A New Control Method for Power-assisted Wheel Chair based on the 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 sensor and a disturbance observer. Then we decide assist torque by combining the filtered myoelectric signal and the estimated human torque signal calculated by the disturbance observer.**

Our designed controller carries out different power-assist method for acceleration and deceleration phase. During the acceleration phase, we provide power assist earlier than the driver's action. On the other hand, when the driver wants to stop the wheel chair, we step down assist torque according to the degree of urgency estimated by the filtered myoelectric signal. Experimental results show that the proposed power-assist method realizes fast, stable and smooth acceleration and safe deceleration.

I. Introduction

The number of people who have difficulty walking and need wheelchairs is increasing due to an aging population caused by low birth rate and advances in medical treatment. There are many types of wheelchair such as self-propelled wheelchairs, attendant-propelled chairs and powerchairs. Power-assisted wheelchairs(shown in Fig.1)which are self-propelled wheelchairs with an electrical motor, not only lighten the burden imposed on operators and make it easier to go long distance but also to aid in maintaining the present abilities of the operators as well as rehabilitation. For these reasons, power-assisted wheelchairs have become widespread. Also from a viewpoint of welfare engineering, it is an interesting field which combines human power and motor output power through the handrim. But current power-assisted wheelchairs on the market only amplify the power through the handrim and assist, leaving much to be improved. For the control of the welfare machines that will be used long time such as wheelchairs, power assistance by force amplification is not only a burden for drivers but also poses a danger of tipping over due to too much assist force, and might be an obstruct to the recovery of the drivers. Therefore various assist methods have been proposed to provide smoother operation [1] [2] [4]. In this paper, we focus on "grasping" and "propelling", which are the two main acts when using the wheelchair handrim. We propose a more comfortable power-assist method by using an electromyogram sensor, which enables us to extract the actions of "grasping" and "propelling", which cannot be realized by a conventional sensor that only detects shaft angle. The advantages of introducing an electromyography signal for power-assist control of the wheelchair are as follows:

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

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

1) 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.
2) For example, let us assume two situations. One is that driver does nothing on a wheelchair, the other is that driver grasps the handrims. These are the indistinguishable if we measure the signals which can be obtained from the sensor that detects shaft angle. But they are distinguishable if we introduce myoelectric signal using an electromyogram sensor. In this way, the myoelectric signal enables measurement of how much power is applied by the operator which cannot be obtained from a change of angle.

3) 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 handrim because of disorder in hands. This is a more effective method for many handicapped people because it can estimate human force and assist in situations where the force is difficult to be generated. Conventional power-assisted wheelchairs cannot realize this.

In this paper, we develop a method which assists wheelchair using the characteristics of a myoelectric signal. However, the myoelectric signal depends on the operators and repeatability is a problem because of the position of sensor and the skin condition. Thus, it is difficult to simply estimate that input torque and myoelectric signal are proportional. Therefore we introduce a disturbance observer to estimate the propulsion. Then we decide assist torque depending on the running condition. At the acceleration phase, we assist the wheelchair earlier than the driver's action by combining filtered myoelectric signal and estimated human torque signal calculated by the disturbance observer. On the other hand, when a driver wants to stop the wheel chair, we step down assist torque according to the degree of urgency estimated by the filtered myoelectric signal.

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

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

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 handrim 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.

$$\tau = \begin{cases} \tau_{fast} & \dot{T}_{human} > 0 \\ \tau_{slow} & \dot{T}_{human} < 0 \end{cases} \tag{2}$$

For example, our experiments adopt the following values respectively,

$$\tau_{fast} = 0.08[s], \tau_{slow} = 1.0[s]. \tag{3}$$

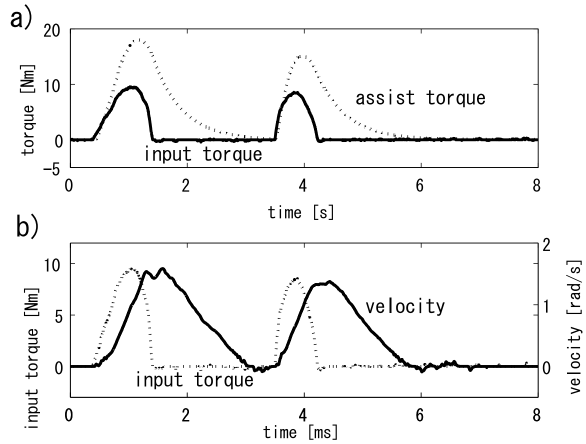

Fig. 2. Power-assist result of the conventional method

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. The result shows that there exist two possible points for improvement in the acceleration phase. First, rising time of wheel speed is slow. Second, acceleration is too sudden. 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 sensor

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 [3]. There are many kinds of EMG sensor, but we use a dry surface electrode MES01(MicroCube) as it is noninvasive and easy to apply. Shown in Fig.4, we carried out measurement by applying 2.5[V] to a reference electrode, having surface electrode attached to the adductor pollicis muscle. As in Fig.4, EMG measured by surface electrode on the skin is called surface electromyogram(sEMG). It is suitable for measuring

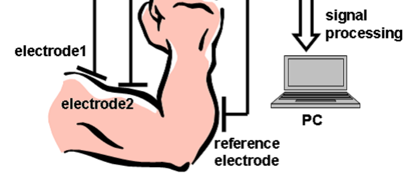

Fig. 3. Sensor system

entire muscle activation because it is a temporal and spatial summation of the action potential of the muscle fiber under the surface electrode. The schematic diagram of the measurement system is shown in Fig.3. Shown in Fig.3, 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.

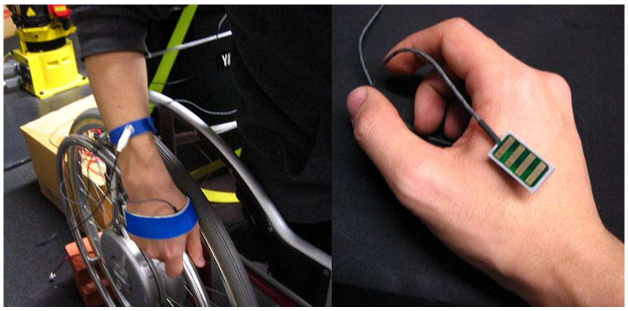

Fig. 4. EMG sensor

B. Signal processing of EMG

The raw signals of EMG and encoder signals when a driver propels the handrim twice are shown in Fig.5. The high-frequency EMG signals are seen before the driver propels handrims. Here, we define the operating condition of the driver as follows:

- grasping phase
- propelling phase
- coasting phase

The first phase is the grasping phase, which is the range of about 0.5[s] before and after propelling the handrim and wheel velocity is slow. The second phase is the propelling phase during which the driver propell the handrim and wheel velocity is increasing. The third phase is the coasting phase which is the condition that driver relaxes his/her grasp on the handrim and wheel velocity is decreasing. While cycling through these three phases, wheelchair is running.

In order to extract grasping phase from EMG signals, it is necessary to apply signal processing to the EMG signals. The signal processing of EMG signals is shown in Fig.6. 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 frecuency of 100 [Hz].

$$EMG_{hpf}(t) = \frac{s}{s+\tau} EMG_{surface}(t) \tag{4}$$

Next, the signals are full-wave rectified and smoothed by taking a moving average per 50 points as the following equation.

$$EMG_{filtered}(t) = \frac{1}{50} \Sigma_{i=0}^{49} |EMG_{hpf}(t-i) - EMG_{ave}| \tag{5}$$

Finally, subthreshold signals are cut to zero.

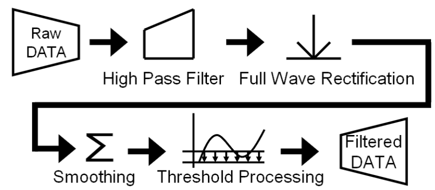

Fig. 6. The block diagram of the filter component

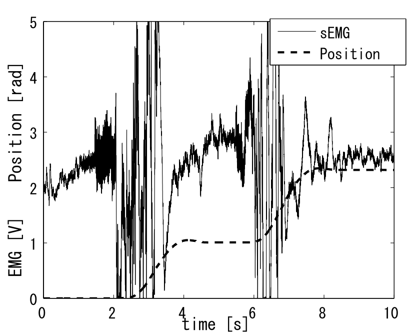

Fig. 5. EMG and encoder recording

The filtered sEMG is shown in Fig.7. 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. However, filtered EMG signals are like a impluse signal at the propelling phase, and it is difficult to use filtered EMG signals as human input torque. Then, we will discuss acceleration method in the following chapter.

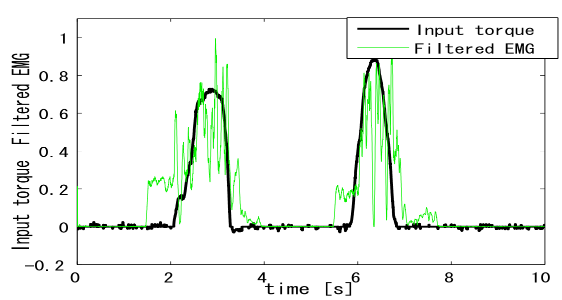

Fig. 7. Comparison between filtered EMG signal and input torque via the torque sensor

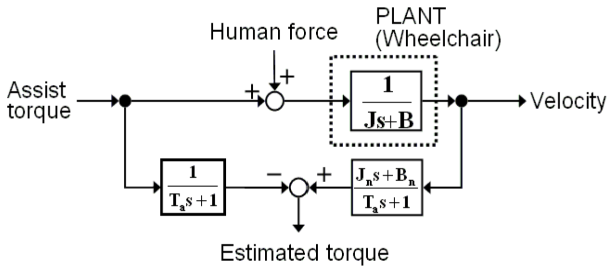

Fig. 8. Block diagram of the disturbance observer

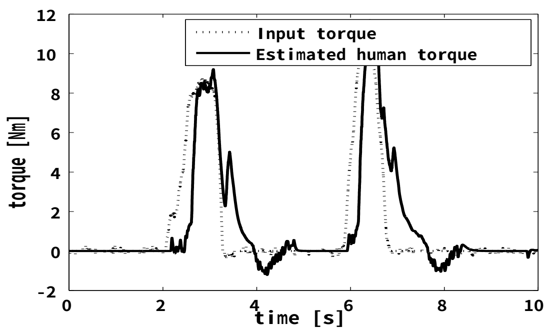

Fig. 9. Estimation result of the human force calculated by the disturbance observer

IV. Acceleration controller of power-assist wheelchair

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.8 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 touque and the inverse plant of nominal model.The experimental result is shown in Fig.9.Compared with input torque signals from the torque sensor, although the disturbance observer doesn't work at the grasping phase, estimated torque signals from the disturbance observer can estimate human input torque at the propelling phase. When we use a torque sensor we sense the only driver's input power, but when we use the disturbance observer we can also obtain care personnel's power because disturbance observer estimates whole input force which added to the wheelchair. Hence, this lightens the burden on the care personnel.

B. Design of power-assist controller

At the grasping phase, we regard filtered EMG signals as the intensity of driver's intention of forward movement and at the propelling phase, we estimate propelling torque calculated by the disturbance observer. Then, we combine filtered EMG signals and estimated torque signals. The block diagram is shown in Fig.10. K_1 and K_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 a large value and set K_2 as a small value 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 a small value and set K_2 as a large 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 acceleration in order to realize smooth acceleration. At the coasting phase, we use the same method as the conventional metod. v_α and v_β are variables which indicate the effect between filtered EMG signals and estimated torque signals. For exsample, we set v_α as a large value when the driver is a severely disabled person who has a weak grip. When we use proposed method for rehabilitation, we gradually reduce v_β according to the degree of recovery. In this way we can realize acceleration according to the driver's taste by chainging v_α and v_β.

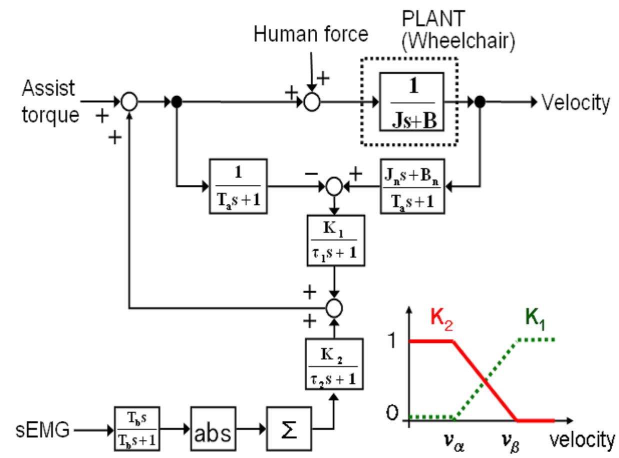

Fig. 10. Block diagram of the proposed acceleration method

C. Experiments

Experiments where the human subject propels handrims two times using proposed acceleration method are performed. The results are shown in Fig.11. Compared between Fig.2(a) and Fig.11(a), we can see that assist torque of the proposed method is generated earlier than that of the conventional method. Shown in Fig.11(b), wheel velocity accelerates smoothly before driver propels handrims so that driver can propell 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.

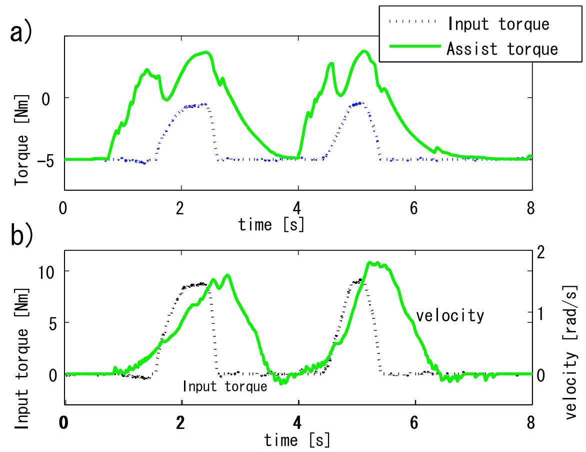

Fig. 11. Power-assist result of the proposed acceleration method

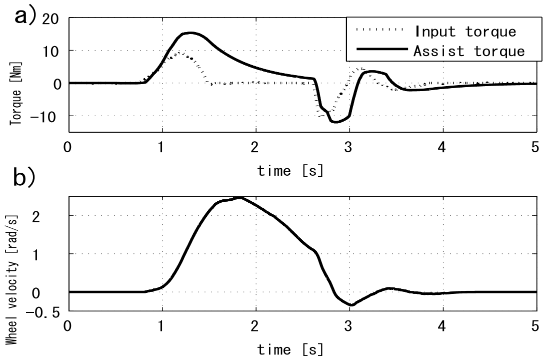

Fig. 12. Experimental result of the conventional method

V. 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.12 when the driver stops the wheelchair quickly. Fig.12 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 handrim 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.13. Fig.13 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.

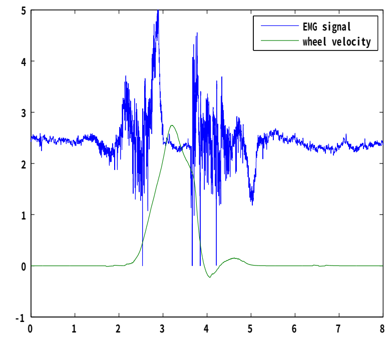

Fig. 13. EMG signal and wheel velocity

B. Design of deceleration controller

One of the typical case 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. The equation of deceleration controller is

$$\tau_{assist} = \tau_{stop} \exp^{-s_{EMG}t} \tag{6}$$

where τ_{stop} is the assist torque and s_{EMG} is the intensity of filtered EMG signals when the estimated input torque signal by the disturbance observer falls below the negative threshold value. When estimated input torque signal falls below the negative threshold value, assist torque is decreased gradually according to the equation (6) and s_{EMG} plays the role of a damping constant. The larger s_{EMG} 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 quickly using the proposed deceleration method were performed. The results are shown in Fig.14 and Fig.15. In Fig.14, the driver applies strong force to stop the wheelchair. In Fig.15, a driver applies weak force to stop the wheelchair.Shown in Fig.14 and Fig.15, assist torque and wheel velocity converge to zero without rocking vibration and wheel velocity in Fig.14 decreases earlier than that in Fig.14.Compared with Fig.12, we can see that proposed deceleration method is safer and more reliable than the conventional method.

For the evaluation of ride quality, we measure angular velocity of pitch surface by gyro sensor. The comparison between the proposed method and the conventional method is shown in Fig.16. In Fig.16, the unnecessary vibration occurs by the conventional deceleration method, on the oter hand, it is recognized that the vibration is suppressed by the proposed deceleration method. Thus, it can be said that our proposed deceleration method is more safe and comfortable for the driver.

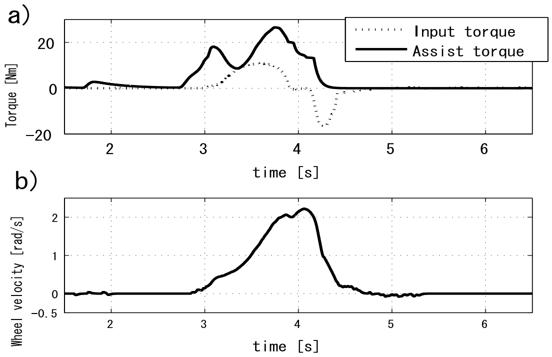

Fig. 14. Experimental result of quick stop by use of proposed deceleration method

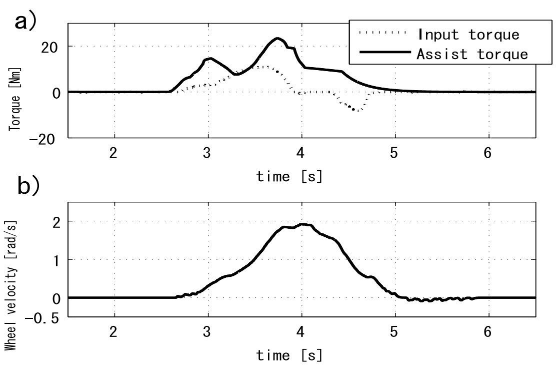

Fig. 15. Experimental result of slow stop by use of proposed deceleration method

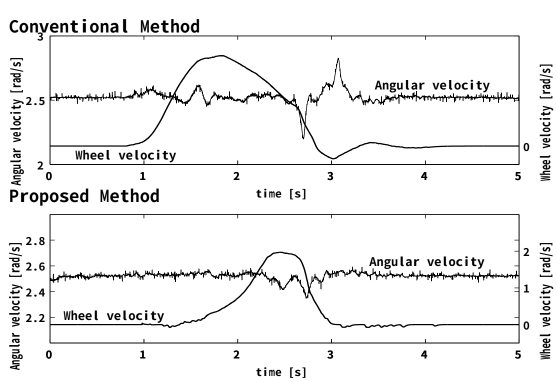

Fig. 16. Comparison of angular velocity between conventional and proposed method

VI. CONCLUSION

In this paper, we design a new controller for power-assisted wheelchair. At the acceleration method, we assist the wheelchair by combining filtered EMG signal and estimated torque signal calculated by the disturbance observer. 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. This enables to lighten the driver's burden when a driver starts to propel handrims. At the propelling phase, we assist wheelchair by using the estimated torque signal so that a caregiver's pushing force can be also amplified. This enables to lighten caregiver's burden. It also provides acceleration according to the driver's taste by changing K_1, K_2, v_α and v_β of Fig.10. On the other hand, 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.

From the view point of the welfare application, smooth control and ride quality are important. Experiments for many test subjects and subject rating are left for further investigation.

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