A Phantom-Sensation Based Paradigm for Continuous Vibrotactile Wrist Guidance in Two-Dimensional Space

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Motion Control of Cycling Wheelchair with Continuously Variable Transmission and Clutch

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Abstract— **The cycling wheelchair, which is pedaled like a bicycle, is an effective device for improving its user's activities of daily living. It is commercially available in Japan, and is currently being used by hospitals and rehabilitation facilities. However, many patients, particularly those with lower-limb disabilities, would like to use it both indoors and outdoors. Indoor use requires relatively slow forward and backward movement combined with a small turning radius, all of which increase maneuverability in confined spaces. In contrast, outdoor use requires the cycling wheelchair to move faster with bicycle-like mobility and a freewheel mechanism for comfortable driving, although the latter precludes pedaling backward. In addition, the user would like to ascend slopes easily and descend them safely. Hence, the cycling wheelchair requires a transmission system that can change the gear ratio based on the environment, and an automatic braking system. In this paper, we discuss in more detail the assistive functions required of the cycling wheelchair, and we realize them by developing a new cycling wheelchair with a continuously variable transmission and clutch. In addition, we conduct experiments to illustrate the validity of the proposed cycling wheelchair and its control method.**

I. INTRODUCTION

Various assisted-mobility systems have been proposed to help elderly and disabled people. For example, walker-type assistive systems can help some people with lower-limb disabilities to walk [1], [2], [3], thereby maintaining and possibly even improving their ambulatory abilities. However, people with severe lower-limb disabilities have to use a wheelchair. In general, there are three types of wheelchair: manual wheelchairs driven by a pair of hand-rims on the sides, electric wheelchairs driven by actuators, and recently proposed robotic wheelchairs [4], [5]. A wheelchair extends the activities of daily living of its user: it improves his or her quality of life by allowing movement that is faster than walking, and allows greater distances to be covered. In addition, wheelchairs lower the risk of falling accidents associated with walker-type systems. However, long-term wheelchair use can cause problems if the lower limbs are not used sufficiently, often referred to as lower-limb disuse syndrome. Although rehabilitation is one way to prevent lower-limb disuse syndrome, patients tend to dislike it because it is time-consuming, expensive, and hard work. Therefore, we seek to develop a new assisted-mobility system that is as easy to operate as a wheelchair while affording exercise for the lower limbs like a walker-type system.

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Takahashi et al. have proposed a new type of wheelchair known as the "cycling wheelchair," which is pedal driven like a bicycle [6]. It was developed initially for rehabilitation at hospitals and rehabilitation facilities, and is operated by functional electrical stimulation (FES) of the lower limbs. However, after extensive clinical tests, it was discovered that many patients with hemiplegia, paraplegia, or Parkinson's disease could pedal a cycling wheelchair without FES [7], [8]. Cycling wheelchairs are now commercially available in Japan [9], and many patients use them at hospitals and rehabilitation facilities. However, most users would like to use their cycling wheelchairs outdoors as well as indoors, as they can with manual and electric wheelchairs, because this would allow them to move independently while exercising their lower limbs, thereby improving their quality of life. To use a cycling wheelchair outdoors, we have to consider obstacles such as slopes and steps. However, the current crop of commercial cycling wheelchairs have no functionality to deal with such obstacles.

In this paper, we propose a new cycling wheelchair for both indoor and outdoor use. Indoor use requires high maneuverability in a confined space, which equates to being able to pedal forward and backward and turn with a small turning radius. Current commercial cycling wheelchairs already afford this level of mobility, as discussed in more detail in the next section. In contrast, for outdoor use, the cycling wheelchair is required to move faster with bicycle-like mobility and a freewheel mechanism for comfortable driving. This mechanism enables the user to coast without having to keep turning the pedals, although it does preclude pedaling backward. Bicycle-like mobility also enables a caregiver to push the cycling wheelchair from behind without the user having to turn the pedals simultaneously. In addition, the user should be able to ascend an upward slope easily and descend a downward one safely. Hence, the cycling wheelchair should have a transmission system that can change the gear ratio based on the environment, as well as an automatic braking system for downhill stretches. The braking system would also be useful for avoiding obstacles or steps if the wheelchair could detect them by means of sensors.

Given the above requirements, we seek to design a cycling wheelchair that has bicycle-like mobility, a freewheel mechanism, a variable transmission, and braking control. The contribution of this paper is to design a new cycling wheelchair that realizes these mechanisms and assistive functions to support its use both indoors and outdoors. In section II, we introduce the mechanism used in commercial cycling wheelchairs and point out its advantages and remain-

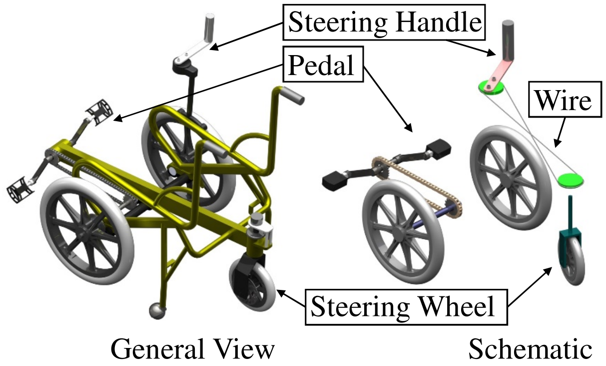


Fig. 1: Commercial cycling wheelchair

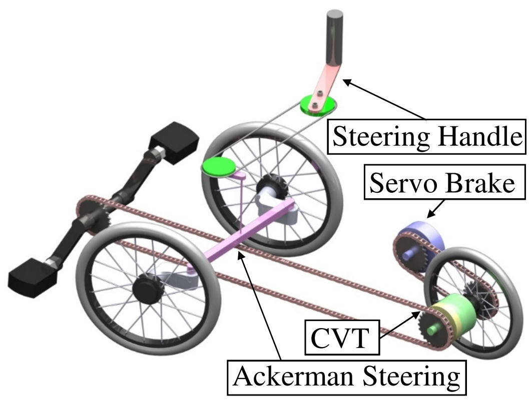


Fig. 2: Sports-type cycling wheelchair

ing problems for outdoor use. In section III, we propose a new mechanism for the cycling wheelchair involving a continuously variable transmission (CVT) and a clutch to realize all of the required functions explained in section II. We also explain the conditions that determine the behavior of these functionalities, and propose motion-control methods for several assistive functions. In section IV, we present the results of experiments to assess the validity of the proposed mechanism and its control methods, and we conclude in section V.

II. Problems of Current Commercial Cycling Wheelchairs

A. Driving Mechanism

A typical commercial cycling wheelchair and its mechanism are shown in Fig. 1. It is equipped with pedals, the torque of which is transmitted directly to only the left wheel via a chain and sprockets; the right wheel is passive. Because the user-applied torque is transmitted directly, the user can pedal the cycling wheelchair both forward and backward easily. The inertia of the wheelchair is conveyed via the direct torque-transmission mechanism back to the pedals when the wheelchair moves, thereby rotating the pedals with no need for additional pedaling torque from the user. This is what allows many people with lower-limb disabilities to propel a commercial cycling wheelchair without the need for powered assistance.

However, the direct torque-transmission mechanism makes it difficult to stop at high speeds. Even able-bodied users cannot stop the high-speed rotation of the pedals by themselves. To prevent the speed from becoming excessive, a bicycle-like handbrake is attached to the steering handle. However, if the user panics in a dangerous situation, he or she may lose control of the handbrake. In addition, with direct transmission, a cycling wheelchair cannot be pushed from behind by a caregiver as a conventional wheelchair without the user's feet rotating with the pedals, which is uncomfortable.

B. Maneuverability of Rotational Motion

In addition to its left and right wheels, the cycling wheelchair has a steering wheel that is connected to the steering handle via a wire. The cycling wheelchair has one degree of freedom around the instantaneous center of rotation determined by the angle of the steering wheel. As explained in the previous subsection, the pedaling torque is transmitted to only the left wheel. This causes the user to perceive right and left turns differently because turning left requires a larger pedaling torque.

C. Difficulty on Slopes

A commercial cycling wheelchair has no actuators with which to realize assistive functions such as power assistance, speed control, obstacle/step avoidance, or gravity compensation on a slope. From the feedback obtained from TESS, a company that sells cycling wheelchairs, speed control is required on downward slopes to use the wheelchair safely outdoors because going too fast can be very dangerous. In addition, if possible, climbing assistance is desirable on upward slopes to allow the user to ascend them using minimal pedaling torque.

D. Required Functions for New Cycling Wheelchair

As explained in the previous sub-sections, several functions are necessary to be able to use a cycling wheelchair safely outdoors. These have braking control, variable transmission for slopes, forward and backward movability, propulsion from behind by a caregiver without the pedals turning, improved turning maneuverability, and a small turning radius. These functions are summarized in Table I.

Previously, we have developed cycling wheelchairs with several assistive functions [10]–[13] based on a commercial cycling wheelchair. We have also proposed a sports-type cycling wheelchair with CVT and a servo brake, which can move faster than commercial ones, and we have realized variable transmission and braking control for easy and safe use outdoors [14]. A mechanism of sports-type cycling wheelchair is shown in Fig. 2. However, we were unable to implement all of the functions given in Table I in a single cycling wheelchair.

Although our sports cycling wheelchair realizes many of the functions in Table I by having a CVT and a servo brake, it cannot be pedaled backward because it was originally developed for a wheelchair marathon and hence has a freewheel

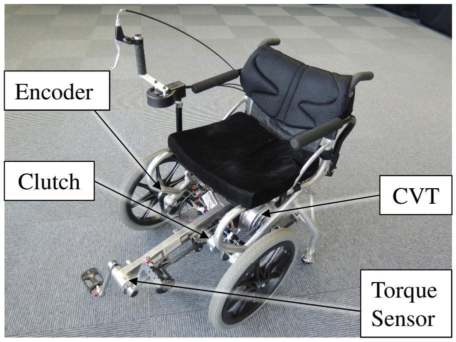


Fig. 3: General view of proposed cycling wheelchair

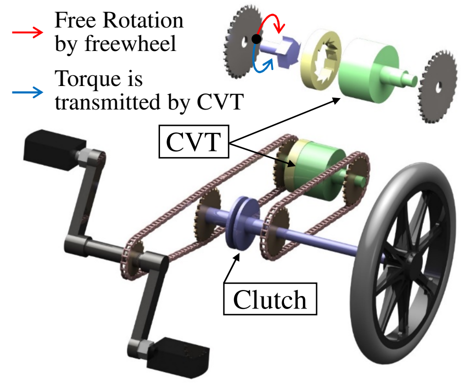
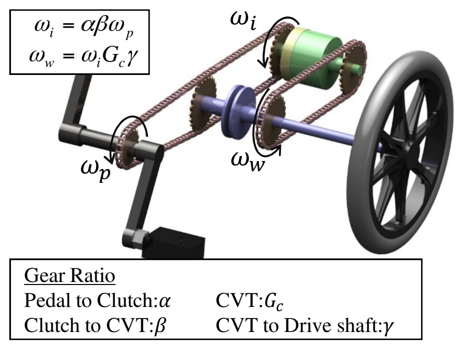
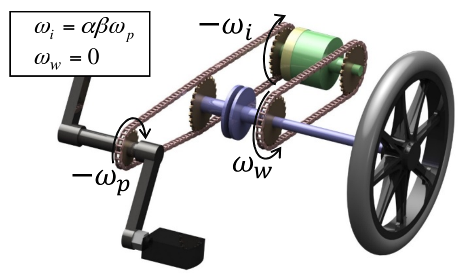


Fig. 4: Schematic of proposed system



(a) Forward



(b) Backward

Fig. 5: Driving conditions at disengagement of clutch

driving mechanism similar to that on a bicycle. In addition, it is equipped with Ackerman steering as in an automobile, and the turning radius is too large. Consequently, this prototype wheelchair cannot be used indoors. A cycling wheelchair that can be used both indoors and outdoors has to be smaller than our previous sports cycling wheelchair size and equipped with a better steering mechanism to give a smaller turning radius. Therefore, we seek to develop a new type of cycling wheelchair based on a commercial one but with a CVT and a clutch.

This new mechanism is also developed under the concept of passive robotics, which was proposed by Goswami et al. [15] to improve safety in human–robot interactions. The proposed cycling wheelchair with CVT and a clutch has no driving force and presents passive dynamics with respect to the user-applied force. Therefore, even if the user is unable to control the wheelchair appropriately, there is no driving force to move it unintentionally. In the next section, we explain the new mechanism of the cycling wheelchair.

III. New Cycling Wheelchair with CVT and Clutch

A. Mechanism of Cycling Wheelchair

We decided to attach a CVT and a clutch to a commercial cycling wheelchair. By controlling the gear ratio of the CVT and the clutch engagement, we can realize the assistive functions of the cycling wheelchair as given in Table I.

Figure 3 shows a general view of the proposed cycling wheelchair, and Fig. 4 shows a schematic of the clutch and CVT. In this system, the pedaling torque is transmitted to both the clutch and the input gear of the CVT. The clutch is connected to the drive shaft of the left wheel. If the clutch is engaged, the pedaling torque is transmitted to the left wheel. If the clutch is not engaged, the input gear of the clutch rotates freely and no pedaling torque is applied to the left wheel through the clutch. This means that the rotation of the wheel is not transmitted to the pedals, thereby allowing the wheelchair to be pushed as a conventional wheelchair without the pedals turning.

We also attached a CVT that was designed for a bicycle and was fabricated by Fallbrook Technologies, Inc. [16]. This CVT can change the gear ratio at a rate of 0.25 s^{-1}. The input gear of the CVT is connected to the pedals and the clutch via a chain, and the output gear of the CVT is connected via a chain to a sprocket attached to the driveshaft of the left wheel. A small motor controls the gear ratio of the CVT. Because the CVT was designed for a bicycle, it has a freewheel mechanism; the torque is transmitted in only one direction. We attached encoders to the wheelchair to measure the angular velocity of the wheels, the steering angle, and the pedaling angular velocity. In addition, to evaluate the user-applied pedaling torque, we attached a torque sensor to the pedals.

B. Conditions for Changing Feasible Motion

With the new hardware, we can control the gear ratio of the CVT and the engagement of the clutch. Based on the conditions of the CVT and the clutch, we can realize the assistive functions listed in Table I.

TABLE I: Required functions for cycling wheelchair

	Brake	Transmission (support on slope)	Feasible motion direction	Support by caregiver	Improved rotational motion	Turning radius
Commercial model [9]	Hand	Fixed	Forward/backward	User must pedal	No	Small
Regenerative-braking model [12]	Hand/auto	Fixed	Forward/backward	User must pedal	Yes, by differential gear	Small
Sports model [14]	Hand/auto	CVT	Forward	No user pedaling (freewheel)	Yes, by symmetry mechanism	Large
Proposed model	Hand/auto	CVT	Forward/backward	No user pedaling (clutch)	Yes, by CVT control	Small

1) Control of Clutch: Firstly, we consider controlling the engagement of the clutch. As shown in Fig. 5, if the clutch is not engaged, the pedaling torque is transmitted to the left wheel via only the CVT. However, because the CVT has a freewheel mechanism, we can only pedal forward; pedaling backward is not possible. If a user applies a pedaling torque to move forward and the pedals rotate at angular velocity w_p, the angular velocity ω_w of the wheels is given by

$$\omega_w = \alpha\beta G_c \gamma \omega_p, \tag{1}$$

where α is the gear ratio between the crank sprocket and the clutch sprocket, β is the gear ratio between the clutch sprocket and the sprocket on the input side of the CVT, γ is the gear ratio between the sprocket on the CVT's output and the sprocket on the drive shaft, and G_c is the gear ratio of the CVT. By changing the value of G_c, we can control the relationship between the velocity of the cycling wheelchair and the required pedaling torque based on the environment (e.g., upward slopes). We can also control the angular velocity of the pedaling, which is called cadence [17]. However, if the user applies a pedaling torque to move backward, the angular velocity ω_{cvt} of the output gears of the CVT is zero because of the CVT's freewheel, as shown in Fig. 5b, and hence the user cannot move backward.

2) Control of CVT: Next, we consider controlling the gear ratio of the CVT under the assumption that the clutch is always engaged. In this case, the pedaling torque is transmitted to the left wheel via both the clutch and the CVT. By changing the gear ratio of the CVT, we can change the feasible motion of the cycling wheelchair.

Firstly, we control the gear ratio of the CVT so that the angular velocity ω_{cvt} of the output sprocket of the CVT is greater than or equal to the angular velocity ω_i of the input sprocket of the CVT:

$$\omega_i \leq \omega_{cvt}. \tag{2}$$

This situation is illustrated in Fig. 6a. The angular velocities of the output/input transmission of the CVT are expressed as follows:

$$\begin{aligned} \omega_i &= \alpha\beta\omega_p, \\ \omega_{cvt} &= \frac{\alpha\omega_p}{G_c\gamma}. \end{aligned} \tag{3}$$

From the above equations, the gear ratio G_c of the CVT is determined as

$$G_c \leq \frac{1}{\beta\gamma}. \tag{4}$$

In this case, the pedaling torque transmitted to the left wheel via the clutch is not affected by the rotational motion of the CVT because of the freewheel mechanism. Therefore, the user can pedal both forward and backward. Note that the resultant angular velocity of the left wheel is expressed as

$$\omega_w = \alpha\omega_p. \tag{5}$$

This means that we cannot use the CVT to change the relationship between the velocity of the cycling wheelchair and the required pedaling torque. However, when the angular velocity ω_i of the input gear of the CVT is larger than the angular velocity ω_{cvt} of the output gear, the velocities are inconsistent between the input and output gears of the clutch. This situation is shown in Fig. 6b, and the angular velocity of the wheels depends on the gear ratio of the CVT as follows:

$$\omega_w = \alpha\beta G_c\gamma\omega_p. \tag{6}$$

In addition, in this inconsistent situation, the clutch slips even if it is engaged. The resulting slippage generates a braking torque at the pedals.

When disabled and elderly people use cycling wheelchairs, it is dangerous for them to drive at high speeds. Therefore, it is important to limit the speed to avoid accidents. In this research, we use the slip of the clutch as a brake. The torque diagram is shown in Fig. 7, where τ_c represents the torque generated by the slip of the clutch. If we assume that the wheel is rotated by an external force, then τ_c (the constant engagement torque of the clutch) is applied to the chain connected to the input of the CVT. From τ_c, we can derive the torque of the drive shaft from the CVT and sprocket as follows:

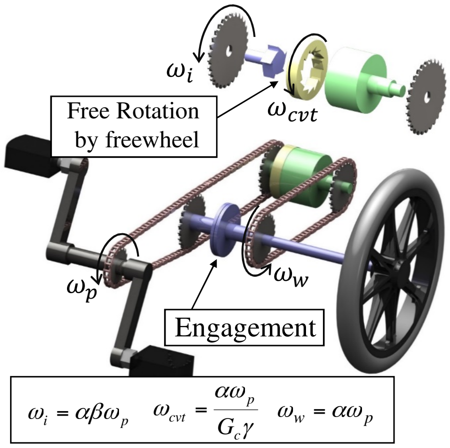
$$\tau_v = \frac{\tau_c}{G_c\beta\gamma}. \tag{7}$$

The torque from the wheel is distributed to the clutch and the CVT. Because the clutch and the CVT are closed systems, the respective torques τ_v and τ_c compete. The engagement torque is exerted in the reverse direction at the time of engagement. The braking torque τ_{bw} of the left wheel can be calculated as follows:

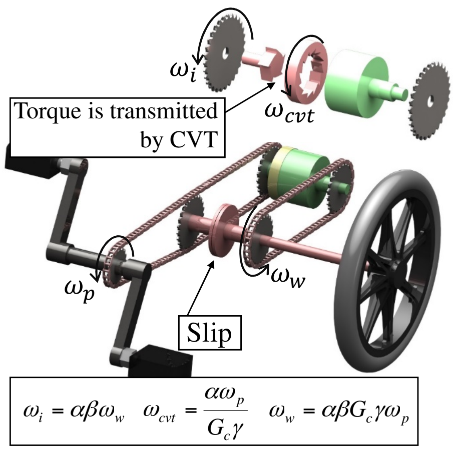
$$\tau_{bw} = \tau_c + \tau_v. \tag{8}$$

We can then rewrite Eqs. (7) and (8) as

$$\tau_{bw} = \frac{\tau_c(1 - G_c\beta\gamma)}{G_c\beta\gamma}. \tag{9}$$



(a) $\omega_i \le \omega_{cvt}$



(b) $\omega_i > \omega_{cvt}$

Fig. 6: Drivable conditions

This equation shows that we can change the braking force by changing the gear ratio of the cycling wheelchair's CVT. Equation (9) has the braking force increasing with the gear ratio of the CVT. In this research, we selected the following gears and clutch:

$$\begin{aligned} \alpha = \frac{22}{30}, \beta = \frac{30}{30}, \gamma = \frac{40}{28}, \\ G_c : 0.5 \sim 1.8, \\ \tau_c = 10.4[\mathrm{Nm}]. \end{aligned} \tag{10}$$

From these values, the magnitude of the braking force when moving forward is calculated as shown in Fig. 8. When the gear ratio of the CVT is close to 0.7, the braking force is

TABLE II: Feasible motion

	Input gear ratio of CVT: Output gear ratio of CVT		
	>	=	<
Forward	Brake	Allow	Allow
Backward	Allow	Allow	Brake

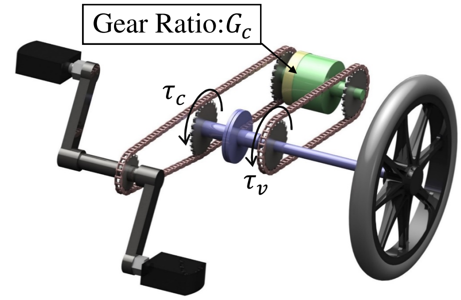


Fig. 7: Definition of the torque

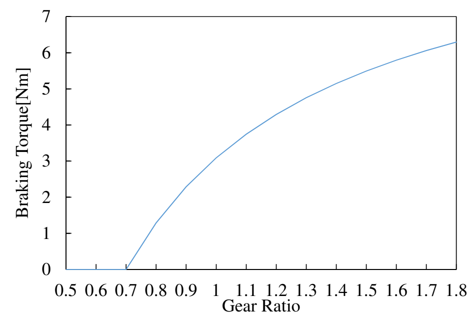


Fig. 8: Braking torque as a function of gear ratio

small. Therefore, it is possible to move forward or backward without a braking force. However, when the gear ratio of the CVT is large, the magnitude of the generated braking force also increases, making it difficult to move forward. In Table II, we summarize the conditions for changing feasible motion.

IV. Experiments

In this section, we present the results of several experiments conducted to assess the validity of the proposed mechanism. Experiments were conducted on forward and backward movement, braking control, and compensation for turning motion.

A. *Confirmation of Forward/Backward Movement*

We conducted experiments to confirm whether forward and backward motion is possible with the clutch engaged. The experimental results with the clutch disengaged are shown in Fig. 9 for a CVT gear ratio of 0.7. When pedaling in the forward direction (positive value), the wheelchair moves forward. However, pedaling backward does not move the wheelchair. The experimental results with the clutch engaged are shown in Fig. 10 for a CVT gear ratio of 0.7. In this case, the user can move the wheelchair forward or backward by turning the pedals accordingly.

B. *Braking Control*

We conducted experiments to confirm the braking force explained in the previous section. The experimental setup was a slope of about 3.2° as shown in Fig. 11. The subject was a man with no prior history of any physical or neurological disorders. The force acting on the cycling wheelchair is shown in Fig. 12, and the results are shown

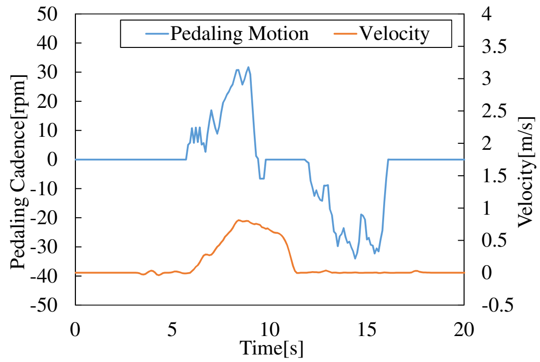


Fig. 9: Clutch disengaged

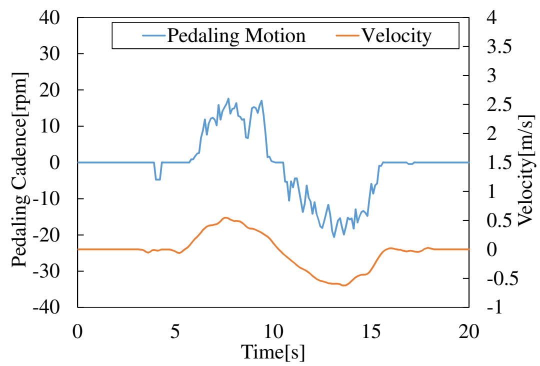


Fig. 10: Clutch engaged

in Fig. 13. Without control, it can be seen that the speed increased gradually as the user descended the slope. We also conducted experiments in which braking control was applied; the braking force is expressed by Eq. 9. The clutch was set to engage if the speed exceeded 1.0 m/s, and our results confirm that the acceleration decreased at speeds greater than that value.

To verify these results, we calculate them theoretically. The force acting on the cycling wheelchair is expressed as follows:

$$F = mg\sin\theta - Dv - \frac{\tau_{bw}}{R_w},$$
$$a = \frac{F}{m}, \tag{11}$$

where m is the total mass of the wheelchair and its user, g is the acceleration due to gravity, θ is the slope angle, D is the viscous damping coefficient, v is the speed, τ_{bw} is the braking torque, and R_w is the wheel radius. We assume $g = 9.8$ m/s^2 and a viscous damping coefficient of 0.005 Ns/m, which was determined experimentally. In addition, the total mass is 70 kg. Thus, the viscous damping is relatively small and the wheelchair would accelerate uniformly; the speed should therefore increase effectively linearly with time. The predicted behavior is indicated by the dotted lines in Fig. 13, whereas the actual behavior is indicated by the solid lines. It can be seen that the theoretical and experimental values are almost equal.

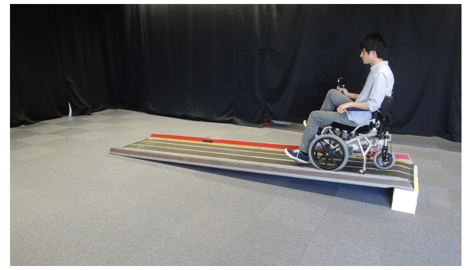


Fig. 11: Experimental setup

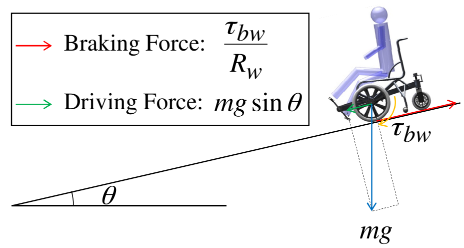


Fig. 12: Experiment on braking control

C. Compensation of Pedaling Torque by Rotational Motion

In daily life, it is often necessary to turn left or right when using a cycling wheelchair. However, the pedaling torque is applied only to the left wheel. Thus, different pedaling torques are involved when turning left or right: turning right involves a lower pedaling torque that does turning left. We propose controlling the CVT to reduce these torque differences. The instantaneous center of rotation is determined by the angle of the steering wheel, as shown in Fig. 14. The distance P between the instantaneous center of rotation and the center of the cycling wheelchair is expressed as follows:

$$P = W_w \tan\left(\frac{\pi}{2} - \theta\right), \tag{12}$$

where W_w is the wheelbase and θ is the steering angle (in degrees). From this instantaneous center of rotation and the wheelbase, the ratio of the left and right rotational velocities

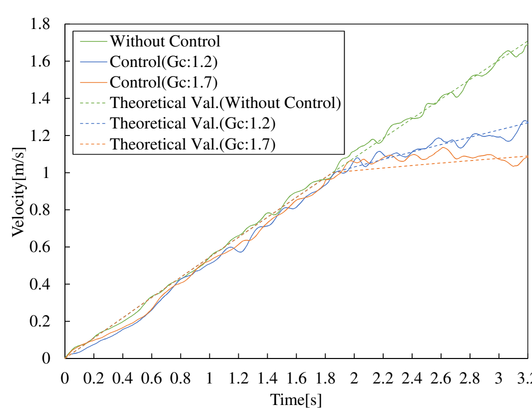


Fig. 13: Experimental results of braking control

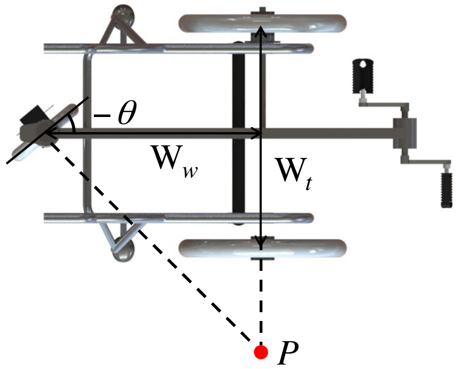


Fig. 14: Frame of cycling wheelchair

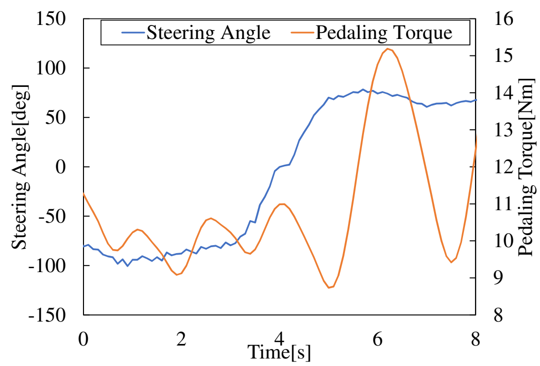


Fig. 15: Pedaling torque without control

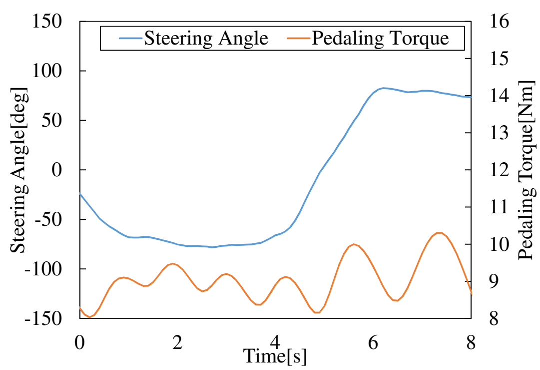


Fig. 16: Pedaling torque with proposed method

can be obtained as follows by using similar triangles:

$$\omega_{wr} : \omega_{wl} = P + \frac{W_t}{2} : P - \frac{W_t}{2}, \tag{13}$$

where ω_{wr} is the angular velocity of the right wheel, ω_{wl} is the angular velocity of the left wheel, W_t is the track, and K is the current CVT gear ratio. From this relationship, we can derive the CVT gear ratio as

$$G_c = K\frac{\omega_{wr}}{\omega_{wl}}. \tag{14}$$

From these equations, it is possible to calculate the rotational difference between the left and right wheels. Here assuming that the user output is constant, the torque is represented by the reciprocal of the angular speed. Therefore, by multiplying the reciprocal of the left and right rotational difference with the gear ratio of the CVT, the torque can be kept constant. Using this difference in the proposed control scheme makes it possible to reduce fluctuations in torque due to left and right turns.

The relationship between the steering angle and the pedaling torque when the gear ratio of the CVT is constant is shown in Fig. 15. Here, a positive steering angle represents a left turn. From this graph, we can confirm that the pedaling torque is increased when turning left. In contrast, in Fig. 16, we see a different effect when this difference is accounted for; the pedaling force is now similar when turning either left or right. Hence, the proposed approach is effective for maintaining pedaling torque while turning.

V. Conclusions and Future Work

In this paper, we proposed a cycling wheelchair equipped with a CVT and a clutch. By combining the CVT and clutch, we could achieve various assistive functions, namely pedaling forward or backward, braking using the clutch slip force, and compensating the pedaling torque when turning left or right. We conducted several experiments to confirm the validity of these functions.

However, the current clutch-engagement torque is too small to be able to apply an appreciable braking force. In future work, we intend to use a clutch with a larger engagement torque. Also, a dry clutch has a relatively small range of torque control, so we intend to try other clutch types, such as a powder clutch.

We also need to consider the fact that the optimal cadence is different for each person. The relationship between a person's joint torques and their cadence has been studied previously [18]. We intend to develop a pedal that can measure the pedal angle, the crank angle, and the force applied to the pedal. Using this information, we could estimate the joint torques by using a human model, and could determine the optimal cadence that minimizes the joint torques for each person. In addition, we would like to evaluate the performance of our system quantitatively.

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