Paper:

Wrist Rehabilitation Device Using Pneumatic Parallel Manipulator Based on EMG Signal

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In this study, we focus on a rehabilitation motion of human wrist joint and aim at supporting these rehabilitation training by introducing a mechanical system. A pneumatic parallel manipulator is introduced since it can drive enough D.O.F to correspond to a wrist motion and has inherent compliance characteristics resulted from air compressibility. We propose a new training method for a muscle strengthening training based on ElectroMyoGraphy (EMG), where a payload is set between joint angle and the muscle we want to train directly. In addition, we propose a method to detect muscle fatigue based on a frequency analysis of EMG. The effectiveness of the proposed method is confirmed through some experiments.

Keywords: wrist rehabilitation, pneumatic parallel manipulator, EMG, MPF

1. Introduction

Although a 2008 Japan physical therapy white paper states that about 12,000 rehabilitation facilities and 35,000 PhysioTherapist (P.T.) are currently available in Japan [1], at least 80,000 P.T. are needed to serve all rehabilitation facilities equally. The use of robot technology is expected to cope with the shortages of nursing labors in a medical / welfare fields and some mechanical rehabilitation devices have been developed [2–6].

In this study, we focus on a rehabilitation motion of human wrist joint and aim at developing a mechanical device to support a rehabilitation training instead of P.T. A pneumatic parallel manipulator [7] is introduced from a view that it has 6 D.O.F. sufficient to correspond to complex wrist motion and has back-drivability resulted from air compressibility, which works as safe function.

In generally, wrist rehabilitation with mechanical system is implemented by giving a payload between joint torque and joint angle / joint angular velocity. We introduce a surface ElectroMyoGram (hereafter called EMG) signal instead of a joint torque into the wrist rehabilitation [8, 9]. EMG-based rehabilitation enables us to select a corresponding muscle for intensive training. EMG signals are widely used as a fatigue index, enabling rehabilitation that takes user fatigue into account.

Medical services in Japan covered by health insurance are currently limited to 180 days. Introducing a mechanical device we propose would thus enable users needing to continue rehabilitation to do so at home without being limited by health insurance conventions. We confirmed the feasibility of our proposal through experiments.

2. Wrist Rehabilitation Device

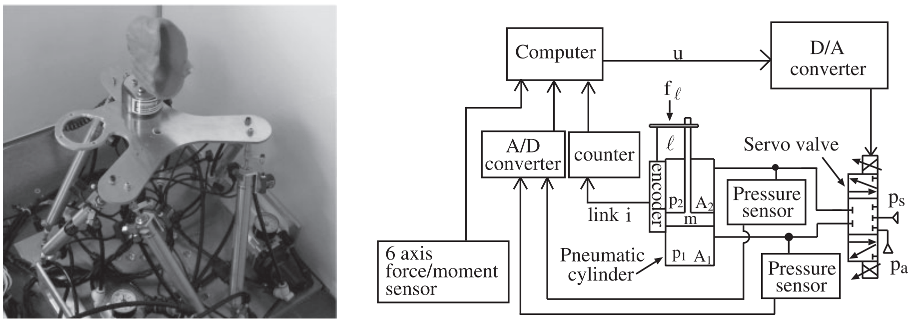
The pneumatic parallel manipulator we propose is shown in **Fig. 1(a)** [10]. Its multiple D.O.F. corresponds to wrist motion [11, 12]. Six low-friction pneumatic cylinders (Airpel Co. Ltd., 9.3 mm diameter, 150 mm rod stroke) operate as drive actuators to form Stewart type platform [13]. **Fig. 1(b)** shows the pneumatic drive circuit. Cylinder chamber pressure p_1 and p_2 are detected by a pressure sensor and piston rod displacement ℓ is measured by a wire rotary encoder with 0.025 mm resolution. Chamber pressure is regulated by a flow control servo valve (Festo MPYE-5) with supply pressure set to 500 kPa. The control algorithm is implemented on a Real-Time Application Interface RTAI, which is a real-time Linux extension with a sampling interval of 5 ms.

The position / orientation of the manipulator's upper platform shown in **Fig. 1(c)** is expressed by hand coordinate frame $\boldsymbol{h} = [x, y, z, \phi, \theta, \psi]^T$ using roll-pitch-yaw angle notation. The $\boldsymbol{h}$ origin is set above the upper platform center, the same as the wrist center. A patient put the forearm above the upper platform along the manipulator x axis and receive rehabilitation exercise by holding a jig mechanically attached with a 6-axis force / moment sensor equipped on an upper platform.

A link vector is similarly defined as $\boldsymbol{\ell} = [\ell_a, \ldots, \ell_f]^T$ with each piston rod element.

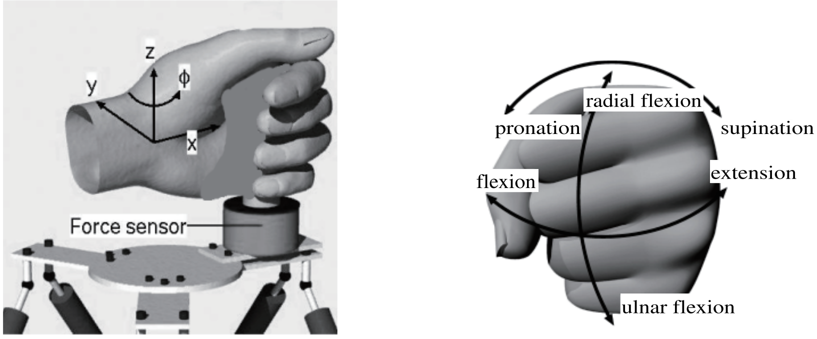
The force / moment vector at the origin of $\boldsymbol{h}$ is defined as $\boldsymbol{f_h} = [\boldsymbol{f_{he}^T} | \boldsymbol{\tau_{he}^T}]^T = [f_x, f_y, f_z, | \tau_\phi, \tau_\theta, \tau_\psi]^T$, which is obtained through coordinate transformation based on the measured force / moment with a force / moment sensor. Also, $\boldsymbol{f_h}$ acts on a piston rod as an external force $\boldsymbol{f_e}$, which satisfy the following relation from a principle of a virtual work.

$$\boldsymbol{f_h} = \boldsymbol{J}^T \boldsymbol{f_e} \quad \ldots\ldots\ldots\ldots \quad (1)$$



(a)Overview

(b)Pneumatic drive circuit



(c)Configuration

(d)Wrist motion

Fig. 1. Wrist rehabilitation device.

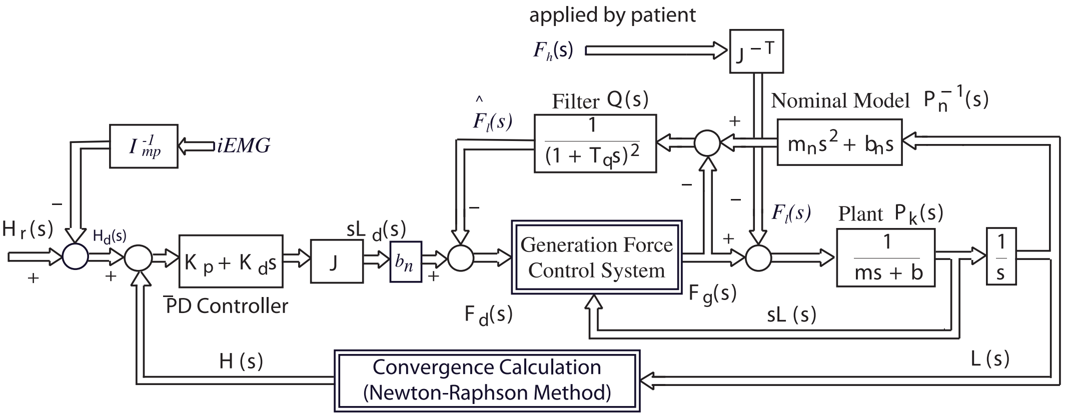


Fig. 2. EMG-based rehabilitation control.

where $\boldsymbol{J}$ is a Jacobi matrix which forms the next relation.

$$\frac{d\boldsymbol{\ell}}{dt} = \boldsymbol{J}\frac{d\boldsymbol{h}}{dt} \quad \text{. . . . . (2)}$$

Figure 1(d) shows a wrist motion, where pronation / supination, radial flexion / ulnar flexion and flexion / extension motion correspond to ψ, θ and ϕ, respectively.

3. EMG-Based Wrist Rehabilitation

In generally, rehabilitation is implemented based on the relation between an applied torque and a joint angle / angular velocity.

In this study, we introduce an EMG for a wrist rehabilitation. The EMG based rehabilitation allow us to train a muscle we want to train and evaluate a property of a muscle directly. Being able to do so is said to be effective in the rehabilitation early after a stroke.

Figure 2 shows EMG-based rehabilitation control. Position-based impedance control is used and the signal input to impedance model I_{mp} is an EMG integral as shown in Eq. (3) as the muscle force index.

$$iEMG(t) = \int_{t-T}^{t} |EMG(p)|dp \quad \text{. . . . . . (3)}$$

where integral time T is set to be 1.0 s. Control parameters are listed in **Table 1**.

Figure 3 shows EMG-based wrist rehabilitation. **Fig. 4** shows forearm muscles alignment.

Table 1. Control parameters.

T_p,T_{pn}	Time constant of pressure response
K_p,K_{pn}	Steady gain of pressure response
K_v	Steady gain between piston velocity and pressure
m,m_n	Equivalent mass for one cylinder
b,b_n	Viscous coefficient
f_h	Force / moment applied by patient
f_{pt}	Force / moment applied by P.T.
f_e	Force equivalently applied on a link
f_s	Force / moment measured by a sensor
A_1,A_2	cross sectional area of head / piston side
p_1,p_2	air pressure in head / piston side
ℓ	displacement of piston rod
J	Jacobi matrix
T_q,T_{pq}	Time constant of filter
u	control input (input voltage of valve)

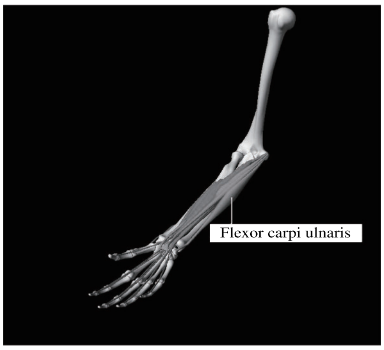
Fig. 3. Rehabilitation based on EMG.

4. Experimental Results

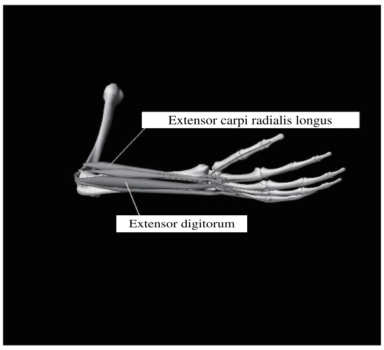
4.1. Case for Setting Stiffness

Figure 5 shows experimental results for setting stiffness ($I_{mp} = K$) between extension (ϕ) direction and the muscle (extensor carpi radialis longus), which is dominant for the extension directional motion. **Figs. 5(a)** and **(b)** correspond to the case where a stiffness K is set small (soft) and large (hard), respectively. A solid line shows the desired angle calculated by dividing iEMG with stiffness K. An actual angle agrees with its desired one, which shows the stiffness is well realized.

Another advantage of EMG-based rehabilitation is that it enables a muscle to be trained for a direction in which the muscle contributes little. **Fig. 6** shows the same experimental results as **Fig. 5**, except that the training muscle is the extensor digitorum, which contributes little to extensor motion.

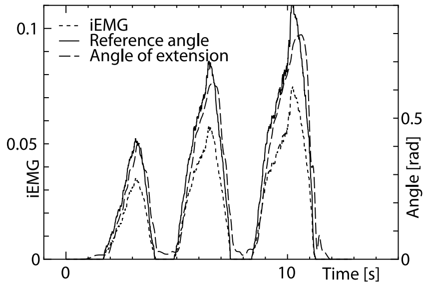


(a) Muscle for flexion

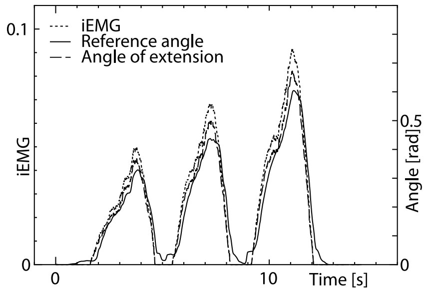


(b) Muscle for extention

Fig. 4. Flexion / Extention Muscle.



(a) High compliance



(b) Low compliance

Fig. 5. Dominant muscle rehabilitation – stiffness control.

4.2. Effectiveness in Avoiding Trick Motion

A trick motion – meaning a motion implemented by an incorrect muscle – becomes a matter of concern in a rehabilitation training. Wrist extension is, for example, implemented using the whole forearm without using the corresponding muscle. This may occur due to fatigue and is difficult to avoid in general torque-based rehabilitation.

Figure 7 shows the same experiment with **Fig. 5** except that the input of impedance model is not an iEMG but a

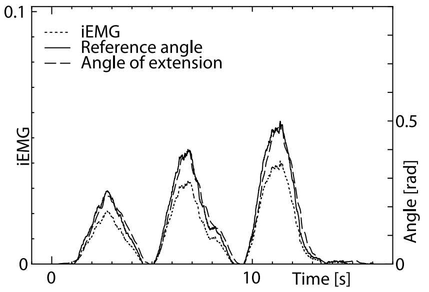
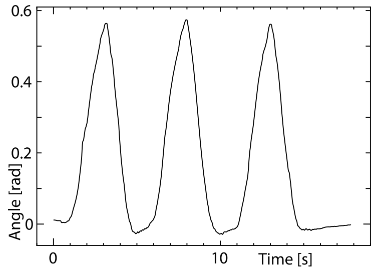
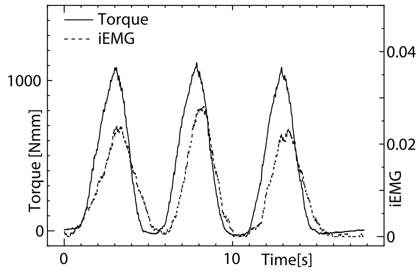


Fig. 6. Non-dominant muscle rehabilitation – stiffness control.

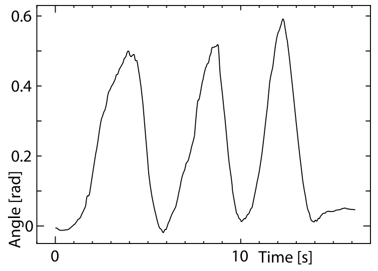


(a)Angle

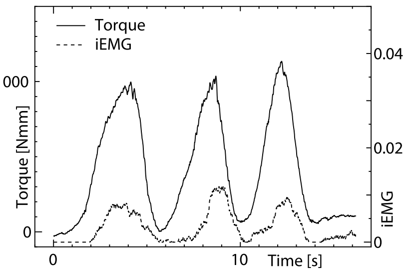


(b)Torque and iEMG

Fig. 7. Rehabilitation without trick motion (control with torque).

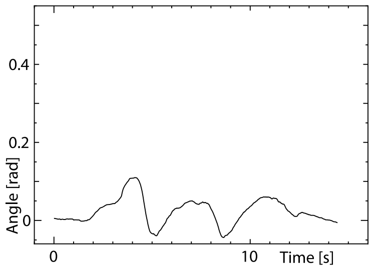


(a)Angle

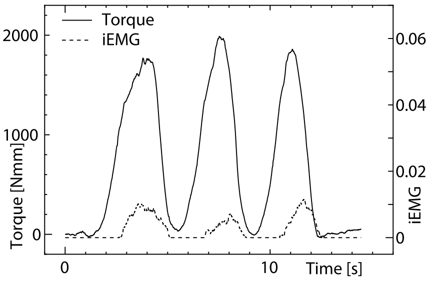


(b)Torque and iEMG

Fig. 8. Rehabilitation with trick motion (control with torque).



(a)Angle



(b)Torque and iEMG

Fig. 9. Rehabilitation with trick motion (control with EMG).

wrist joint torque. In the **Fig. 7**, we can confirm the proper iEMG since trick motion is not occurred in this case.

Figure 8 shows the trick motion, in which iEMG is low because the intended muscle is not activated even though angle and torque are high. Despite appearing to be a correct rehabilitation motion, effectiveness is poor, showing the difficulty of avoiding trick motion in torque-based control.

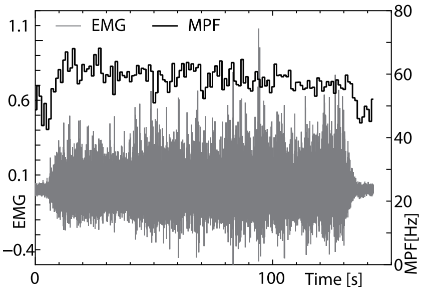
Figure 9 shows trick motion with EMG-based control as proposed in **Fig. 2**. In **Fig. 9(b)**, joint torque applied is high but the joint angle in **Fig. 9(a)** is not produced due to low iEMG because the intended muscle is not activated. Basically, EMG-based rehabilitation motion is not executed even in trick motion.

4.3. Effectiveness in Fatigue Evaluation

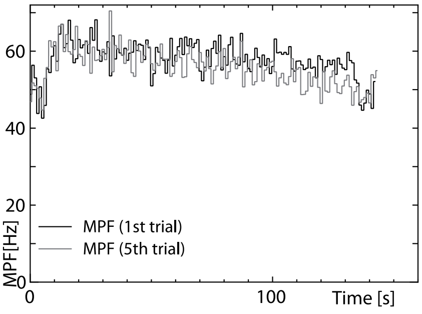
Because rehabilitation is mainly long-term, control system must take user fatigue into account. It is well known that Mean Power Frequency (MPF) of EMG shifts to low frequency with increasing muscle fatigue [14].

Figure 10(a) shows EMG MPF for the extensor digitorum muscle during continuous 2-minute contraction. The black line is EMG MPF calculated using a Fast Fourier Transform (FFT) function from the GNU Science Library (GSL), where calculation is done each 200 sampling periods (= 1.0 s). The GSL is called by a rehabilitation program written in C thanks to the real-time Linux multitasking function. The gray line is raw EMG data showing that a muscle continuously produced force for 2 minutes. Note how MPF deteriorates gradually into low frequency range over time.

This 2 minutes trial is repeated 5 times with 1 minute intervals. **Fig. 10(b)** compares with results for trials 1 and 5. The slope in trial 5 is larger than that in trial 1, meaning that fatigue increases as training progresses.



(a)EMG and MPF



(b)Variation of MPF

Fig. 10. Fatigue index with MPF.

Using our proposal, a P.T. and a patient can confirm current fatigue quantitatively without stopping rehabilitation, demonstrating the possibility of regulating rehabilitation based on a fatigue index.

5. Conclusion

The pneumatic parallel manipulator we introduce for wrist rehabilitation features multiple D.O.F. suitable for complex wrist motion and safety thanks to air compressibility.

The EMG-based rehabilitation we propose to train a muscle directly provides the following advantages:

1. A muscle is trained selectively and intensively for a variable payload.
2. A motion / muscle pair selected based on the multiple D.O.F. robotic mechanism enables a muscle to be trained even for a motion it is not dominant for.
3. Trick motion is avoided because it is not occurred in proposed EMG-based rehabilitation.
4. User fatigue is evaluated quantitatively simultaneously with rehabilitation training using real-time EMG frequency analysis.

We are now planning how to concretely evaluate our proposals at a rehabilitation facility and how to further improve EMG-based rehabilitation.

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