New Control Method for Power-assisted Wheelchair Based on Upper Extremity Movement Using Surface Myoelectric Signal

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Abstract— **In this paper, we design a new controller of power-assist wheelchair. Conventional power-assist controllers only focus on the amplification of the human input force using torque sensor. Our proposed controller does not use a torque sensor but an electromyogram(EMG) sensor.**

Our designed controller is composed of EMG filter block, disturbance observer, direction discrimination function and fuzzy controller. We estimate whether the driver wants to propel forward or backward by the EMG signal placed at the triceps brachii muscle and the biceps brachii muscle. Then we provide power assist by combining the filtered myoelectric signal of the adductor pollicis muscle and the estimated human torque signal calculated by the disturbance observer. If the driver wants to stop or decelerate the wheelchair, our designed controller detects the changes of EMG signal and step down assist torque and realize vibration-free speed reduction. Experimental results show that the proposed power-assist method realizes quick and smooth power assist according to the driver's will.

I. Introduction

In Japan, according to demographic forecasts, the ratio of elderly people at least 65 years old is expected to top 40% in 2050. Therefore, there needs technical assistance for elderly people from the aspect of welfare engineering based on the coming aging society with fewer children. We focus on the power-assisted wheelchairs (shown in Fig.1) which are self-propelled wheelchairs with an electrical motor. There exist two advantages of using power-assisted wheelchairs. First, they reduce fatigue to go long distance. Second, they aid in maintaining the present abilities of the operators as well as rehabilitation. In other words, power-assisted wheelchairs have advantages of both self-propelled wheelchairs and electric-powered wheelchairs. For these reasons, power-assisted wheelchairs have become widespread. But current power-assisted wheelchairs on the market only amplify the power through the handrims and assist, leaving much to be improved. Therefore various assist methods have been proposed to provide smoother operation [1] [2] [3].

In this paper, we don't use torque sensor and joystick but EMG sensor for input interface considering developing cooperative control between human and machine. First, we develop two methods which assist wheelchair by filtered EMG signal of the adductor pollicis muscle and estimate input torque signal using disturbance observer respectively. Then we discuss the advantages and disadvantages and describe how to integrate these methods by the fuzzy controller. Additionally, we also develop decelerating method. When the power-assisted wheelchairs stop, it often happens undesirable vibration because of misconstruing input torque to stop wheelchair as reversely-directed input torque. To address this problem, we introduce EMG signal and step down assist torque according to the degree of urgency.

Fig. 1. Power-assisted wheelchair as an experimental tool (YAMAHA JW II)

II. Analysis of conventional method of power-assist wheelchair

A. Fundamentals of conventional method

In this paper, we first describe a conventional power assisting control, and carry out assistance using myoelectric signal. The conventional control of power-assisted wheelchairs simply amplifies the manual inputs from the push handrims with first order delay. The equation of the power-assist controller is

$$T_{assist} = K\frac{1}{1+\tau s}T_{human} \tag{1}$$

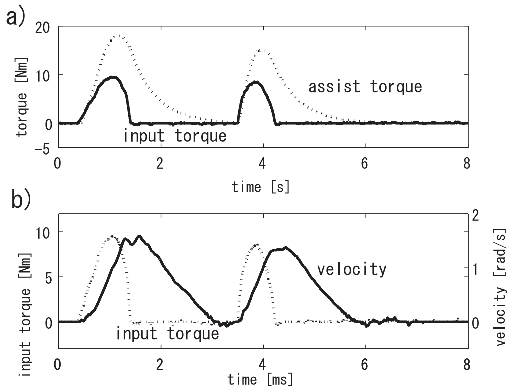

Fig. 2. Power-assist result of the conventional method

where K is the power-assist-ratio, T_{human} is input torque from the push hanrim, T_{assist} is the amplified torque from the push handrims and τ is the time constant of first order delay. τ should be a suitable value to realize an inertia for the wheelchair. Therefore, τ at the beginning of propelling should be small value and should be large at the ending, as shown by the following relations.

$$\tau = \begin{cases} \tau_{fast} = 0.1[s] & \dot{T}_{human} > 0 \\ \tau_{slow} = 0.5[s] & \dot{T}_{human} < 0 \end{cases} \quad (2)$$

B. Experimental result of conventional power-assist method

Experiments were carried out with real time control using YAMAHA JWII wheelchair with ARTLinux PC, with a torque sensor, gyro sensor and rotary encoder. The behavior of this controller is shown in Fig.2. However, there exists trade-off between comfortable ride and safety. When τ_{fast} is set small value, acceleration become sudden and wheelchair overreacts to unintended input torque and road irregularity. On the other hand, when τ_{fast} set big value, rising time of wheel speed become slow and driver doesn't achieve enough benefit from power-assist. These problems tend to cause driver discomfort and dangerous feeling. Therefore we solve these problems by using a myoelectric sensor, rather than a conventional torque sensor.

III. Signal processing of EMG

A. EMG signal

Electromyogram (EMG) records action potential of the muscle fiber which causes contraction of the muscles. EMG is established as an evaluation tool for applied medical rehabilitation, physical education, ergonomics, and so on [4]. There are many engineering applications such as myoelectric hand, myoelectric switch, robot suit and wheelchair [5]. The advantages of introducing an electromyography signal for power-assist control of the wheelchair are as follows:

1) Control with no delay
2) Ability to detect how much power is applied
3) Ability to assist people who do not have enough power to propel a wheelchair

First, a myoelectric sensor measures a signal that occurs when the brain orders the muscle to propel the wheelchair, not a change of rotational angle that occurs when wheelchairs are propelled. We can carry out the control without delay as the myoelectric signal can be measured earlier than human movement.

Second, the myoelectric signal enables measurement of how much power is applied by the operator which cannot be obtained from a change of angle.

Third, By using an electromyogram sensor, operators can be provided with power-assist and also the effect of rehabilitation even if they have little grip force or having difficulties in propelling the handrims because of disorder in hands.

B. EMG sensor

There are many kinds of EMG sensor, but we use a dry surface electrode as it is noninvasive and easy to apply. Shown in Fig.3, we carried out measurement by applying 2.5[V] to a reference electrode, having surface electrode attached to the adductor pollicis muscle and the triceps brachii muscle and the biceps brachii muscle. As in Fig.3, EMG measured by surface electrode on the skin is called surface electromyogram(sEMG). It is suitable for measuring entire muscle activation because it is a temporal and spatial summation of the action potential of the muscle fiber under the surface electrode. The sMEG is 1200 times amplified by the differential amplifier as the raw EMG signals are between a few microvolt and 2-3 millivolt. The sampling rate is 1[kHz] through a 12-bit A/D converter. The human subject was a 23-year-old healthy man.

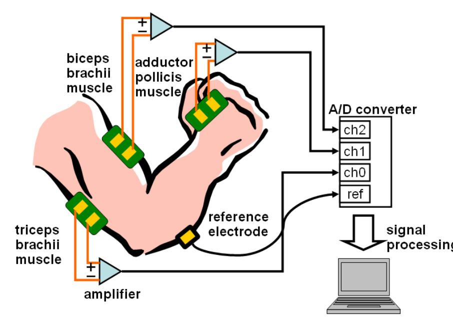

Fig. 3. Measurement system

C. Signal processing of EMG

The raw signals of EMG and encoder signals when a driver propels the handrims twice are shown in Fig.4. The high-frequency EMG signals are seen before the driver propels handrims. Here, we define the grasping phase as the range of about 0.5[s] before and after propelling the handrims. We also define the propelling phase as the range during which the driver propels the handrims and wheel velocity is increasing.

In order to extract grasping phase from EMG signals, it is necessary to apply signal processing to the EMG signals of

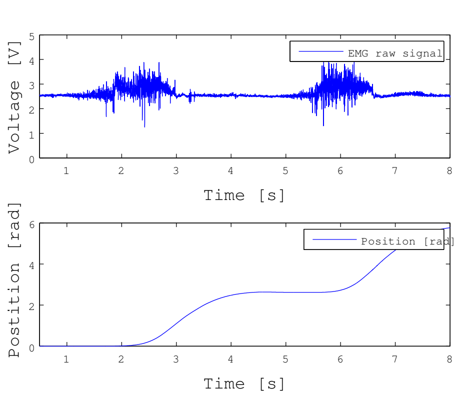

Fig. 4. EMG and encoder recording

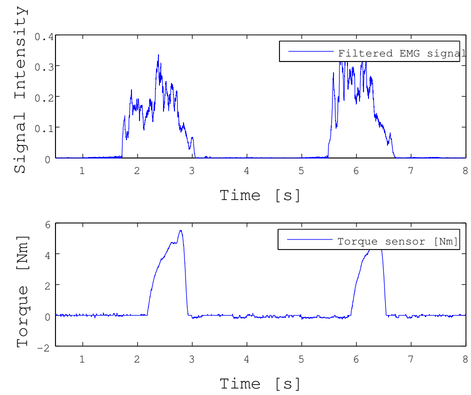

Fig. 6. Comparison between filtered EMG signal and input torque via the torque sensor

the adductor pollicis muscle. The signal processing of EMG signals is shown in Fig.5. If the sEMG signals are low-pass filtered, it is difficult to distinguish between overlapping noise and low-pass filtered EMG signals at the grasping phase. Therefore, the sEMG signals are digitally high-pass filtered with a cut-off frequency of 100 [Hz]. Next, the signals are full-wave rectified and smoothed by taking a moving average per 50 points as the following equation. Finally, subthreshold signals are cut to zero.

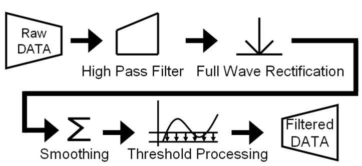

Fig. 5. The block diagram of the filter component

The filtered EMG signal of the adductor pollicis muscle is shown in Fig.6. Compared with input torque signals from the torque sensor, the filtered EMG signals occur at the grasping phase. Therefore we can assist a wheelchair at the grasping phase according to filtered EMG signals since the larger the filtered EMG signals are, the stronger the grasping power of driver is.

D. Power-assist method by filtered EMG signal

The block diagram of the power-assist method by the filtered EMG signal is shown in Fig.7 where $\tau = 0.2$ when wheel acceleration is positive and $\tau = 0.5$ when wheel acceleration is negative.

The experimental result is shown in Fig.8. Fig.8 shows that assist torque is generated about 0.3[s] earlier than torque sensor detect input torque. However there are two disadvantages. First, filtered EMG signal can't detect whether the driver wants to propel forward or backward. Second, filtered EMG signals are like a impulse signal at the propelling phase, and it is difficult to use filtered EMG signals as human propelling

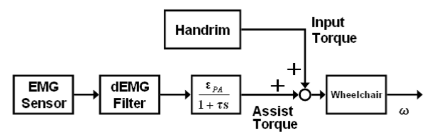

Fig. 7. Block diagram of power-assist method by filtered EMG signal

torque. Therefore, we use filtered EMG signal at the grasping phase.

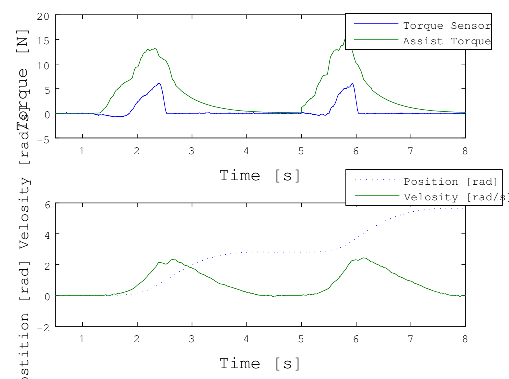

Fig. 8. Power-assist result by filtered EMG signal

IV. Disturbance Observer

A. Estimation of the propelling torque

As stated above, it is problematic to simply estimate that input torque of drivers and myoelectric signal are proportional. Therefore, we calculate assist torque by using a disturbance observer at the propelling phase. The block diagram of the disturbance observer is shown in Fig.9 where J_n is the nominal value of J, B_n is the nominal value of B and T_a is the time constant of the low-pass filter. We can estimate human input torque by calculating the difference between the commanded value of assist torque and the inverse plant of nominal model.

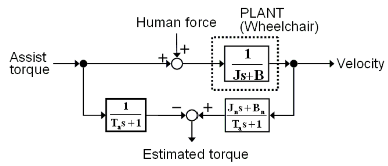

Fig. 9. Block diagram of the disturbance observer

The experimental result is shown in Fig.10. Compared with input torque signals from the torque sensor, although the disturbance observer doesn't estimate the grasping torque, estimated torque signals from the disturbance observer can estimate propelling torque.

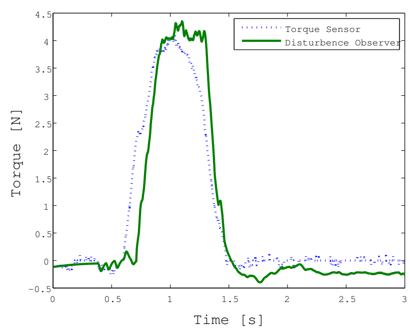

Fig. 10. Estimation result of the human force calculated by the disturbance observer

B. Power-assist method by disturbance observer

The block diagram of the power-assist method by disturbance observer is shown in Fig.11 where $\tau = 0.2$ when wheel acceleration is positive and $\tau = 0.5$ when wheel acceleration is negative.

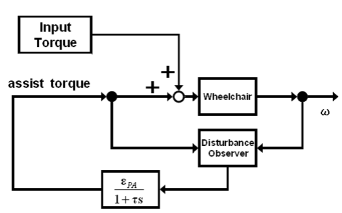

Fig. 11. Block diagram of power-assist method by disturbance observer

The experimental result is shown in Fig.12. Although estimated torque signal is about 0.2[s] later than torque sensor, disturbance observer estimates almost exactly propelling torque. Therefore, we use disturbance observer at the propelling phase.

V. Design of Proposed Power-Assist Controller

A. Fuzzy Controller

At the grasping phase, we regard filtered EMG signals as the intensity of driver's intention of forward movement and at

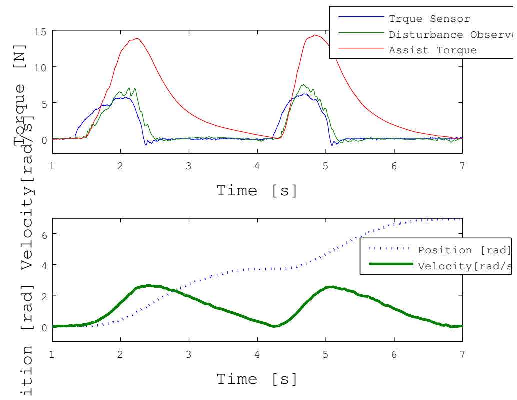

Fig. 12. Power-assist result by the disturbance observer

the propelling phase, we estimate propelling torque calculated by the disturbance observer. Then, we combine filtered EMG signals and estimated torque signals by the fuzzy controller (Fig.13). K_1 and K_2 are constant value and F_1 and F_2 are variables which indicate effect extent of power-assist. These values change according to the wheel velocity. At the grasping phase when the wheel velocity is under v_α, we set K_1 as 1 and set K_2 as 0 so that power assistance is based on the filtered EMG signals. On the other hand, at the propelling phase when the wheel velocity is over v_β, we set K_1 as 0 and set K_2 as 1 value so that power-assist is based on the estimated torque signals. At the transition phase from grasping to propelling, K_1 is larger and K_2 is smaller in proportion to wheel velocity in order to realize smooth acceleration. v_α and v_β are variables which indicate the effect between filtered EMG signals and estimated torque signals. Direction is the plus or minus which decided by direction discrimination function. Direction discrimination function outputs plus when filtered EMG signals of the biceps brachii muscle is stronger than those of the triceps brachii muscle at the beginning time of grasping phase. On the other hand, when the filtered EMG signals of the triceps brachii muscle is stronger than those of the biceps brachii muscle at the beginning time of grasping phase, direction discrimination function outputs minus.

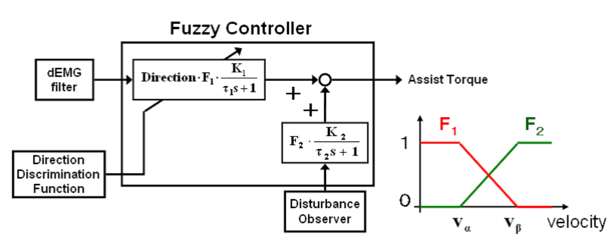

Fig. 13. Internal structure of fuzzy block

B. Block diagram of the proposed method and experimental result

The integrated block diagram of the proposed power assist method is shown in Fig.14.

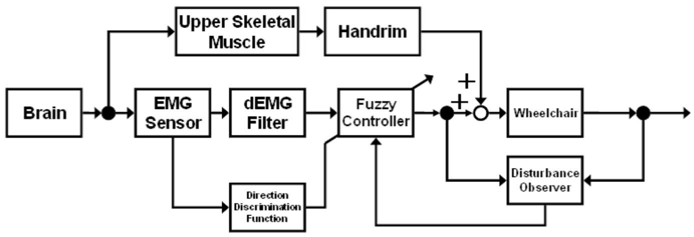

Fig. 14. Block diagram of the proposed method

Experiments where the human subject propels handrims forward and backward respectively using proposed method are performed. The results are shown in Fig.15. Compared between Fig.2 and Fig.15, we can see that assist torque of the proposed method is generated earlier than that of the conventional method. Shown in Fig.15, wheel velocity accelerates smoothly before driver propels handrims so that driver can propel with less power. When the wheelchair runs at the dirt road or lawn, it may need strong power for driver to start to move wheelchair. Proposed method has a beneficial effect on these situations.

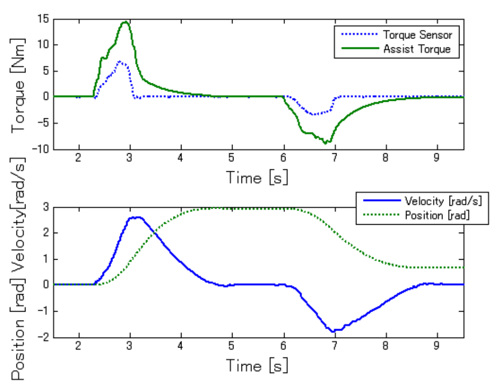

Fig. 15. Power-assist result of the proposed acceleration method

When the driver has a healthy upper body and easy to propel handrims, K_1 set as a small value. Fig.16(a) shows the experimental result when $K_1 : K_2 = 1 : 1$. Assist torque at the grasping phase is small and insusceptible to the filtered EMG signals. On the other hand, when the driver has a weak grip and have difficulty propelling handrims, K_1 set as a big value. Fig.16(b) shows the experimental result when $K_1 : K_2 = 5 : 1$. Assist torque at the grasping phase is big and susceptible to the filtered EMG signals. When we use proposed method for rehabilitation, we gradually reduce K_1 according to the degree of recovery. In this way, we can realize acceleration according to the driver's taste by changing parameters of the fuzzy controller.

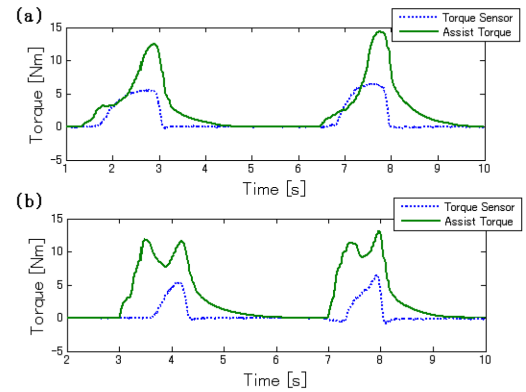

Fig. 16. Power-assist result ((a)K1:K2=1:1 (b)K1:K2=5:1)

VI. Deceleration Controller of Power-Assist Wheelchair

A. Experiment of conventional method at the deceleration phase

The behavior of the conventional controller is shown in Fig.17 when the driver stops the wheelchair quickly. Fig.17 shows that when the driver stops the wheelchair, a negative torque is applied to the wheelchair detecting the input torque against driver's will. Then, the driver reflexively pushes the handrims forward in response to the negative torque. Therefore, a positive torque is applied to the wheelchair, and positive and negative torque are applied to the wheelchair alternately. As a result, vibration occurs and it takes a long time to stop the wheelchair. The sEMG signals when the driver stops the wheelchair quickly is shown in Fig.17. Fig.17 shows that strong grasping power is necessary to stop the wheelchair at the acceleration phase during power assisting. This might be a burden for elderly people who have a weak grip. Therefore, from the viewpoint of safety, a deceleration controller which stops quickly with little input torque is required.

B. Design of deceleration controller

One of the typical cases of human behavioral pattern in face of danger is clenching his own hand. This is thought to be similar to wheelchair. We regard the filtered EMG signals as the degree of urgency and decrease assist torque according to the filtered EMG signals. Then we define the range from the time when the filtered EMG signals of both triceps brachii muscle and biceps brachii muscle are over threshold value to the time when the wheel velocity converge to zero as the stop phase. At the stop phase, assist torque is decreased gradually according to the equation (3). The equation of deceleration controller is

$$\tau_{assist} = \tau_{stop} \exp^{-K \frac{EMG_{stop}}{EMG_{max}} t} \quad (3)$$

where τ_{stop} is the assist torque and EMG_{stop} is the intensity of filtered EMG signals of the adductor pollicis muscle at the beginning of the stop phase. EMG_{max} is the maximum signal intensity of the adductor pollicis muscle which measured

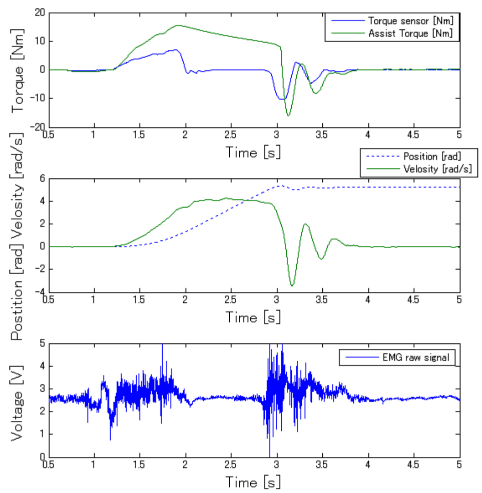

Fig. 17. Experimental result of the conventional method

before the experiment. Here, EMG_{stop} plays the role of a damping constant. The larger EMG_{stop} is, the sooner assist torque is decreased. This enables a driver to stop the wheelchair according to the driver's will.

C. Experiments

Experiments where the human subject stops the wheelchair using the proposed deceleration method were performed. The results are shown in Fig.18 and Fig.19. In Fig.18, the driver applies strong force to stop the wheelchair. In Fig.19, a driver applies weak force to stop the wheelchair. Shown in Fig.18 and Fig.19, assist torque and wheel velocity converge to zero without rocking vibration and wheel velocity in Fig.18 decreases earlier than that in Fig.19.Compared with Fig.17, we can see that proposed deceleration method is safer and more reliable than the conventional method.

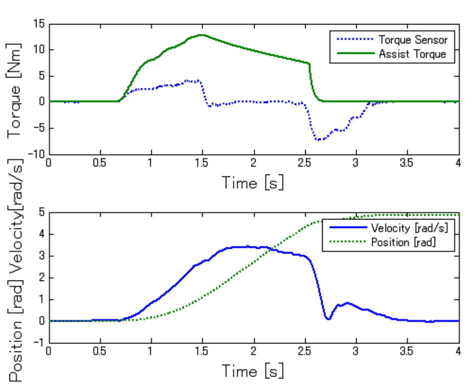

Fig. 18. Experimental result of quick stop by use of proposed deceleration method

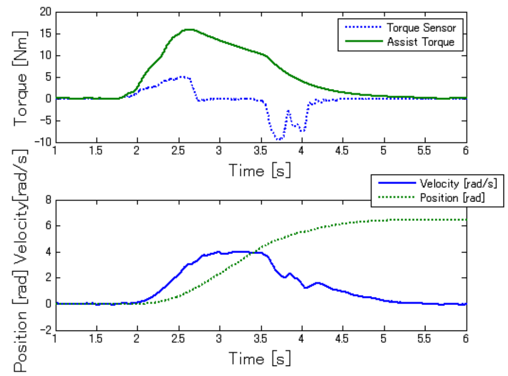

Fig. 19. Experimental result of slow stop by use of proposed deceleration method

VII. Conclusion

In this paper, we presented a new controller for power-assisted wheelchair based on the EMG signal. We explain the advantages of both filtered EMG signal and disturbance observer. Then we combine two methods by the fuzzy controller. At the grasping phase, we assist the wheelchair by using the filtered EMG signal so that the wheelchair can move before the driver propels handrims. The direction of power-assist is decided depending on the relative strength of filtered EMG signals between the triceps brachii muscle and the biceps brachii muscle. At the propelling phase, we assist wheelchair by using the estimated torque signal so that the wheelchair can move according to the human propelling torque. It also provides acceleration according to the driver's taste by changing K_1, K_2, v_α and v_β of Fig.14. In addition, during deceleration, we step down assist torque according to the degree of urgency estimated by the filtered EMG signals. This enables to stable stop without rocking vibration.

The future work will consider the cooperation control of both right and left wheel. In addition, the evaluation of this proposed method in some elderly care facility is considered.

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