IPT-EGVE Symposium (2007)
B. Fröhlich, R. Blach, and R. van Liere (Editors)

Experimental Study of Haptic Interface Considering Myoelectrical Activity

J. Sone[1], R. Inoue[1], S. Tsuyuki[1], K. Ishikawa[1], K. Yamada[1], I. Kaneko[1], Y. Tokuyama[1], K. Fujita[2], and M. Sato[3]

[1]Faculty of Engineering, Tokyo Polytechnic University, Japan
[2]Graduate School of Tokyo University of Agriculture and Technology, Japan,
[3]Precision and Intelligence Laboratory, Tokyo Institute of Technology, Japan

Abstract

Many types of haptic interface for virtual reality have been proposed. These haptic interfaces mostly contain earth grounding linkages. In this study, we considered wearable haptic interfaces for improved flexibility. It was necessary to consider information regarding the human body and we focused on the role of muscle. First, we measured the activation pattern of muscle for actions by electromyogram. For an action, we designed a wearable suit to constrict the muscle for activation. An experiment was performed using the developed wearable haptic interface for simple pulling and lifting the weight action. We confirmed the force feedback effect of haptic interfaces. Here, we also propose a new mechanism to resolve the problems identified.

Categories and Subject Descriptors (according to ACM CCS): H.5.2 [User Interfaces]: Haptic I/O

1. Introduction

Many haptic device methods [Bur96] that can assign the direction and force of humans are being studied. These devices are used for interactive communication and manipulation and to improve perceived reality in the virtual world. One such system is the earth-grounded device, including the Phantom [MS94], Arm Exoskeleton System [BAB*94], Spidar [Sat02] and Haptic Master versions. To improve haptic device portability, body-grounded devices [HOY99] have been developed. One generates the gyro moment force [YYI02], which uses an inertia roller and gimbal. Another device is the GyroCube [TSK*01], which was developed at Tsukuba University. This device has a large inertia wheel for 3 directions, and assigning the torque is generated with a motor by changing the angular momentum of the wheel. This system can present an arbitrary direction of three-dimensional spaces. In this method, devices to control the motor's electric current are needed to accurately control the speed. In the third method, the angular momentum of the wheel is changed by a brake mechanism [AOW*03]. In this system, 2 motors and 2 brake mechanisms are needed to assign one degree of freedom. Ando et al. also developed a haptic device that uses a piston-type link mechanism [AAM05], and in which the direction is assigned by difference in velocity between the increase and decrease cycles. In this method, a servo system and precise control are needed, and the torque-assigning time is short. These haptic interfaces are limited to the finger or palm.

Nishisaka et al. [NIF06] confirmed that myoelectrical signals were effective for haptic broadcast, by which the force sense is transmitted over long distance person. In this case, it is necessary to measure the myoelectrical signal in real time. The relation between muscle contraction force and electromyogram (EMG) was examined [KMKS06]. In addition, the correlation between the contraction force and average rectified value (ARV) of EMG has been discussed. The results reported by Nishisaka indicated that handling force and rectified value have a proportional relationship.

In the case of human operation, the activating signal is sent to the muscle and the muscle then contracts accordingly. The power needed for the muscle depends on the operation load. Therefore, the load of real operation is given to muscle. It acts as resistance for operation and the brain must control muscle activation during operation. This study was performed to develop a haptic interface to allow the subject to feel the actual operation. Therefore, it has a different mode of working with a power suit [Kob02, KLKS03] for assisting human operation. We have developed a wearable-type inter-

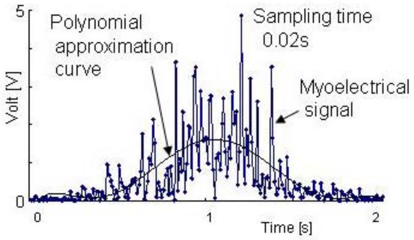

Figure 1: *Relation between myoelectrical signal and polynomial approximation curve.*

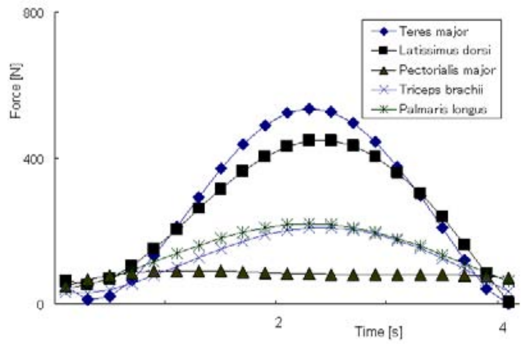

Figure 2: *Regression curves of measuring electromyogram data for teres major, latissimus dorsi, pectoralis major, triceps brachii, palmaris longus and trapezius.*

face the operation of which is affected to a greater degree by muscular power rather than external force. Therefore, it is important to consider the load of each main muscle, and it is sufficient to consider a simple simulation. The muscle load is controlled by generating a muscular activation pattern measuring the myoelectrical signals beforehand. To set the muscle load, the actuator is set at a disturbed position for muscle operation.

We examined the effectiveness of this system, and designed and built a new device to resolve the associated problems.

2. Development of the system

2.1. Design of load curve

Our system was designed based on the application of a force disturbing to that of muscle activation, which would be the load felt by the subject and the subject controls their muscles to execute the operation. It is necessary to measure the pattern of muscular activation using myoelectrical signals beforehand. Naturally, the human mechanism results in some delay by following the path: load sensing, recognition, assigning the activation command to the muscle by the brain and activation of the muscle. Mechanisms must be put in place to compensate for these delays.

The appropriate load is different according to the muscle condition. In this study, the operation involved a simple pulling action. We selected this operation of which is affected to a greater degree by muscular power rather than external force. The target is smooth operation and load curves are smoothed and multiplied by a constant value to interface various loads. The action in this study, pulling a string, involves considerable activation of pectoralis major, trapezius, latissimus dorsi, teres major, triceps brachii, biceps brachii, palmaris longus and carpi radialis longus from our investigation

2.2. Results of myoelectrical signal measurement and calculation of myoelectrical signal curve

In this study, we felt that a smooth load curve should be used for the haptic interface, as small irregularities in the wave will cause a sense of incompatibility with the interface. ARV sometimes makes use of a lowpass filter, and small irregularities will remain using lowpass filter. Therefore, after rectification, we used a 6th order polynomial approximation. Figure 1 shows the relation between the myoelectrical signal and the polynomial approximation curve. There were few differences in the approximation curve between sampling times of 0.02 and 0.1 s from trial experiment. Therefore, a sampling time of 0.1 s was selected and load curves were generated by polynomial approximation. The trapezius muscle was measured at the same time for all measurements to determine the starting point of each curve and the starting time is the activation point of the trapezius. Measurement data were rectified, averaged and approximated to obtain the myoelectrical signal curves, which plotted voltage vs. time.

It has been reported [KMKS06] that even with the muscle supplying the same power, myoelectrical signals can differ due to differences in thickness of soft tissue, etc. Thus, the myoelectrical signal gain will differ according to the percentage body fat. Our experiment is feasible, and myoelectricity curves were measured from subjects with low body fat percentages.

The load curve of teres major, latissimus dorsi, pectoralis major, triceps brachii, palmaris longus and trapezius is shown in Figure 2. The load curve was generated from the myoelectrical signal by 5 V conversion to 8N. Here, 8N is the standard value and the maximum value of load curve was changed as a parameter for experiments in the verification. The effects of the biceps brachii and carpi radialis longus were neglected as the experimental results indicated that the loads on these muscles were small.

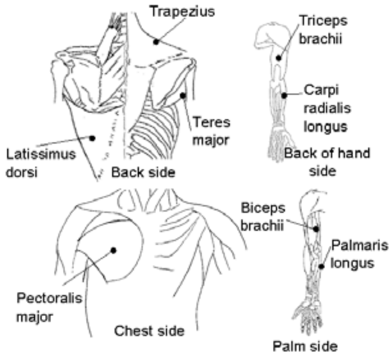

Figure 3: *Important muscles for this experiment*

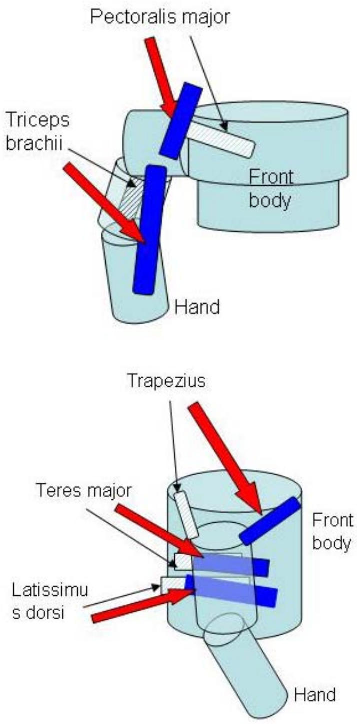

Figure 4: *Connection of artificial muscle.*

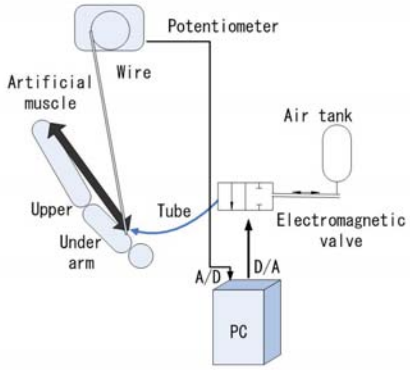

Figure 5: *System configuration.*

2.3. Development of interface

There has been recent research into power assistance suits [Kob02] to support human power in the field of welfare. In this study, a multi-point haptic interface was developed with McKibben-type [Sch61] artificial muscle using the air pressure used in the power assistance suit. In this system, artificial muscle must be placed to act as load for the corresponding muscle. Here, an ice hockey protector is used to arrange the artificial muscle. Figure 3 shows the important muscles for our experiment. Figure 4 shows the placement of artificial muscle for pectoralis major and triceps brachii, trapezius, teres major and triceps brachii. The thin and thick arrows show the actual muscle position and that of the artificial muscle, respectively. Each artificial muscle is placed close to the opposite direction of the corresponding muscle.

2.4. Development of control system

Figure 5 shows the outline of the control system. Load curves are generated by the attitude of the hand. The attitude can be determined by the position of the hand. In this method, trajectories are almost the same between the measured EMG pattern and verification experiment. The position of the hand is measured with a potentiometer via the wire (A very small amount of tension is generated with a spiral spring to prevent the wire becoming slack), and data are collected with an A/D converter as a voltage value. Then, the wire load was sufficiently small. Moreover, the load is generated from the artificial muscle with the air pressure controlled by a personal computer; it is controlled by the D/A converter with a pressure control valve. This control system was developed in C++. First, some action positions were measured because the position is changed in each subject. In addition, the time scale of the load curve is transferred to the corresponding position and the load is assigned using the curve. Here, we assumed the relation between input air pressure and generated force was proportional and load was controlled based on a force of 49N generated by input

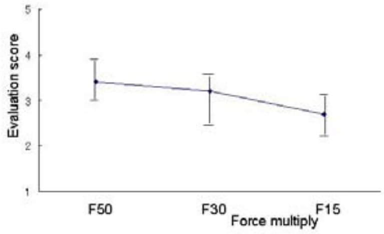

Figure 6: *Average effect of force magnification (enquete 1).*

air pressure of 0.05MPa. The data were obtained with 10% contraction of the artificial muscle. Here, the sampling time of the control was set to 10 ms.

3. Verification experiment method

The effectiveness of the developed system was verified according to the following procedures.

(1) Operation involved pulling the wire from front to rear by dominant hand. The orbit was set for the right under in the middle motion. We set the following conditions: force magnification (Amplitude of motion) of 50, 30 and 15 times; lead time, 400 ms and 1000 ms. Here, to verify the effectiveness of the load, the magnification of the load was set in the wide range. One process time of this experiment set to almost constant time (3 s). We must consider the velocity can be computed from position, and force magnification coefficient must be changed according to the velocity in the next step. The lead times compensate for the delay of artificial muscle and correct for the delay of actual muscular activity from the load. Compensation of the delay of artificial muscle is larger than that of muscular activity (<100 ms).

(2) A total of six combinations (three force magnifications and two lead times) were repeated Sixth. Each experiment was performed using random conditions and questionnaires were collected after each experiment. Five minutes rest was given to the testee at each of the 20 experiments.

Five male subjects ranging in age from 20 to 45 participated in this study, and experiments were performed with the subjects in the standing position.

The questionnaire items were as follows:

(1) Was the sense of the wearable suit's force recognisable?. (5) Perfectly recognised. (4) Almost perfectly recognised. (3) Little recognised. (2) Difficult to recognise. (1) Not recognised.

(2) How was the timing of the force?. (5) Perfectly synchronised. (4) Almost perfectly synchronised. (3) Little synchronised. (2) Poorly synchronised. (1) Not synchronised.

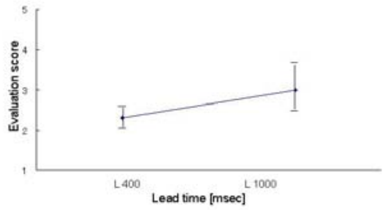

Figure 7: *Average effect of lead time (enquete 2).*

Table 1: *Probability of t-test between each force magnification factor for enquete 1.*

Enquete	F50 and F30	F50 and F15	F30 and F15
1	0.202	0.025	0.220

4. Result of verification experiment

All values in the questionnaire for each parameter were averaged and characteristic graphs were generated. F means force magnification and L means lead time. Figures 6 show the average effect of force magnification for questionnaires 1. Table 1 shows t-test result for force magnification F50 and F30, F50 and F15, F30 and F15. This Figures 6 shows that a force magnification of F50 yielded the best results. Statical analysis indicated significant difference between force multiply 50 and 15 at a probability level of 5 %. As shown in Figures 6, the gradient between F50 and F30 is smaller than that between F30 and F15. Therefore, we inferred that the increment over F50 is small. Figure 7 shows the average effect of lead time. A lead time (L) of 1000 ms yielded good results. Table 2 shows lead time of 1000 ms has significantly different from 400 ms at a probability level of 10 %.

From feasible experiment, some improvemnet issues were finded. When it is necessary to correct the load curve with the muscle condition, inference of muscle condition system must be designed using encoder data. And wealabe haptic display must be lightweight and detail and flexible control were needed.

5. Improvement of haptic interface

The results described above confirmed that operation feeling can be generated from muscle resistance. This resistance is generated by artificial muscle load using the EMG curve. Kawamoto et al. [KLKS03] reported that the position and

Table 2: *Probability of t-test between each lead time factor for enquete 2.*

Enquete	Lead time 400 and 1000
2	0.081

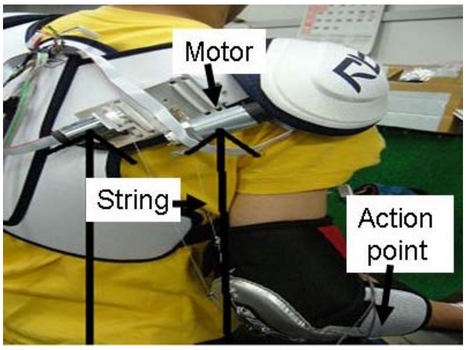

Figure 8: *Improved interface*

contraction conditions are important for the muscle activation process. Detailed load controls are needed in one action cycle. Detailed control can be achieved using a servo motor.

The design guidelines were as follows:

(1) The haptic interface must be lightweight.
(2) Detailed control can be achieved using the servo motor.

A new haptic interface was designed and developed taking the above-mentioned requirements into consideration. Figure 8 shows a developed mechanism. This system is a prototype and it can be used for the biceps brachii. The DC servo motor with a spinning wheel was installed attached to the body and the mechanism was activated with pulling a string. Here, weak pull torque is set each time to avoid slackening of the string, and the position, speed and acceleration information are detected by the encoder connected directly to the motor. The motor is controlled by the current control. Maxon RE-max 29 was used and max torque values were 26.1 mNm. The weights of the mechanisms were 0.2 kg, and these were designed to fit onto a wide range of the parts of the body. Moreover, the sense of touch is presented by pulling back with the gloves.

We have completed the basic design. In addition, we have confirmed that the designed load curve can be correctly activated by the biceps brachii based on the EMG curve comparing lifting action with actual lifting.

6. Conclusion

A wearable haptic interface using McKibben-type artificial muscle was developed, which supplies a force disturbing and equivalent to that of muscle activation. This muscle activation curve is generated by measuring myoelectrical signals beforehand. Here, we investigated operation and determined important muscles from the results of myoelectrical signal measurement and plotted appropriate load curves. Load is controlled by the position of operation. These results indicated that it was possible to recognise the sense of force by adjusting the load value. Moreover, it was confirmed that timing is important for haptic display.

Although problems remain to be resolved, the results presented here confirmed the effectiveness of our system. One of the problems to be solved involves changes in load with movement velocity. We consider that it is possible to correspond if the load curve is compressed in the time axis direction. In addition, to solve the problems associated with the delay, a compensation control system must be designed based on the experimental results. When it is necessary to correct the load curve with the muscle condition, inference of muscle condition system must be designed using encoder data. In addition, it is necessary to express the palm force using a glove. In conclusion, we have proposed a new haptic system and confirmed the effectiveness of the new interface using lifting action.

7. Acknowledgement

This research was supported by basic research (C) 17500080.

References

[AAM05] AMEMIYA T., ANDO H., MAEDA T.: Virtual force display: Direction guidance usin gasymmetric accelerationvia periodic translational motion. In *Proc. of IEEE World Haptics Conference 2005* (2005), pp. 619–622.

[AOW*03] ANDO H., OBANA K., WATANABE J., SUGIMOTO M., MAEDA T.: Development of a rotation moment-type force display using mechanical breaks. *Journal of Human Interface Society 5*, 2 (2003), 181–188.

[BAB*94] BERGAMASCO M., ALLOTTA B., BOSIO L., FERRETTI L., PERRINI G., PRISCO G. M., SALSEDO F., SARTINI G.: An arm exoskeleton system for teleoperation and virtual environment application. In *IEEE Int'l Conf. Robot. Automat.* (1994), no. 2, pp. 27–33.

[Bur96] BURDEA G. C.: *Force and touch feedback for virtual reality*. John Wiley & Sons, Inc., 1996.

[HOY99] HIROSE M., OGI T., YANO H.: Development of wearable force display(hapticgear) for immersive projection display. In *Proc. IEEE Virtual Reality 79* (1999).

[KLKS03] KAWAMOTO H., LEE S., KANBE S., SANKAI Y.: Power assist method forhal-3 using emg-based feedback controller. In *Proc. of Int'l Conf. on Systems, Man and Cybernetics (SMC2003)* (2003), pp. 648–1653.

[KMKS06] KIZUKA T., MASUDA T., KIRYU T., SADOYAMA T.: *Parctical Usage of Srface Electromyogram*. Tokyo Denki University Press, 2006.

[Kob02] KOBAYASHI H.: Development on wearable robot for human power support. In *Proc. of the 28th Annual Conference of the IEEE Industrial Electronics Society* (2002), pp. 3091–3096.

[MS94] MASSIE T., SALISBURY J. K.: The phantom haptic interface: A device for probing virtual objects. In *Proceeding of 3rd Annual Symp. Haptic Interfaces for Virtual*

Environment and Teleoperator Systems (1994), pp. 295–301.

[NIF06] NISHISAKA S., IKEDA Y., FUJITA K.: Haptic broadcast:remote haptic sharing of elastic objects allowing active pinch using electromyogram for force estimation. *Transaction of Virtual Reality Society of Japan 11*, 1 (2006), 19–26.

[Sat02] SATO M.: Development of string-based force display :spidar. In *Proc. The Eighth International Conference on Virtual Systems and Multi Media VSMM 2002* (2002), pp. 1034–1039.

[Sch61] SCHULTE H. F.: *The characteristics of the McKibben artificial muscle*. The Application of External Power in Prosthetics and Orthotics Publication 874, National Academy of Sciences - National Research Council, Washington DC, Appendix H, 1961.

[TSK*01] TANAKA Y., SAKAI M., KOHNO Y., FUKUI Y., YAMASHITA J., NAKAMURA N.: Mobile torque display and haptic characteristics of human palm. In *ICAT 2001* (2001).

[YYI02] YOSHIE M., YANO H., IWATA H.: Development of non-grounded force display using gyro moment effect. *Transaction of Virtual Reality Society of Japan 7*, 3 (2002), 329–337.