Active and Passive Control Algorithm for an Exoskeleton with Bowden Cable Transmission for Hand Rehabilitation

Shuang WANG, Jiting LI, Ruoyin ZHENG

Abstract—**This paper investigates the control algorithm of an exoskeleton for hand rehabilitation, which accomplishes both active and passive rehabilitation training. In the passive mode control the PID control algorithm is executed in the velocity mode of the driver. In the active mode control, control architecture is proposed to deal with in both free space and constraint space. A resistance compensation control method is proposed to reduce the resistance in free space which is caused by the friction of the Bowden cable as well as the moment of inertial. To realize the compensation, force sensors are used to measure the force exerted by the human fingertip. A commercial driver, which could switch between the two control modes by a programmable digital switch rather than changing the physical connection manually, guarantees the realization of the required functions. The experiments are conducted to verify the proposed method, and the results show that in the active control mode, the maximum finger-exerted force with compensation is about two fifths of the force without compensation which means the resistance is greatly reduced. And in the passive mode, the maximum joint position error is about 1.2 degree, which satisfies the requirement in hand rehabilitation application. The experimental results demonstrate the validity of the proposed method.**

I. INTRODUCTION

Hands are crucial and important for human being's daily life. However, they are apt to be injured in accident. Moreover, diseases, stoke for instance, can also result in the loss of hand function. In order to recover the motor capability, hand rehabilitation training is needed. Current rehabilitation trainings are usually performed manually by therapists. And this approach is costly for patients and laborious for therapists.

Recent researches showed that the method of incorporating mechatronic devices and the technology of virtual reality into hand rehabilitative training is feasible and effective [1], [2] and is attracting much research interests [3]–[6]. Dependent on different mechanical structure and rehabilitation modes, some control algorithms are investigated [7]–[11]. Despite of the researches, there are still some problems to be investigated.

For some existing exoskeleton-type hand rehabilitation devices, the Bowden cable transmission is utilized to satisfy the need of remote and changeable distance transmission and reduce the weight exerted on the patient hand. Although there reaches a consensus that the friction between the cable and sheath accounts for the great part of the mechanical resistance, the problem to compensate the friction is still open and unresolved.

Aimed at the hand rehabilitation, our research group developed a wearable exoskeleton for index finger rehabilitation [12]. In this paper we propose a control method which can realize both passive and active control modes. A position control algorithm is used for accomplishing the passive rehabilitation training and a force control for active rehabilitation. Meanwhile a resistance compensation method is investigated to compensate the mechanical resistance in free space of the active mode. A commercial driver is adopted which could switch between the two modes by programmable digital switch and don't need to change the physical connection or adjust switch manually. With the proposed approach, the measured maximum force that the index finger outputs is reduced to about two fifths which greatly improved the backdrivability of the device and the maximum joint position error is less than 1.2 degree, which satisfies the requirement in hand rehabilitation application.

The remainder of the paper is organized as follows. Section II introduces the system architecture. Section III presents control algorithm for active and passive control modes. Section IV depicts experiment and the results. Section V gives conclusion and discusses the future work.

II. SYSTEM DESCRIPTION

The hand rehabilitation system is comprised of the hand exoskeleton integrated with angle and force sensors, the control (including controller and driver), and the virtual environment. The system architecture is shown as Fig.1.

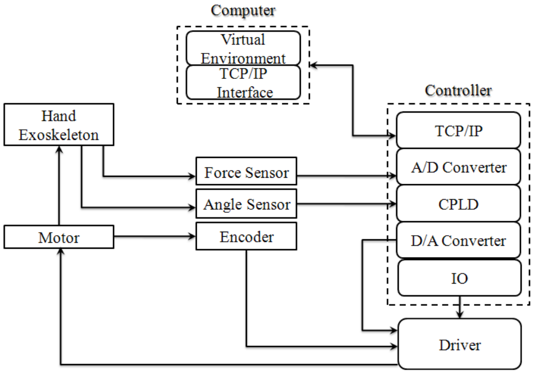

Fig.1. The system architecture

The exoskeleton device for index finger rehabilitation has 4 degrees of freedom and consists of three parts: the actuation module, the Bowden cable transmission and the exoskeleton,

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as shown in Fig 2.

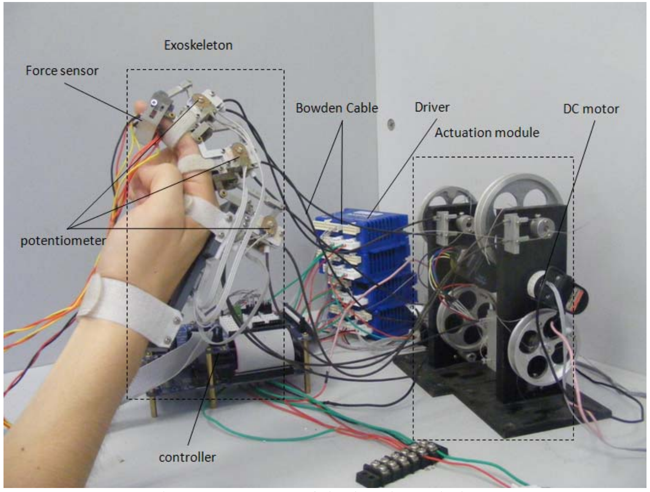

Fig. 2. Prototype of the hand exoskeleton

The exoskeleton is worn on the dorsal side of the hand. It is actuated by four actuators and can implement the motion of flexion/extension for the DIP, PIP and MCP1 joints, and the motion of adduction/abduction for the MCP2 joint. Based on the human hand anatomical structure, the exoskeleton is comprised of three parts which are attached to the phalanges – distal, middle and proximal phalanges, respectively. For each finger joint, two cables, each of which is housed in a sheath, are used to transmit force and motion from the actuator to the exoskeleton.

On the each joint shaft of the exoskeleton, a potentiometer (SV01A103 MURATA manufacturing Co., Ltd, Japan) is fixed to measure the rotational angle of the finger joint. Three force sensors (Honeywell_FSS, USA) are installed on the bottom of the distal module of the exoskeleton to measure the fingertip force exerted by the human fingertips (see Fig. 2). Because the measure part (a small sphere) of the sensor is too shallow for soft fingertip to measure the reliable force, three force sensors are assembled together and covered with a metal plate as shown in Fig. 3 to ensure that the force exerted by the finger could be measured accurately. The forces on three contact points read from the force sensors are summed to obtain the resultant force which is exerted by the finger [13].

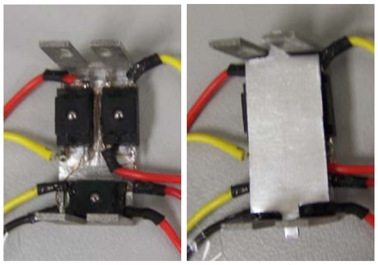
(a) (b)

Fig.3. (a) Layout of three force sensors (b) covered by a metal plate

The real-time controller (shown in Fig.2) is developed by our research group. It could sample the angle data and the force data in real time. The controller links to the host computer by the TCP/IP, and the sampling frequency is 1000Hz.

The motors are driven by four drivers (Accelnet, Copley Inc, USA). Each driver is connected to the controller through analog output channels. It could run under the torque mode, velocity mode or position mode. The different modes could be switched by a programmable digital switch. When the input voltage of the digital switch is greater than 3.65v, the velocity mode is triggered; if the input voltage is less than 1.35v, the torque mode is triggered.

The host computer runs the virtual rehabilitation and feeds the virtual interactive force back to the patient.

III. CONTROL ALGORITHM

There are various modes of rehabilitative motion. We focus on the fundamental passive and active modes. For the passive and the active rehabilitation modes, the control requiements is different. In general, a position control is suitable for the passive mode to enable imposition of specific trajectories. A force control is needed in the active mode to provide feedback force to the patient.

In this paper, the position control and force control are realized by the velocity control mode and the torque control mode of the driver, respectively

A. Active mode control

During the active rehabilitation training, the virtual hand works in two different states. When it is in contact with the virtual object, we call it is in constraint space and the virtual contact force is fed back to the human finger. Otherwise it is called in free space and the human finger is supposed to move without resistance of the device. Therefore, the mechanical resistance is demanded to be compensated in free space. For the continuous motion of abduction/adduction, the resistance of our device is not obvious. We only focus on the resistance compensation for the flexion and extension in the subsequent part.

In constraint space we use the impedance control algorithm [14] which has been widely applied, and the detail will not be described in this paper. The block diagram of control structure is shown in Fig. 4.

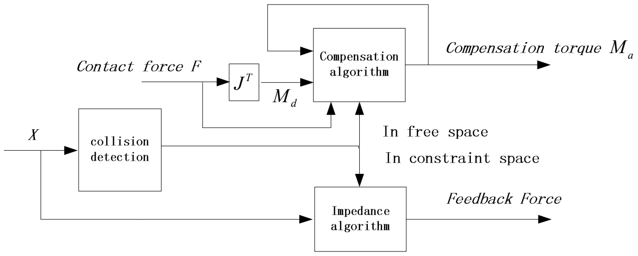

Fig.4. The block diagram of control structure

As aforementioned, in free space the exoskeleton is supposed to follow the motion of the finger and be compliant with the finger's motion. However, the friction between the cable and sheath, as well as the moment of inertial of the system, cause great resistance to the finger flexion/extension. Thus we propose a method to compensate the resistance.

For the continuous motion of flexion and extension of the finger, the direction of resistant torque is opposite, and so is the compensation torque supposed to. Therefore the motion (flexion or extension) must be identified. Because the force sensors are assembled on the side of the finger pad, the contact force only exists during the flexion theoretically. So we can use the contact force information to identify the flexion from the extension. Considering that the tiny and involuntary shake of the finger may happen and the generated contact force may provide false information for identification of motion, we use a threshold force F_1 to avoid the misjudgment. If the measured contact force F is greater than F_1, the finger is considered in the motion of flexion. Otherwise, it is in the extension. F_1 is determined by experiment.

1) Compensation during the flexion

Taking one joint for example, the actuator module, the cable/sheath transmission and the exoskeleton module are simplified as follows (Fig 5).

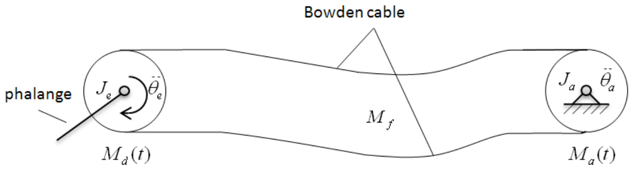

Fig.5. The simplified model for one finger joint

According to the equilibrium equation of the moment at a certain instant *t*, we have:

$$M_d(t)+M_a(t)=J_e*\ddot{\theta}_e(t)+J_a*\ddot{\theta}_a(t)+M_f(t) \tag{1}$$

where $M_d(t)$ is the driving torque which is produced by the fingertip-exerted force;

$M_a(t)$ is the output torque of the actuator module which is used to compensate the system resistance;

$M_f(t)$ is the resistance torque due to the friction between the cables and the sheaths;

J_e and J_a are the moment of inertia of the exoskeleton module and of the actuator module, respectively;

$\ddot{\theta}_e(t)$ and $\ddot{\theta}_a(t)$ are the angular acceleration of the exoskeleton module and the actuator module, respectively.

If we want to compensate the mechanical transmission caused by the friction and the moment of inertia, the actuator module output $M_a(t)$ should be:

$$M_a(t)=J_e*\ddot{\theta}_e(t)+J_a*\ddot{\theta}_a(t)+M_f(t) \tag{2}$$

Because $M_f(t)$ at instant *t* is unknown, $M_a(t)$ chould not be calculated. We estimate $M_a(t)$ with the value at instant *t*-1. Considering that the time interval is small so the estimation is reasonable. According to (1) we have：

$$J_e*\ddot{\theta}_e(t-1)+J_a*\ddot{\theta}_a(t-1)+M_f(t-1)=M_d(t-1)+M_a(t-1) \tag{3}$$

Substituting (3) in (2), we have:

$$M_a(t)=M_d(t-1)+M_a(t-1) \tag{4}$$

The active rehabilitative motion requires that it is the human hand that drives the exoskeleton. However, if the estimated $M_a(t)$ is greater than the actual resistance, the exoskeleton is then actually driven by the actuator, and the active rehabilitative motion is changed to passive rehabilitative motion which is not desired. On the other hand, the human fingertip will not contact with the sensors which means the contact force cannot be measured. To avoid the undesired problem we modify (4) as follows.

$$M_a(t)=\xi*[M_d(t-1)+M_a(t-1)],(\xi\le 1.0) \tag{5}$$

where the driving torque $M_d(t-1)$ could be calculated from the contact force measured by the force sensors. The coefficient ξ is determined by experiment.

2) Compensation during the extension

In this case there is no available contact force, so we use a constant torque to compensate the resistance.

$$M_a(t)=-M \tag{6}$$

The constant torque M is determined by experiments which just overcomes the static resistance. The negative symbol means the direction is opposite to that during the flexion.

3) Compensation in the transition state

Besides flexion and extension, the case when the finger stops moving should be considered as well since compensation is not needed. Additionally, over-compensation may possibly happen with the aforementioned method which may cause the vibration of the motor. So only with the aforementioned compensation is inadequate. We define the third state (we call transition state) to improve the method. In the transition state, the compensation torque is set to be zero.

$$M_a(t)=0 \tag{7}$$

As shown in Fig.6, another threshold F_2, together with F_1, is used to define the range of transition state, which is also determined by the experiment.

Extension state	Transition state	Flexion state
$F\le F_1$	$F_1<F\le F_2$	$F>F_2$

Fig.6. flexion, extension and the transition states

Summarily, the compensation control algorithm can be written as follows:

$$M_a(t)=\begin{cases} -M & (F \le F_1) \\ 0 & (F_1 < F < F_2) \\ \xi * [M_d(t-1)+M_a(t-1)] & (F \ge F_2) \end{cases} \qquad (8)$$

B. *Passive mode control*

In the passive motion mode, the human finger is driven by the hand exoskeleton. Thus, the task of control is to control the device to the desired positions. Considering the system security and stability, the PID controller is adopted independently for each joint. The potentiometer is used for measuring the angle of the finger joint at the exoskeleton as a position feedback in the closed loop. Thus, we have the relationship as following:

$$\Delta\theta = \theta_e - \theta_d \qquad (9)$$

$$U = K_P * \Delta\theta + K_I * \int \Delta\theta * dt + K_D * \frac{d\Delta\theta}{dt} \qquad (10)$$

where

θ_d is the desired angle of joint;

θ_e is the actual angle read from the potentiometer;

$\Delta\theta$ is the difference between θ_d and θ_e;

U is the input voltage of the driver;

K_P is the proportional coefficient;

K_I is the integral coefficient;

K_D is the differential coefficient.

The driver is executed in the velocity mode which could control the speed of the motor directly. In the interior of the driver, the relationship between the input voltage and the speed of the motor could be pre-set as follow:

$$\dot{\theta}_{motor} = K * U \qquad (11)$$

where

$\dot{\theta}_{motor}$ is the speed of the motor;

K is the constant speed parameter set in the driver.

Therefore

$$\dot{\theta}_e = \frac{\dot{\theta}_{motor}}{n} = \frac{K * U}{n} \qquad (12)$$

where n is the transmission ratio.

Substituting (10) into (12), we have:

$$\dot{\theta}_e = \frac{K}{n} * (K_P * \Delta\theta + K_I * \int \Delta\theta * dt + K_D * \frac{d\Delta\theta}{dt}) \qquad (13)$$

The block diagram of passive control mode is shown in Fig. 7.

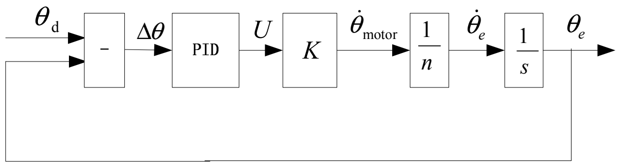

Fig.7. The block diagram of passive control mod

IV. EXPERIMENTAL RESULT

A. *Experiment on active control mode*

In the experiment, the human hand wears the exoskeleton and index finger makes the motion of flexion and extension continuously for three times with and without resistance compensation, respectively. In two conditions, the relative position of the exoskeleton module and the actuator module is approximately kept unchanged to let the experiments carried out on the same conditions. The contact forces exerted by human finger during the flexion in the two conditions are measured, respectively, as shown in Fig.8, where the input voltage of the digital switch is 0V. The coefficients for three finger joints are $\xi_{MCP} = 1.0$, $\xi_{PIP} = 0.9$, $\xi_{DIP} = 0.75$, and $F_1 = 400$ mN, $F_2 = 500$ mN . The constant compensation torques for the three finger joints are $M_{MCP} = 42$ mNm, $M_{PIP} = 24$ mNm, $M_{DIP} = 21$ mNm.

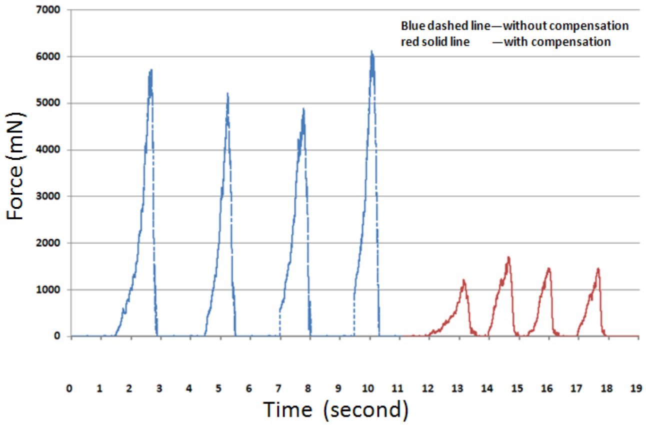

Fig.8. The measured forces exerted by the human fingertip

With the compensation control algorithm, the maximum output force of the human fingertip is less than two fifths of the maximum force without compensation during the motion of flexion. During the motion of extension, the fingertip force could not be measured, but the subject felt that it is much easier to rotate the finger with the compensation control.

B. *Experiment on passive control mode*

The target is to control the exoskeleton joint to rotate along a predefined trajectory, here a sinusoidal path with the amplitude of 80 degrees, and the period of 12 seconds. The actual joint angles are measured by the potentiometer. All the joints are controlled with the same method, so we only take the DIP joint for example. The parameters are K_P =0.5V/degree, K_I = 0.0625 V/degree, K_D = 0.007 V/degree/s, K =275rpm/V, The desired and actual trajectories and the error are shown in Fig. 9 and Fig. 10, respectively. The average error is about ±0.225 degree and the maximum error is within ±1.2 degree.

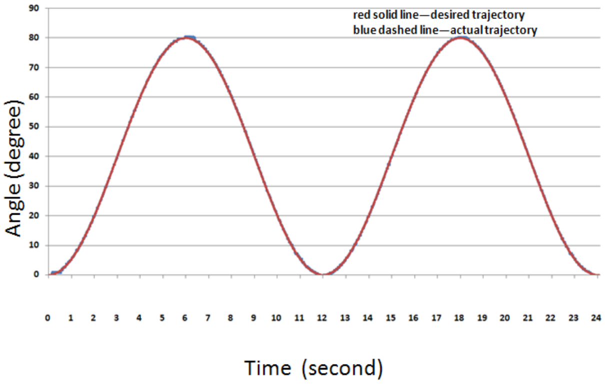

Fig.9. The trajectories of DIP finger joint

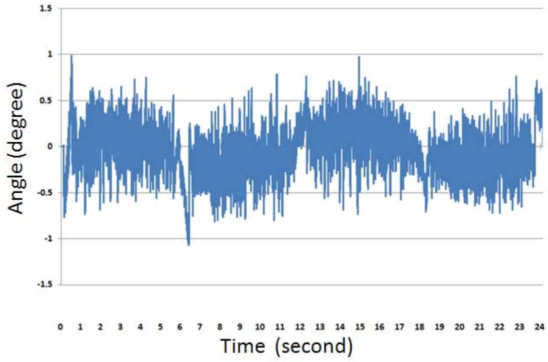

Fig.10. The position error between the expected and the actual Trajectory

V. CONCLUSION

Focused on the special requirements of the hand rehabilitation, different algorithms for active and passive control modes are realized. The two control modes, switched by a programmable digital switch rather than changing the physical connection manually, guarantees the realization of the required functions. Besides, a compensation method is investigated to compensate the bidirectional mechanical resistance, and the undesired resistance is greatly reduced.

For future work, we are going to further test the performance of the proposed control algorithm, such as the force feedback in the constraint space. We will also expand the method to the other rehabilitation modes, such as assisted and resisted modes.

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