate smoothly. θ_c and θ_h denote the angles of the steering caster and the steering handle, respectively. To achieve power assist control of the steering handle, we design an apparent motion dynamics for the steering caster as follows:

$$m_c\ddot{\theta}_c + d_c\dot{\theta}_c = \alpha n_h - n_{cp} \ , \tag{7}$$

where α is the power assist ratio and $a > 1$.

When we assume that the user operates the steering handle without acceleration or deceleration, we have $\ddot{\theta}_h = 0$. From eq.(5), eq.(6), and eq.(7), the moment n_{cb} around the axis of the caster wheel is expressed by the following equation:

$$n_{cb} = k(\alpha - 1)(\theta_h - \theta_c) \ . \tag{8}$$

In this study, we measure θ_h using a potentiometer attached to the steering handle, and θ_c is calculated by encoders attached to the right and left wheels. From the geometry of the cycling wheelchair, θ_c is derived as follows:

$$\theta_c = \tan^{-1}\frac{r_{wb}}{r_{ICR}} + \sin^{-1}\frac{r_c}{\sqrt{r_{ICR}^2 + r_{wb}^2}} \ , \quad (9)$$

where r_{ICR} denotes the distance between the center of the wheel axis and the instantaneous center of rotation (ICR) as shown in Fig. 3. The position of the ICR is determined by the angle of the steering caster.

From the derived moment n_{cb}, we can calculate the moment n_b around the center of the wheel axis as follows:

$$n_b = \frac{n_{cb} r_{wb}}{r_c \cos\theta_c} \ . \quad (10)$$

By generating the moment n_b from the servo brakes attached to the wheels, we can achieve power assist control of the steering handle.

In this study, we have assumed that the wire has elasticity; as a result, the power steering control is achieved without using a force/torque sensor. If the wire does not have elasticity, we would have to design a steering system with a compliance that mimics the elasticity of the wire. That is, when we measure the relative angles between the steering handle and the steering wheel, power steering control is achieved using the proposed method. Thus, we can develop an intuitive steering interface, similar to a joystick, to suit the user's level of disability by considering the compliance of the interface. Certainly, if a force sensor is attached to the steering handle, the power steering control is realized without using the elasticity of the wire or the compliance of the new interface.

B. Force Display Control

When the wheelchair is used in a complex environment containing obstacles, the system should provide the user with environmental information. While this could be presented via an audiovisual interface, our system provides environmental information by applying an intentional force to the steering handle.

In addition, patients with severe impairments of the upper extremities may not be able to apply appropriate force to the steering handle. According to the observations of a medical doctor, some patients moved the steering handle in an oscillatory manner and could not keep the wheelchair moving in a straight line. In such cases, it would be beneficial to display the force to the steering handle to avoid steering the wheelchair in an unintended direction. Using this force display control, we can implement path following functionality.

To implement force display control, we determine the direction of the cycling wheelchair from environmental information, and then generate a force for rotating the steering caster in the intended direction. In our system we redesign the apparent dynamics of the steering caster expressed in eq.(7) as follows:

$$m_c\ddot{\theta}_c + d_c\dot{\theta}_c + k_c(\theta_{cd} - \theta_c) = \alpha n_h - n_{cp} \ , \quad (11)$$

where k_c denotes the stiffness and θ_{cd} denotes the desired angle of the steering caster determined by the environment. From the above equation, we can calculate the braking moment as follows:

$$n_{cb} = k(\alpha - 1)(\theta_h - \theta_c) + k_c(\theta_{cd} - \theta_c) \ . \quad (12)$$

When we do not use the force display control, θ_{cd} is equal to θ_c. In this case, the motion dynamics of the steering caster is the same as that shown in eq.(7).

C. Braking Control Method

To implement the power steering and force display controls, we have to derive the braking forces of the right and left wheels of the cycling wheelchair $\boldsymbol{F}_{bw} = [f_{br},\ f_{bl}]^T$ using the Jacobian $\boldsymbol{J}$ as follows:

$$\boldsymbol{F}_{bw} = \boldsymbol{J}^T\ \boldsymbol{N}_b \ , \quad (13)$$

where $\boldsymbol{N}_b = [0,\ n_b]^T$ denotes the braking moment determined by eq.(10).

In most cases using the above equation, the sign of the braking torque for each wheel is the opposite for generating the moment around the center of the cycling wheelchair. However, the servo brake cannot generate active torque, that is, the output direction of the braking torque is restricted by the rotational direction of the wheel. Thus, we have to consider a feasible braking control region and then determine the appropriate braking torques of the wheels [7].

As explained above, to achieve rotational motion based on the moment expressed by eq.(10), one of the wheels must accelerate. However, the braking system cannot generate wheel acceleration, and in this case, the desired moment is outside the feasible braking control region $A(U)$ as shown in Fig. 4. Therefore, a wheel that can output an adequate braking torque generates a high braking torque to achieve the braking moment shown in eq.(10).

We determine the output braking force $F_o = [f_{or},\ f_{ol}]^T$, which is applied to each wheel using the following equation when the wheelchair turns left.

$$f_{or} = \frac{n_b - f_{ol}(-T/2)}{T/2} \ , \quad (14)$$

where T is the distance between the right and left wheels. As the angular velocity of the left wheel is decreased and cannot generate a large braking force, the right wheel compensates for this loss. Conversely, when the wheelchair turns right, we have

$$f_{ol} = \frac{n_b - f_{or}(-T/2)}{T/2} \ . \quad (15)$$

There are multiple solutions for f_{or} and f_{ol} in the above equations. We have to consider how the output braking force/moment in the feasible braking control region can be determined. As shown in Fig. 4, the braking moment determined by eq.(10) is generated by combining the output braking force/moment with a compensation force provided by a part of the human applied pedaling force. This suggests that steering assist control is achieved using the braking

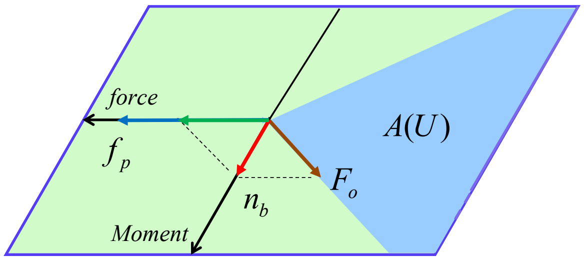


Fig. 4. Braking Control based on Feasible Braking Control Region

moment generated by the servo brakes in combination with the human pedaling force. In general, the user can apply a large force to the pedal, and this can be used to compensate for the insufficient force provided by the arms for steering. However, the greater the compensation force, the greater the burden on the user. To reduce this burden, the braking force/moment should be determined at the boundary of the feasible braking control region as shown in Fig. 4.

IV. Experiments for Assessing Steering Assist Control

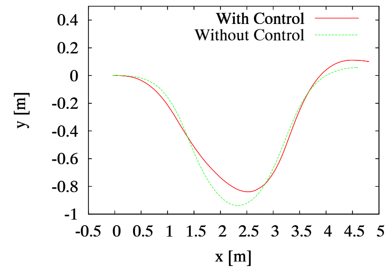
We modified a cycling wheelchair according to the proposed methods and experimentally confirmed their validity. First, we evaluated the power steering control of the cycling wheelchair, followed by the force display control, which enabled path following functionality. To evaluate the experiments, we attached a 6-axes force/torque sensor (WDF-6A200-3, WACOH-TECH Inc.) to the steering handle and measured the force applied by the user during the experiments. This sensor did not user any controls of the cycling wheelchair. The experiments were performed by a healthy subject, who could operate the cycling wheelchair smoothly. The results are presented in the following sections.

A. Power Steering Control

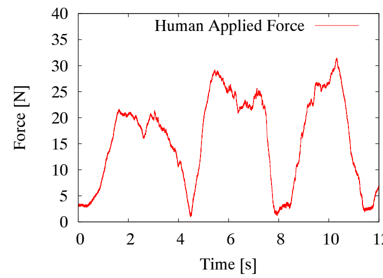
In this experiment, we assumed that the elastic coefficient of the wire k in eq.(8) was 2.5 Nm/rad, and conducted the experiment by specifying the power assist ratio α. First, we specified the ratio as $\alpha = 1$. This equation indicates that the experiment was performed without power steering control. Next, we specified the ratio as $\alpha = 2$ to reduce the required force by half.

In these experiments, the subject operated the cycling wheelchair to generate rotational motion. Fig. 5(a) shows the paths of the cycling wheelchair with $\alpha = 1$ (without control) and $\alpha = 2$. The subject attempted to move the cycling wheelchair along the same path in each experiment. Fig. 5(b) and (c) express the absolute value of the forces applied to the steering handle during the experiments. These results indicate that the power steering control reduces the force required for steering by approximately half.

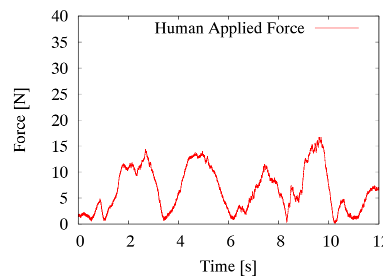
As the proposed method requires both braking force/moment and human pedaling force to implement power steering control, the user must expend greater effort in pedaling. However, the subject did not perceive his



(a) Path of Cycling Wheelchair



(b) Force (without control)



(c) Force (α = 2)

Fig. 5. Experimental Results for Power Steering Control

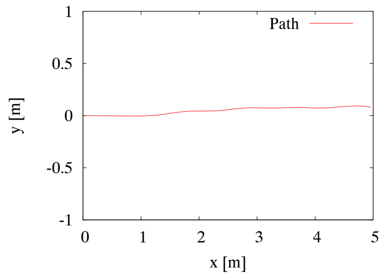
exertions to be excessive during the experiment because he could apply greater force to the pedal relative to the compensation force of the power steering control. We will evaluate user exertion quantitatively in a future study.

B. Force Display Control

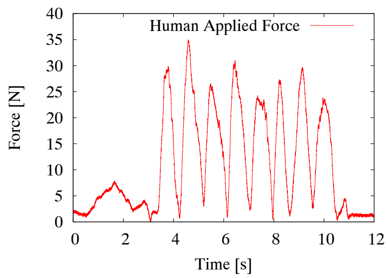
In this set of experiments, we displayed the force to the steering handle based on environmental information. First, we specified two types of paths to the cycling wheelchair: straight and S-shaped. We did not use the power steering control to evaluate only force display control. That is, $\alpha = 1$.

The desired angle of the steering caster θ_{cd} was initially specified as zero. In this experiment, the cycling wheelchair moved along a straight path. The subject turned the steering handle in an oscillatory manner to mimic a patient with severe impairment of the upper extremities. The results in Fig. 6 show that the subject moved along a straight path, although he applied a large force to rotate the cycling wheelchair.

Next, we specified the desired angle of the steering caster as $\theta_{cd} = 45 \sin x$ degree, where x is the moving distance of the cycling wheelchair with respect to the forward direction. Using this equation, the cycling wheelchair moved along an

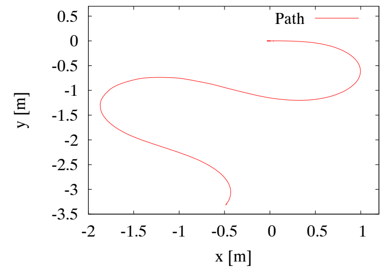


(a) Path

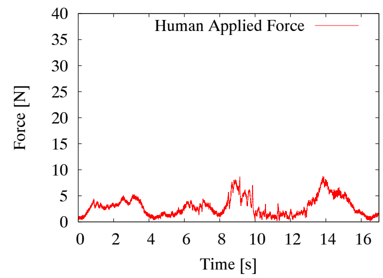


(b) Force

Fig. 6. Experimental Results for Following a Straight Path



(a) Path



(b) Force

Fig. 7. Experimental Results for Following an S-shaped Path

S-shaped path. If the wheelchair has a sensor to detect a line on the ground, we can calculate the desired angle of the steering caster from the sensor data. In this experiment, the subject did not apply a large force to the steering handle, and moved his hand according to the force display information at the steering handle.

The results of this experiment are shown in Fig. 7. Fig. 7(a) shows the actual path of the cycling wheelchair and Fig. 7(b) shows the force applied to the steering handle. Although the cycling wheelchair achieved rotational motion, the human force applied to the steering handle was very small compared to that observed in the previous experiment shown in Fig. 5(b). This result shows that the force display guides the user's hand appropriately based on environmental information.

V. Conclusions and Future Work

In this study, we designed a power assist control for the steering handle of a cycling wheelchair. In addition, we implemented force display control at the steering handle based on environmental information. The proposed methods were realized by controlling the servo brakes attached to the wheels of the cycling wheelchair. We applied the proposed methods to the newly developed cycling wheelchair and experimentally verified their validity.

The cycling wheelchair has great potential not only as a rehabilitation device but also as a mobility assist device to improve participation in daily activities. Many patients have already used cycling wheelchairs for rehabilitation, and most of them want to use these wheelchairs in outdoor environments. Medical doctors and therapists have also evaluated the cycling wheelchair. In future research, we intend to further evaluate our proposed control methods by performing extensive tests with patients, medical doctors, and therapists, and will improve the system based on their feedback.

Acknowledgments

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