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ULERD-based Active Training for Upper Limb Rehabilitation

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***Abstract* - In this paper, we proposed a control method to implement the upper-limb active training which is performed with the proposed exoskeleton device. It provides a wide approach for Human Machine Interface (HMI) in which the device is of high inertia, high friction and non-backdrivability and it is difficult to obtain the contact force between human and the device directly. The main idea of this method is to measure the motion of human body rather than the motion of device. This method is more suitable to the HMI in which the contact between human and device can be assumed as a spring-damper model. According to two kinds of experiments designed, different contact resistance was exerted to the forearm of the user. The sEMG signals detected from biceps brachii and triceps brachii were processed and the two kinds of resistance exerted to human forearm were confirmed.**

Index Terms –active training, Human Machine Interface, spring-damper model. rehabilitation device

I. INTRODUCTION

Stroke is a leading cause of disability in the United States, affecting an estimated 6.4 million Americans [1]. Traditional rehabilitative therapies can help regain motor function and ameliorate impairment [2]. However, they depend on the therapists' experience and need lots of therapists, which overburdens family and society. With the development of robotics, some rehabilitation robots appeared to help stroke survivors to recover motor function. Thereunto, the robots used upper limb rehabilitation mainly differentiated into exoskeletons and end-effectors. In the end-effector category, the user grasps the end-effector of the robot. The typical robot is MIT-MAUNS. It allows two degrees of freedom for movement of upper limbs including wrist, elbow and shoulder movements by performing task-oriented training [3], [4]. Other upper limb rehabilitation devices of end-effector strategy were developed based on different characters [5]-[9]. This kind of robots is simple and versatile, but can not target specific joints of limbs. The exoskeleton strategy can solve this problem obviously. One of typical device MEDARM, developed by the Canadian Institutes of Health Research (CIHR), is based on a cable driven curved track mechanism that provides independent control of all five major degrees of freedom (DOFs) at the shoulder complex [10]. ARMin [11] is an exoskeleton device with six independently actuated degrees of freedom and one coupled DoF. It can provide passive and active rehabilitation to stroke patients. It can significantly improve motor function of the paretic arm in some stroke patients, even those in a chronic state [12]. Most existing rehabilitation robots are heavy and large and not suitable for home-rehabilitation. In my research, a novel light exoskeleton device was designed and developed.

Rehabilitation robots for upper limb are typical Human-Machine Interaction (HMI) device. However, they are different from other HMI device because they should perform training strategies in clinic whatever end-effectors or exoskeletons according to Evidence Based Medicine (EBM). Previous studies have found that the neurons of some animals and humans are of plasticity [13]-[15], and motor cortex functions can be altered by individual motor experiences [16]. According to these researches, training strategies mainly include passive rehabilitation, active rehabilitation and bilateral rehabilitation [9], [17] and [18]. The rehabilitation robots mentioned above can perform one or several kinds of strategies. Rehabilitation strategy should be adapted according to individual impairment. Generally speaking, the patients following severe stroke could perform passive rehabilitation strategy and the mild stroke survivors could obtain better effect to perform active rehabilitation. Hemiparalysis patients tend to perform bilateral rehabilitation to implement recovery of motor function of the affected limb. In our previous work, we discussed the implementation of passive rehabilitation and bilateral rehabilitation using the ULERD [19]-[21] and we also did some preparation for active rehabilitation focusing on elbow joint. In this paper, we proposed a control way to implement the active rehabilitation which is suitable for the ULERD.

Active rehabilitation is a kind of strategy of physical recovery, especially in motor function recovery. During processing of this rehabilitation, patients move their impaired limbs to perform some tasks, and the robots provided a viscous resistance in the direction of the desired movement, like haptic device. Some studies have reported that repetitive practice of hand and finger movements against loads resulted in greater improvements in motor performance and functional scales than Bobath-based treatment, transcutaneous electric nerve stimulation and suprathreshold electric stimulation of hand and wrist muscles [22].

To implement this kind of rehabilitation using an electro-mechanical system, there are two fundamental control methods categorized by different inputs and outputs [23]. One is impedance control, in which motion input by the user is measured and force is fed back to the user. Alternative method, in which forces exerted by a user are measured and the device will react with the proper displacement, is called admittance control. For shortly, the paradigm of impedance control is: motion in and force out; the paradigm of admittance control is: force in and motion out. Both of them implement the same goal but adapt inverse approaches. It seems that it is difficult to understand. In fact, they were developed due to different applications. Impedance control devices require nature lightly built and highly backdrivability. Many commercial haptic devices adapt impedance control. Its performance is lacking in the region of higher forces, high mass and high stiffness and it difficult to add complex end effectors. On the other hand, admittance control devices allow considerable freedom in the mechanical design of the device, because backlash and tip inertia can be eliminated. They are able to add complex end effectors with many DoFs. In fact, impedance control was more widely used than admittance control, which can be explained the high accuracy force/torque sensor is more expensive than potion sensor and the algorithms in admittance control is more complex than impedance control. Considering the discussion about two kinds of control above, there is a problem: if the device is non-backdrivable and multi-DOFs and it is difficult to get pure and accurate force between human and device, how to obtain the programmable various resistances using the device.

We just met this problem when the effective active rehabilitation was implemented using the ULERD. Because of the application flied and design goal of the ULERD, it is not backdrivable to exert enough torque to human in passive rehabilitation mode. On the other hand, it is an exoskeleton device and it is supported by users other than wheelchair or ground for portability, so that the accurate contact force is difficult to obtain directly by using the force/torque sensor. Therefore, a new control method was proposed to generate programmable various resistances under this condition by using inertia sensors to detect the motion of user's limb. Based this method, we assumed that the resistance exerted on user's limb was proportional to the displacement between the device and human limb. In fact, the user wore the ULERD by using several elastic belts. We conducted experiments to evaluate the effect of proposed method, which is focus on the motion of elbow joint. Different weight virtual objects were required to lift up and put down. Meanwhile, the skin surface electromyogram (EMG) of were detected and analysed to assess the motion.

II. Experimental Setups

A. The upper limb exoskeleton rehabilitation device (ULERD)

The motivation of design the ULERD is to provide passive training and active training to the patients with motor dysfunction to recover the motor function of upper limb including elbow and wrist joints. Meanwhile, it was the aim to design it to be wearable and portable device. The basic design structure of the ULERD from upper view and lower view is depicted in Fig.1 and Fig.2. Three active DoFs were designed in elbow and wrist including the elbow flexion/extension, forearm pronation/supination and wrist flexion/extension. These three DoFs are both actuated and sensorised. On the other hand, four passive DoFs were added including two DoFs (one is rotation and the other is translation) in elbow joint, other two in wrist joint with considering many factors, for example, variation of flexion/extension axis (FEA), personalized otherness in physical dimension of joint and correlation between wrist and elbow joint during elbow flexion and extension. Two passive rotational DoFs are sensorised with a potentiometer. To decrease the mass of device, not only are the BLDC motors with high power density used, the main frame of device is fabricated in aluminum. The upper limb is fixed to the device using several elastic belts passing through the slotted holes were used to connect the upper limb to the device in upper arm, forearm and palm. It is easy for users to wear it by themselves.

To decrease the mass of device, BLDC motors (Maxon) are adapted to drive each active joint due to their high power density. For each of actuated DoFs, the torque is delivered from the motor to the corresponding joint by means of a reduction gearhead and steel cables (Fig.2). The motor in elbow joint was mounted perpendicularly to the axis of the upper arm considering the stability.

Because the device is to be worn, special care must be taken to ensure safety of the wearer. Safety mechanism was also designed besides safety strategy was created in software. Motion range of each joint can be limited within that of human joints by fixing the cables onto main pulleys. On the other hand, Grooved pulley is connected to the shaft sleeve by a friction facing which is pressed by a clamp nut. When transmission torque is over a certain value, the grooved pulley will deviate from shaft sleeve. This simplified clutch mechanism is difficult to be used to set an accurate threshold, but it is important as safety precaution of hardware if some errors appear in software.

Approximately with some haptic devices, the cable winds around a grooved pulley and main pulley in "8" mode in each joint of the ULERD. We improved it because this mechanism is not robust in passive training mode. The cable is fixed to the grooved pulley through passing an inner hole of grooved pulley. This improvement is also simple, but it is important to prevent deviation between the cable and grooved pulley in passive training mode.

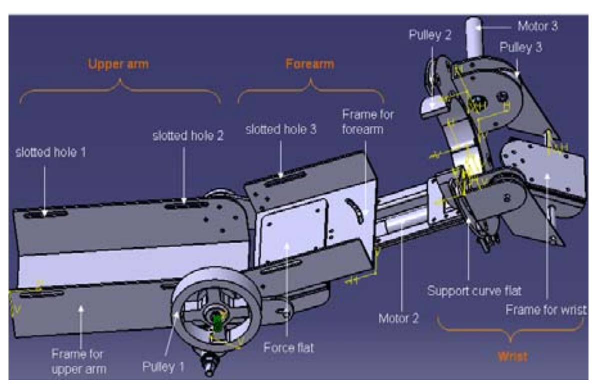


Fig.1 The upper view of the ULERD.

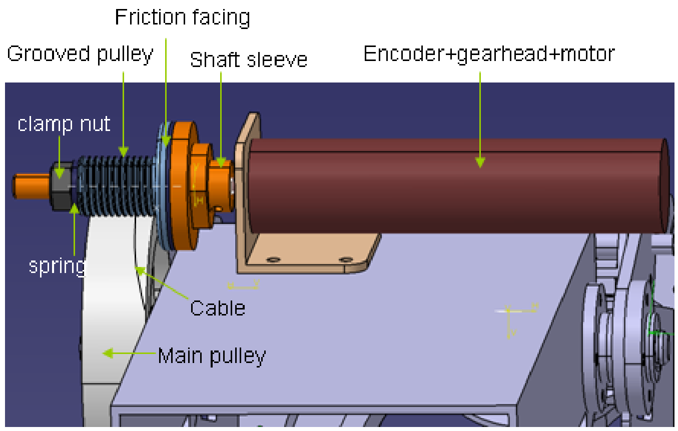


Fig.2 The transmission structure in elbow joint.

B. *MTx sensor*

The MTx is complete miniature inertial measurement unit with integrated 3D magnetometers (3D compass), with an embedded processor capable of calculating roll, pitch and yaw in real time, as well as outputting calibrated 3D linear acceleration, rate of turn and magnetic field data(Fig.3) [20].

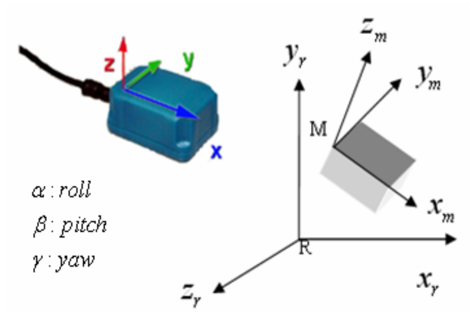


Fig. 3. MTx coordinates M relative to the reference coordinates R

III. METHODOLOGY

A. *The proposed method*

According to literature, there are two main control methods widely used in HMI systems, impedance and admittance control. The main difference between them is application condition. Impedance control requires high backdriviablity, low inertia and mass. Typical application is Phantom (Sensable Tech.). Admittance control can be used in high inertia system, but requires high accurate contact force between human and system. They have the same goal but adapt inverse approaches. In the ULERD, high gearhead ratio results in non-backdrivability, on the other hand, it covers human limb closely, which results in difficulties to get accurate contact force. Though a force sensor mounted on the forearm plate, it can detect the general contact force, and we used it to keep safe and assess the experimental results. To generate a variable resistance with the ULERD, a new method was proposed in this paper. Two control methods mentioned above are implemented by detecting information from device side, displacement or force. In proposed method, the motion even motion trend from user side is detected and the device will react with the proper displacement. The scheme of control system is show in Fig.4. The motion information of human limb can be obtained by using an inertia sensor, including Position, Velocity and Acceleration (PVA). They are calculated by the motion of virtual model and then sent to controller of motors as input. Motors are driven in close-loop with encoders. On the other hand, a force sensor mounted on the forearm frame is used to detect the force between user's forearm and device. It is only used as evaluation indicator. For measurement of biological feeling, EMG signals are analyzed during experiments to assess the effect of the proposed control method. In detail, the contact force between device and human limb can be considered as a spring mode in the ULERD. The contact force can be obtained through controlling the relative position of their ideal axes.

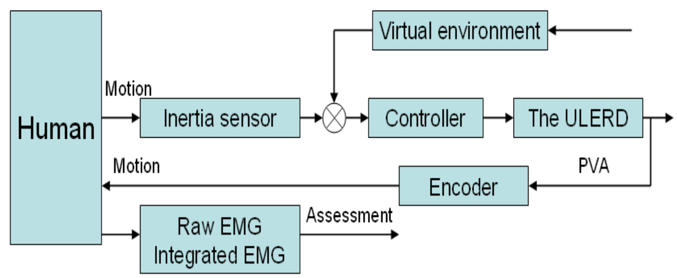


Fig.4 The control scheme of the system

Beside spring model, damp model is also widely used to generate the force exerted on human in virtual reality. The equation (1) show the various resistances related to spring and damp model. Because the motor is controlled in velocity and position close loop, gravity of system can be compensated.

$$F = k(\theta_1 - \theta) + c\dot{\theta}_1 \tag{1}$$

Where k stands for the spring coefficient, which can be set in different stiffness systems; c stands for the damp coefficient, which can be set in different virtual environment; θ_1 stands for the angle between the user's limb and horizontal plane; θ stands for the angle between the forearm frame and horizontal plane; $\dot{\theta}_1$ stands for the angle velocity of user's forearm; F stands for the resistance to user when the user moves his arm in constant velocity and $F - c\dot{\theta}_1$ is the resistance to user.

From the speaking of input and output, this method is approximate to impedance control, (e.g the motion information of user's limb is sent to system as input; output is the resistance force). Impedance control requires the device is backdrivable, and the motion can be detected by the device. The proposed control adapts sensors to detect the motion of user's limb, and control the motion of device based on created model. In fact, force can be felt by user through the deformation of muscle beside some elastic attachment. Equation (2) can be got according to equation (1).

$$\theta = \theta_1 + \frac{1}{k}(c\dot{\theta}_1 - F) \tag{2}$$

According the mechanism of elbow joint of device (Fig. 5), rotational angle of elbow joint can be obtained by Equation (3).

$$\alpha_1 = \alpha_2 + \theta \tag{3}$$

Where α_1 stands for the angle of elbow joint of device; α_1 stands for the angle of passive rotational joint which can be detected by a potentiometer.
Control equation (4) can be get according to equation (2) and (3)

$$\alpha_1 = \alpha_2 + \theta_1 + {}^{1}\!/_{k}\,(c\dot{\theta}_1 - F) \tag{4}$$

Velocity relationship shown in Equation (5) can be obtained by differentiation with respect to equation (4).

$$\dot{\alpha}_1 = \dot{\alpha}_2 + \dot{\theta}_1 + {}^{1}\!/_{k}\,(c\ddot{\theta}_1 - \dot{F}) \tag{5}$$

From this equation, if the desired resistance force is constant, rotational velocity of user's limb and forearm frame of device are possible different as long as the equation (5) is satisfied. F can be set as input reference force without concerning the damp. While $F - c\dot{\theta}_1$ is the resistance to user limb and the target of control is to find the θ (or α_1) to meet the resistance by using velocity close loop control.

B. Integration with VR enviroment

A 3-D interface was created by using OpenGL. In the virtual environment, two virtual upper limbs were created (Fig.6). One is tracked virtual arm which can move randomly within range of motion of user's limb; the other is the manipulated virtual arm which shows the motion of user's limb. The experiment requires user to manipulate the ULERD to make the manipulated virtual arm to move to follow the tracked virtual arm. During this experiment, the virtual force is programmed and a certain resistance will be exerted on the user. In this experiment, performance focusing on the elbow joint is discussed.

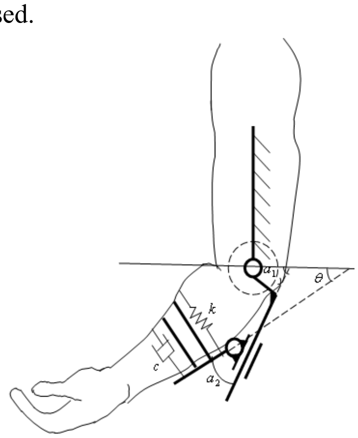


Fig.5 The mechanism scheme of elbow joint of the ULERD.

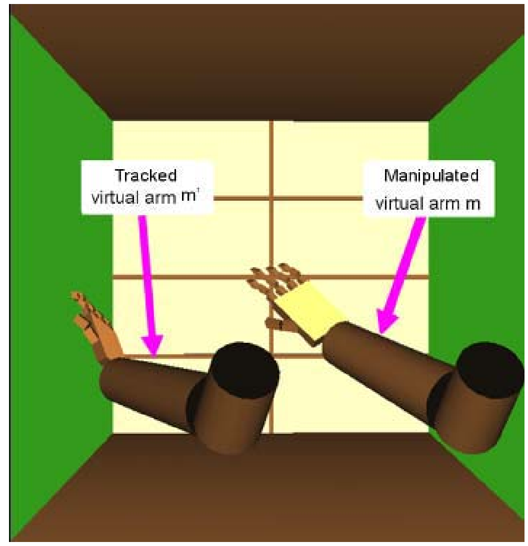


Fig.6 The virtual environment of experiment

IV. Experiments and Results

A. sEMG signals acquisition

Skin surface electromyogram (sEMG) has certain relationship with the muscle activation. To evaluate the performance of user's limb during the experiments, sEMG signal was adapted to detect the activation of relative muscles. In this paper, elbow flexion and extension was discussed; therefore, the sEMG signals from biceps brachii and triceps brachii were recoded by using a commercial sEMG acquisition and filter device (Oisaka Electronic Device Ltd. Japan.) with 8 channels (Fig.7). We used the bipolar surface electrodes with 12mm in diameter, located 18mm apart, and the sampling rate is 1000Hz. It is not obvious to distinguish the activation of muscles using the raw sEMG data. Therefore, sEMG data should be processed. Among various features extraction method, e.g. mean absolute value, average rectified value, root mean square (RMS), integrated value. We chose the integrated value. Integrated sEMG is the mathematical integral of the absolute value of the raw sEMG signals. When the absolute value of the signal is taken, noise will make the mathematical integral have a constant increase.

Fig. 7. Experimental setup of sEMG acquisition

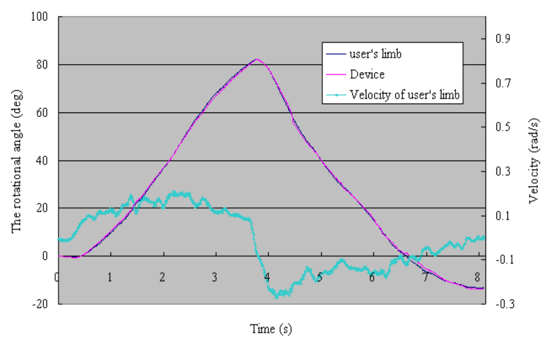
B. Experiments

In this experiments, two healthy subjects (A: 28 years old and B: 23years old) were invited to participate in the experiments. In this paper, only elbow flexion and extension motion were required. Each subject was required to perform two level experiments. One was elbow flexion and extension with no resistance and the other was elbow flexion and extension with resistance. During the performance, two kinds of sEMG signals from the elbow joint (e.g. biceps and triceps) of the user are monitored and used as motion indicator. The EMG signal is sampled at 1 kHz by using a 12-bit A/D converter. In the Experiment I, spring coefficient and damp coefficient are 0 and the device performed a tracking motion following with user's limb. In the Experiment II, spring coefficient k_1 was set as 3.5 during flexion motion and k_2 was set 1.5 during extension motion. Damp coefficient c was set 0.

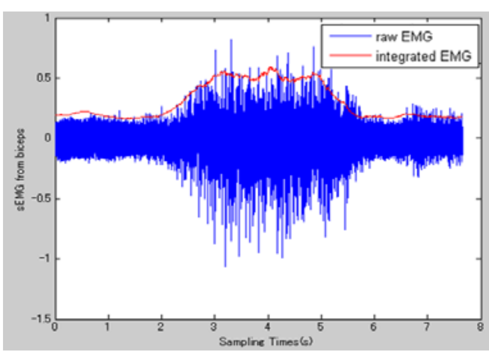
C. Experimental results

The Fig.8 shows the experimental results of the Experiment I. (a) shows the rotational angle of one user's limb and the device during elbow extension and flexion. Both of trajectories are almost the same. Blue curve shows the velocity of user's limb. (b) and (c) show the raw EMG and integrated EMG signal of biceps and triceps respectively.

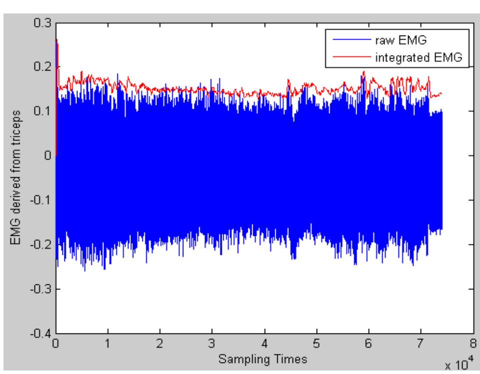
In experiment II, rotation phase of user's limb is different from that of device, so that resistance effect can be obtained. In this paper, we did not calibrate the accurate resistance value with force sensor. Figure 9 shows the typical experimental results of the Experiment II. (a) shows the rotational angle of one user's limb and the device during elbow extension and flexion. From this figure, we can know that the angle of user's limb is higher than that of the device before 4.5s and the case is reverse after 4.5s.



(a) The rotational angles of user's limb and the device and velocity of the user's limb during elbow extension and flexion

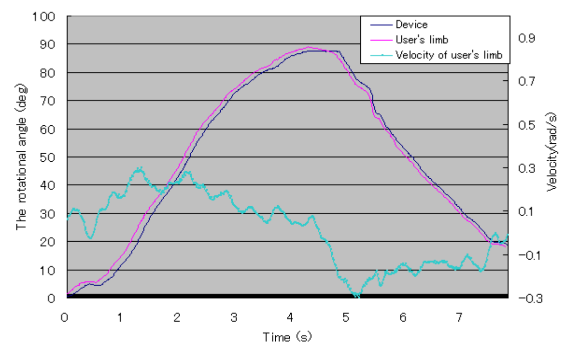


(b) The raw sEMG and integrated sEMG signals from biceps brachii

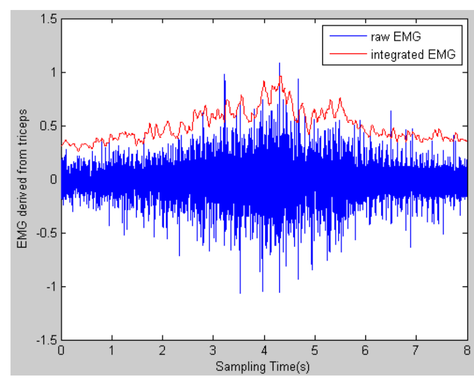


(c) The raw sEMG and integrated sEMG signals from triceps brachii

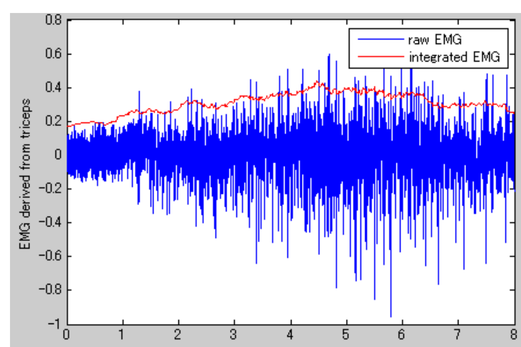
Fig.8 The experimental results in Experiment I.



(a) The rotational angles of user's limb and the device and velocity of the user's limb during elbow extension and flexion



(b) The raw sEMG and integrated sEMG signals from biceps brachii



(c) The raw sEMG and integrated sEMG signals from triceps brachii

Fig.9 The experimental results in Experiment II.

(b) and (c) show the raw EMG and integrated EMG signal of biceps and triceps respectively. The amplitude of sEMG from triceps is high in about 4 seconds and that from triceps become higher than that in Fig.8 (c) especially after 4 seconds.

V. Conclusions and Results

In this paper, we proposed a control method to implement the active rehabilitation which is performed with the ULERD. It provides a wide approach for HMI in which the device is high inertia, high friction and non-backdrivibale and it is difficult to obtain the contact force information directly. The main idea of this method is to measure the motion of human body rather than the motion of device. The desired resistance derived from the device can be obtained by using virtual force model. This method is more suitable to the HMI in which the contact between human and device can be assumed as a spring- damper model. According to two kinds of experiments designed, different contact resistance was exerted to the forearm of the user which is evaluated by processing the sEMG signals detected from biceps brachii and triceps brachii and the method is proved effective and will be used in active training in the future work.

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