# COMPUTATIONAL MUSCLE REFLEX MODEL OF WHEELCHAIR USERS TRAVELING IN MOTOR VEHICLES

## *Evaluation of the Motion and the Myoelectric Potential of People with Disabilities*

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Abstract: In this study, a human motion computer model in a wheelchair was developed to evaluate the effectiveness of a seatbelt for people with disabilities traveling in a motor vehicle. The human model was composed of two rigid links and three masses. This model was characterized with muscle reflection defined by Hill's equation. A sudden stop experiment by using a carriage on which a wheelchair was fixed with a subject was performed to obtain the human muscle parameters and to evaluate the model. Volunteer subjects including people with disabilities participated in the experiment. The motion and muscle activity of a subject wearing a seatbelt were simulated by this model. The muscle reflection of people with disabilities was stronger than that of normal people in the case of not using a seatbelt, but in the case of using a seatbelt the muscle reflection of people with disabilities was similarly weak with normal people. The result of computer simulation showed that a seatbelt is more important for people with disabilities than for normal people.

## 1 INTRODUCTION

Wheelchair users traveling in motor vehicles have been increasing dramatically. There is a need to improve the safety standards for transporting wheelchair users (ANSI/RESNA, 2002) (W.E.Fisher, 1987). The authors carried out a questionnaire for wheelchair users who frequently traveled in motor vehicles. The results showed that even though they used seatbelts, they still felt a strong insecurity during sudden car stops (Aomura, 2007). The motion of people with disabilities at a sudden stop should be examined more in detail.

Computer models are quite effective and widely used in order to simulate the human motions (Bertocci, 2001) (Bertocci, 2000), by many researchers. In recent years, human motion models have been used in simulations for crash experiments. For example, Bertocci (Bertocci, 1999) has used a dynamic lumped mass crash simulator to develop a model of a restrained occupant subjected to a 20g/30mph frontal motor vehicle crash. And Moorcroft (Mooreroft, 1999) demonstrated that motor vehicle crashes could be simulated with the use of MADYMO, a program designed specifically for occupant safety analysis.

However, the model that can simulate the motion of people with disabilities has not been developed. The goal of this study is to make a computer model which can simulate the motion of people with disabilities supported by a seatbelt.

In this study, a rigid link model of a human body sitting in a wheelchair was designed with muscle reflection characteristics to demonstrate the motion of people with disabilities. Although normal people generally show a similar muscle reflection pattern, the muscle reflection pattern of people with disabilities depends on the extent of each disorder. Therefore, the experiment was carried out on two hemiplegia subjects and a quadpledia subject.

As a result, the motion of subject with disabilities could be simulated by using muscle parameters and this computer model was a useful tool to evaluate the motion of wheelchair users with a seatbelt.



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# 2 PHYSICAL MODEL

## 2.1 Rigid Link Model

A two dimensional rigid link model sitting in a wheelchair is shown in Fig.1. This model is a three degrees of freedom and composed of a head, a trunk and a hip. A center of gravity for the head and trunk is located in the center of each rigid body. A dynamic equation for the rigid link model is given in equation(1).

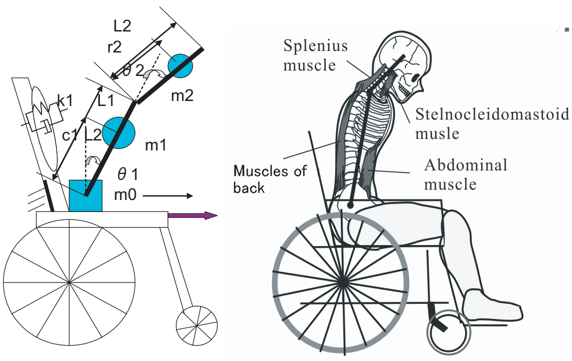


Figure 1: Schematic image of a human rigid link model fixed by a seatbelt in the wheelchair.

$$D\Big(q(t)\Big)\ddot{q}(t)+C\Big(q(t),\dot{q}(t)\Big)+S\Big(q(t),A_{x0}(t)\Big)$$
$$=M\Big(q(t),\dot{q}(t),a(t)\Big) \quad (1)$$

Input is the surface acceleration $A_{x0}$ and outputs are the horizontal displacement of lumbar spine $x$ and the angle of cervical and trunk $\theta_1$, $\theta_2$, $C$ is the term of the mass of each part and velocity of angle, $D$ is the term of the moment of inertia of each link, $S$ is the term of the tension of shoulder and waist of a seatbelt, $M$ is the torque of each joint generated by a muscle activity. $M$ is composed of an active and passive muscle activity. $a$ is the myoelectric potential.

## 2.2 The Model of a Seatbelt and the Friction Force of a Seating Surface

For this study, the seatbelt made by synthetic fibers of polyamide was used. The seatbelt is called energy absorbing webbing. The overhead view of the cross section of a human body with a seatbelt is shown in Fig.2. A human body was fixed by the seatbelt with tension $S$. The tension of the seatbelt $S$ works on the shoulder and the waist. The tension of the seatbelt is given by Nakajima's equation as follows.

$$S_i = 2S_m\left(1-\exp\left(\frac{-2x}{\varepsilon_0 L_0}\right)\right) \quad (2)$$

where $L_0$ is the length of a seatbelt, $S_m$ is the maximum tension of a seatbelt and $\varepsilon_0$ is the strain amount of a seatbelt.

The hip sinks down and moves forward when a motor vehicle stops causing friction force between hip and seating surface. The motion in the horizontal direction is affected by friction force. Fig.2 shows the frictional force between seating surface and hips. It is difficult to measure friction force by the experiment. In this study, the friction force was calculated by using the Kaminishi's (Uenishi, 2005) equation(3) of static analysis.

$$F f_0 = 0.4S_0 \quad (3)$$

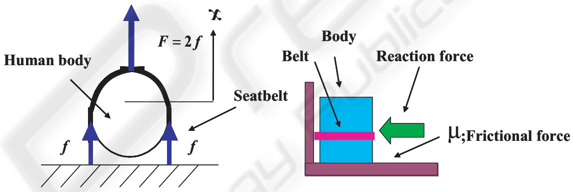


Figure 2: The overhead view of cross section diagram of a human body with a seatbelt and frictional force of seating surface.

## 2.3 The Muscle Activity Model

There are many reports about muscle activity. For example, Hase carried out a computer simulation of the human movement by using a musculo-skeletal model (Hase, 2002). In this study, the muscle reflection results were obtained by using the Hill's equation to express the motion of people with disabilities. A force of muscle contraction $f$ is described by Hill's Eq.(4).

$$f = aF(l,V) \quad (4)$$

Where $l$ is the length of muscle, $V$ is the contraction rate and $a$ is the myoelectric potential. When the length of muscle and the contractile rate is known, the muscle tension force $F_{max}$ can be calculated.

$$F_{max} = \begin{cases} \frac{(B-0.3V)}{B+V}F_a(l) & \text{if } (V>0) \\ (\frac{-1.3V}{B}+1)F_a(l) & \text{if } (\frac{-B}{3}<V<0) \\ (\frac{1.3}{3}+1)F_a(l) & \text{if } (V<\frac{-B}{3}) \end{cases} \quad (5)$$

$$F_a(l) = \begin{cases} \bar{F}[1-K_1(l_r-l)] & \text{if } (l<l_r) \\ \bar{F}[1-K_2(l-l_r)] & \text{if } (l>l_r) \end{cases} \quad (6)$$

Where $l_r$ is the initial length of each muscle, $\bar{F}$ is the maximum muscle force of isometric contraction, $B$ is

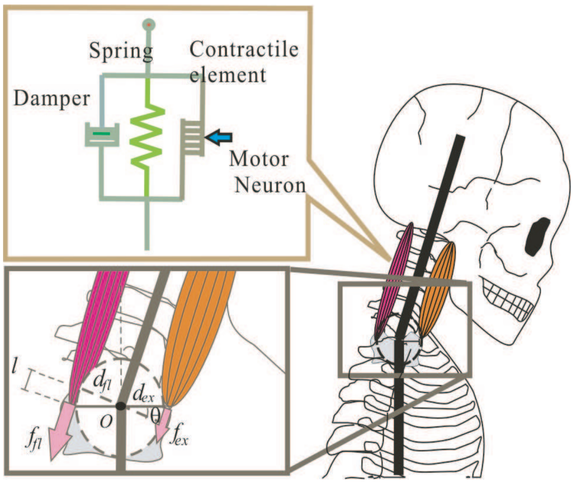


Figure 3: A muscle model of a sternocleidomastoid and a splenius.

the parameter of each muscle, $K_1$ and $K_2$ are the constant. The sternocleidomastoid muscle, the splenius muscle, the abdominal muscle and the back muscles are used to return the body to its original position. The length of extensor and flexor muscles are shown in Fig.3.

### 2.4 The Model of Myoelectric Potential

The myoelectric potential is the electrical manifestation of the neuromuscular movement associated with a contracting muscle. It is an exceedingly complicated signal which is affected by the anatomical and physiological properties of muscles (Luca, 1979).
In this study, it was assumed that the myoelectric potential would be greater when the angular displacement $\theta$, the angular velocity $\dot{\theta}$ of subjects and the acceleration $A_{x0}$ of the carriage device were stronger. The myoelectric potential is given by following equation.

$$a = 1 - \exp\left[-\left\{k\theta(t-\tau) + b\dot{\theta}(t-\tau) + cA_{x0}(t-\tau) + a_0\right\}\right] \quad (7)$$

where $k, b$ and $c$ are the human muscle parameters, $\tau$ is the time lag and $a_0$ is the steady state myoelectric potential.

### 2.5 The Passive Torque

In this study, a passive torque on ligament and soft tissue was described by Yamazaki's passive muscle model (Ogihara, 2000) and given in Eq.(8).

$$T_p = p_1 \exp p_2(\theta - p_3) - p_4 \exp p_5(p_6 - \theta) + q\dot{\theta} \quad (8)$$

where $q$ is the viscosity resistance, $p_3$,$p_6$ are the coefficients that detemines angle when elasticity resistance increases. $p_1$,$p_4$ are the coefficients that determines elastic constant when ligament and soft tissue become stiff. $p_2$,$p_5$ are the coefficients which describes nonlinearity of an elastic resistance.

## 3 CRASH EXPERIMENT

### 3.1 The Method of Experiment

In this study, two experiments were performed. The purpose of the first experiment was to determine the muscle parameters of a subject. The purpose of the second experiment was to evaluate the motion of the subjects and the model.
A carriage experiment device was used instead of a motor vehicle. The carriage experiment device is shown in Fig.4. The forward crash can simulate a sudden stop. A wheelchair (JIS T 9201) was attached to a metal structure which was harnessed to the carriage device.

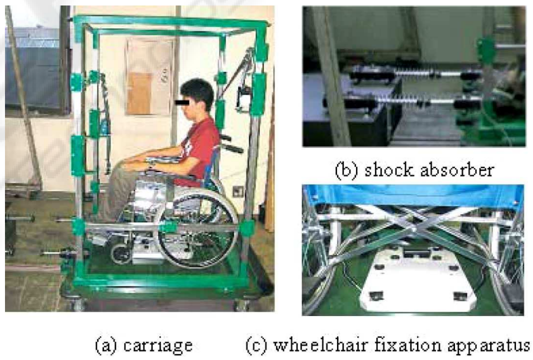


Figure 4: A wheelchair fixed on a carriage experiment device instead of a motor vehicle.

The acceleration of the carriage device was measured to compare with that of a car. Fig.5 shows the comparison of seating face acceleration between the carriage device and a car. The carriage device can be used instead of a motor vehicle because the maximum acceleration and convergence time of the carriage device matched that of a car.

A normal subject (aged 24), right and left hemiplegia subjects by bleeding in the brain (aged 55, 59) and a quadriplegia subject by a cerebral palsy (aged 29) participated in the experiment. All subjects were male. The disabled subjects were selected from many volunteers through the cooperation of a welfare organization. All subjects signed informed consent forms.

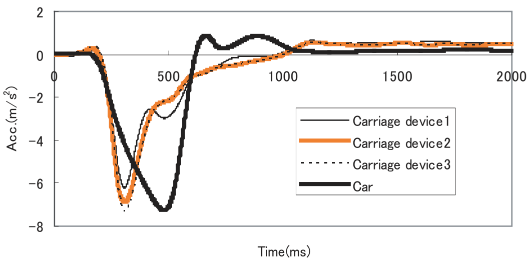


Figure 5: Comparison of seating surface acceleration of a car and a carriage device.

The sudden stop experiment was carried out and the seating surface acceleration was measured by an accelerometer. A tensimeter was attached to the seatbelt to measure the tension of the seatbelt. The trunk angle, the cervical angle were measured by a goniometer. The myoelectric potential of the sternocleidomastoid, the splenius, the abdominal and the muscle of the back were measured by a myoelectric potential sensor. Myoelectric potential sensors were attached to the sternocleidomastoid, the splenius, the abdominal and the back muscles. To measure the sternocleidomastoid muscle, a myoelectric sensor was attached between the sternoclavicularis and mastoid. To measure the splenius muscle, a myoelectric sensor was attached one centimeter away from the fifth cervical vertebra. To measure the abdominal muscle, a myoelectric sensor was attached three centimeters away from the navel. To measure the back muscle, a myoelectric sensor was attached three centimeters away from the fifth lumbar vertebra. Fig.6 shows the location to attach the myoelectric sensors and goniometers.

# 4 SIMULATION

## 4.1 Method

The muscle and physical parameters are need to be determined in the muscle reflection model. The first crash experiment was performed at a low acceleration to obtain the human muscle parameters. Both the experiment and the simulation results of the myoelectric potential were matched to determine the human muscle parameters($k, b, c, a_0$ in Eq(7)). In addition to determining the muscle parameter, another crash experiment was performed at a high acceleration in order to evaluate muscle reflection model. The human physical parameters(height, weight) were measured. The trunk and cervical angle were simulated by Runge-Kutta method which based on parameters obtained. Input was the seating surface acceleration ($A_{x0}$) and

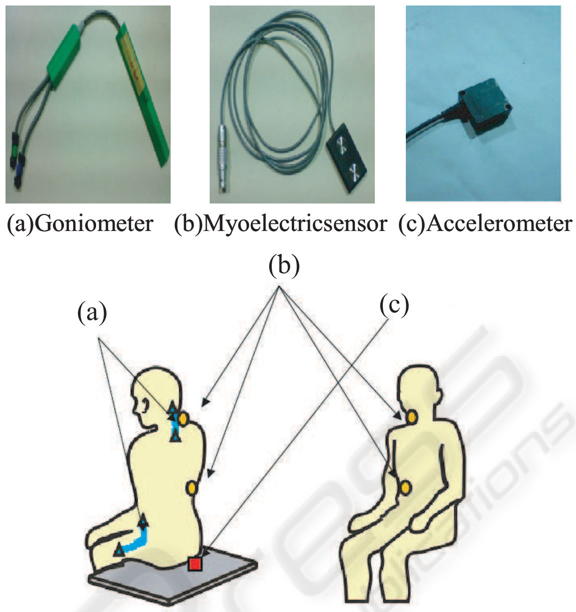


Figure 6: Location to attach the myoelectric sensors, accelerometers and goniometers.

outputs were the trunk and cervical angle and the myoelectric potential of measuring part.

## 4.2 Result

The relationship between simulation and experimental results of the trunk angle without/with seatbelt is shown in Fig.7 and Fig.8. The relationship between simulation and experimental results of the myoelectric potential of the back without/with seatbelt is shown in Fig.9 and Fig.10. Fig.7(a), (b), (c) and (d) show that the trunk angle increases around 400ms and returns to 0 degree. And, Fig.8(a'), (b'), (c') and (d') show that trunk angle doesn't incline. Therefore, the simulation results of trunk angle without/with seatbelt are similar to the experiment results in every subject. Fig.9(a), (b), (c) and (d) and Fig.10(a'), (b'), (c') and (d') show that the simulation and experimental results of myoelectric potential increase around 400ms and return to 0 degree. And the peak time of both results of myoelectric potential is matched. Therefore, the motion and the muscle reflection of disabled subjects could be simulated by the muscle reflection model without/with seatbelt. And Fig.9(c), (d) and Fig.10(c'), (d') show that the muscle reflection of right/left hemiplegia subjects without seatbelt is larger than that of using seatbelt.

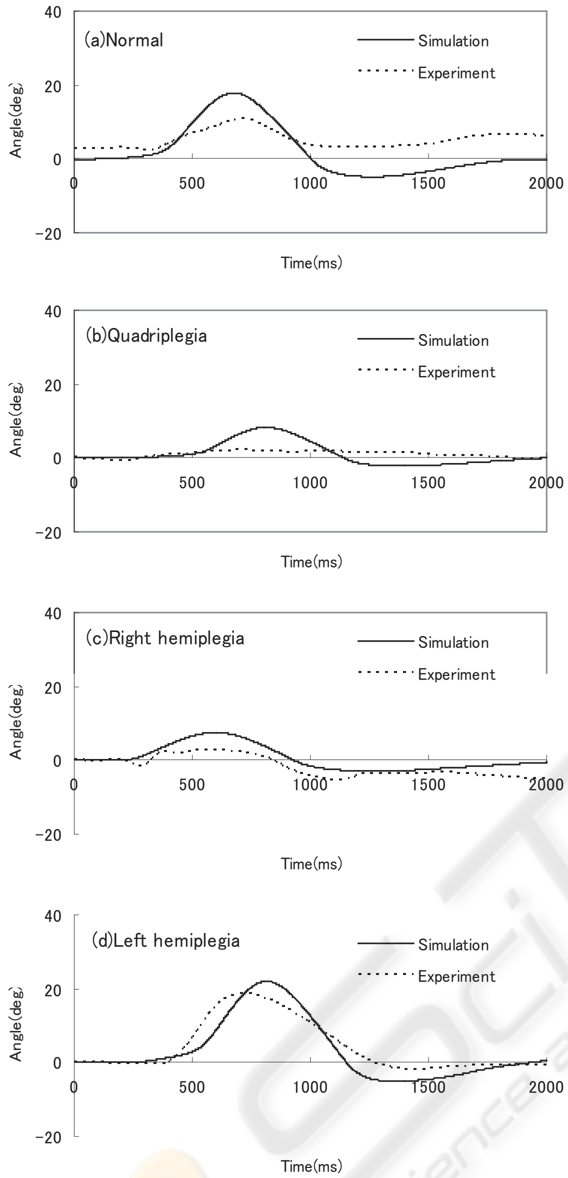


Figure 7: Comparison of the result of computer simulation with crash experiment in measuring of trunk angle without seatbelt.

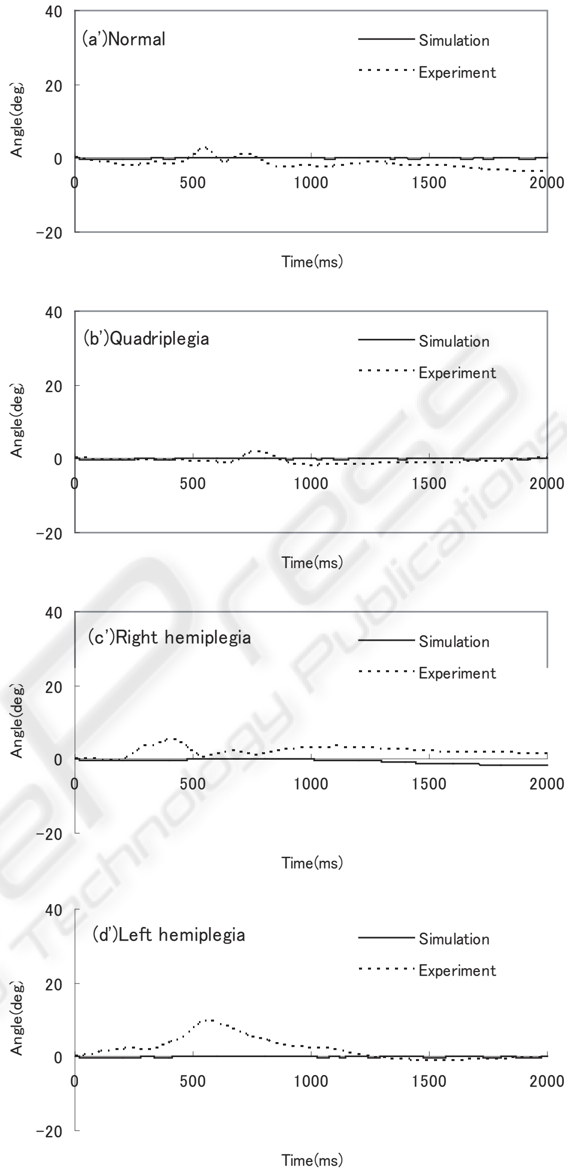


Figure 8: Comparison of the result of computer simulation with crash experiment in measuring of trunk angle with seatbelt.

## 4.3 The Calculation of the Seatbelt Tension

The tension of a seatbelt and horizontal displacement of the subjects were simulated in order to evaluate the effectiveness of a seatbelt when the car stops suddenly. The acceleration change by sudden stop of the carriage device in the experiment is shown in Fig.11(a). The comparison of the tension of the seatbelt between simulation and experiment is shown in Fig.11(b).

Although Fig.11(b) indicated that the maximum tension of the simulation matched that of the experiment, the simulation result increased sharply around 400ms. This indicated that the simulation result didn't take time lag into account until the seatbelt started to pull the human body.

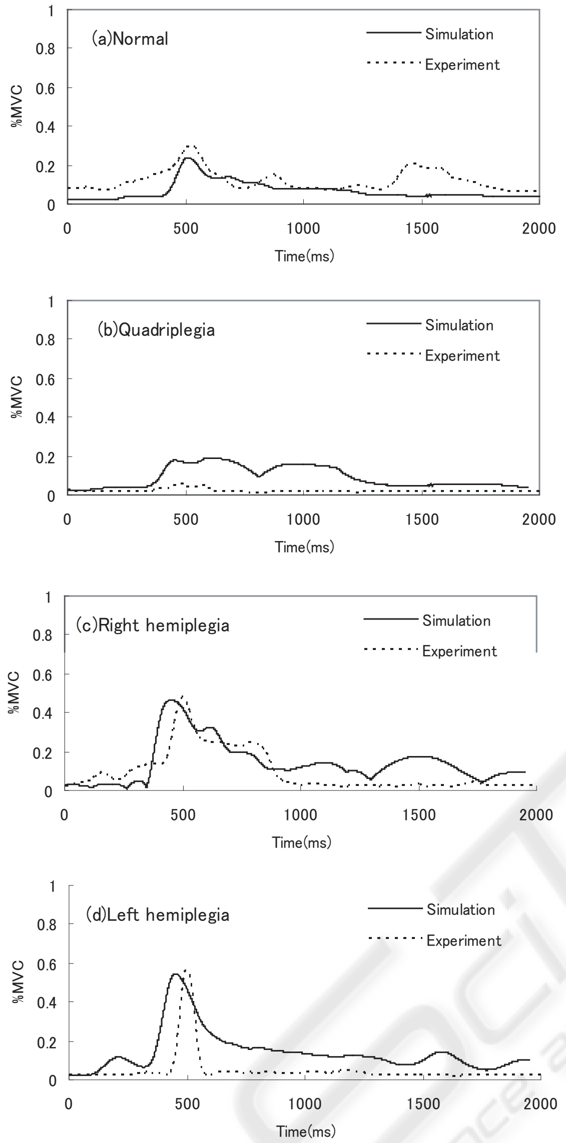


Figure 9: Comparison of the result of computer simulation with crash experiment in measuring of myoelectric potential of the back without seatbelt.

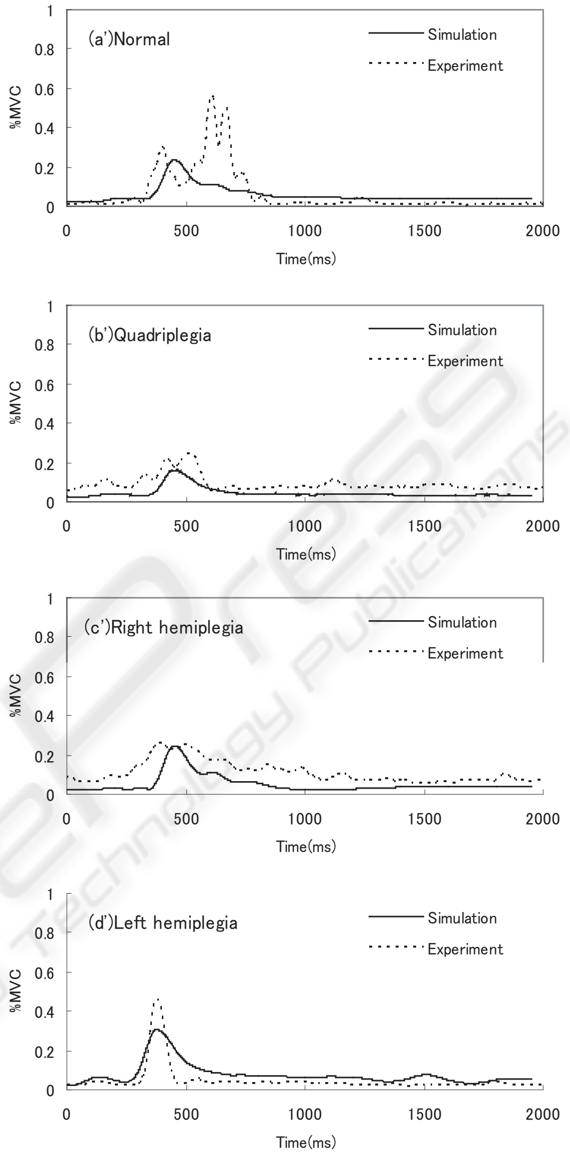


Figure 10: Comparison of the result of computer simulation with crash experiment in measuring of myoelectric potential of the back with seatbelt.

## 5 DISCUSSION

Fig.7 and Fig.8 show that the trunk angle of each subject without seatbelt increases more than that of each subject with seatbelt, because the seatbelt works effectively. Fig.10(a') shows that the normal subject uses his muscle to control his trunk angle when he is wearing a seatbelt. Fig.9(b), (c) and (d) and Fig.10(b'), (c') and (d') show that the muscle reflection of disabled subjects with seatbelt is stronger than that without seatbelt. This indicated that the trunk angle of disabled subjects is restored by the seatbelt instead of muscle reflection. When disabled subjects don't wear seatbelt, they need to use their muscles more than normal subject. Therefore using a seatbelt is more important for people with disabilities than in the case of normal people.

Next, the simulation results are focused. The motion and the muscle reflection pattern in each subject are similar. The experiment results can be simulated by the model based on muscle reflection. The muscle reflection of people with disabilities responds more strongly than that of normal people in the case without a seatbelt. Therefore, people with disabilities experi-

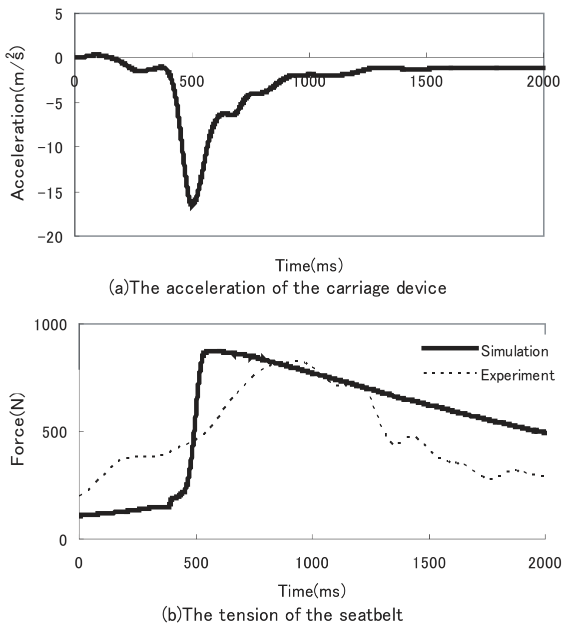


Figure 11: The comparison of the tension of a seatbelt of simulation and experiment result.

ence stronger physical burden than others do. Even though a serious traffic accident doesn't occur, a seatbelt is a useful tool for wheelchair users because it helps to control their movement instead of using their muscles while traveling in a motor vehicle.

## 6 CONCLUSIONS

In this study, a human motion model in a wheelchair traveling in a motor vehicle was developed. The model was expressed the motion of subjects with disabilities by using muscle parameters of each subject. The motion of subjects with disabilities can almost be simulated by this model. The simulation results show that the a seatbelt help to control the motion and the muscle activity of subjects with disabilities. However, the seatbelt and muscle reflection model couldn't simulate accurately. The results show that disabled people need to wear a seatbelt more than in the case of normal people.
In the future, relationship between myoelectric potential and muscle force will be investigated by fundamental experiments, improvement of myoelectric potential's equation and optimal calculation of muscle parameters in order to make safety guideline which are acceptable for people with disabilities.

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