Design of a New Hybrid Control and Knee Orthosis for Human Walking and Rehabilitation

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***Abstract*—Simultaneously considering the physical interaction between the user and the robot within safety and performance constraints in rehabilitation and human walking situations, this paper proposes a new backdrivable torsion spring actuator (BTSA) with hybrid control that switches between direct electromyography (EMG) biofeedback control and zero impedance control, to provide a novel rehabilitation training and walking assistance mechanism for humans. The proposed backdrivable 1-DOF serial elastic actuator is designed to achieve intrinsic safety, compliance properties, and control performance. The proposed mechanical system can provide desirable backdrivable property and softer stiffness than that of traditional robots. In additional, the proposed hybrid control not only considers the assistive function, when human assistance is required, but also the compliance property, when assistance is not needed. Compared to state-of-the-art assistive methods, the BTSA with the proposed hybrid control system is unique in that it can simultaneously achieve assistance control through EMG biofeedback and compliance control through zero impedance control. A simple human-robot interaction model is built to investigate performance and explain the whole control concept. Further, a knee exoskeleton is built and three kinds of controls are used on a human subject to demonstrate the difference between them. Both simulation and experimental results show that the proposed BTSA mechanism with hybrid control offers the desired properties.**

I. Introduction

In the field of physical human-robot interaction (pHRI), there is a tradeoff between human safety and robot performance and detection of human intention [1-3]. The most important issues are how to build a safe robot that can protect humans from danger and estimate human intention, and how to design an intelligent control to allow people to effectively use the robot.

A. Design of a Knee Orthotic with Intrinsic Safety

To achieve safety and efficiency, several techniques and approaches have been devised to maintain intrinsically safe robot actuation, such as SEA [4-6] and variable stiffness actuators [2, 7]. Variable stiffness actuators have more flexibility to dominate the bandwidth and payload capacity of the overall system and the safety level of the pHRI between different human users and versatile tasks [2]. However, it is usually too complicated to build a variable stiffness mechanism and the volume required is usually larger than traditional SEA mechanisms. For knee motion assistance, important features are backdrivable property, compact size, light-weight, and enough power to help human locomotion. Therefore, a simple and compact mechanism is more suitable for actual use. To increase output force, non-backdrivable systems usually use high reduction ratio transmission, i.e., worm gears [5, 6], while backdrivable systems use low reduction ratio transmission, i.e., bevel gears, cable transmission, or direct drive [4]. To achieve backdrivable property, there are many kinds of SEA designs with constant stiffness devised. They can be roughly separated into rotation [4, 6] versus linear SEA [5] and backdrivable versus non-backdrivable systems. In general, the linear SEA is composed of a linear spring and the rotatory SEA is composed of a torsion spring.

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TABLE I Comparison of rehabilitation and assistive exercise

	Rehabilitation	Assistive Exercise
Frequency	Low	High
Load	High	Medium
User	Patient with Movement Impairment or Disability	Full Abled Person, Elder, Patient with Impairment
Mechanical System	Non-Backdrivable System High-Gear Ratio	Backdrivable System Medium-Gear Ratio

The benefit of a non-backdrivable system with high gear ratio is that it can assist those with severe motion disabilities and support human limbs. On the other hand, it makes the bandwidth of most human-robot systems lower than backdrivable systems. It can also consume a great deal of human energy. Therefore, it is not suitable for those with mild motion disabilities. The comparison between rehabilitation and assistive exercise is shown in Table I.

B. Comparison with Existing Biofeedback Rehabilitation Robots

How are those rehabilitation devices controlled by using biofeedback signals? The biofeedback signals used to control or estimate the subject’s performance can be categorized as electromyography signals (EMG) [8, 9], electroencephalo -graph signals (EEG) [10, 11], and human motion and external force sensing [12, 13]. The most common biofeedback signals are motion detection and force sensing because these signals are more stable than EMG signals, and especially, EEG signals. However, motion detection and force sensing do not work well when the subjects have partially or totally impaired motor abilities. For some patients having cardiovascular accidents and spinal cord injury with a partially impaired motor ability, EMG and EEG signals provide alternate choices, but EEG signals can only be used to execute certain simple motions. EMG signals can not only be used to detect abnormal situations, such as spasticity, but also to estimate human torque directly, or provide a direct feedback signal for control.

C. Comparison with Direct EMG Biofeedback Control and Modeled-based EMG Biofeedback Control

The most common control methods using EMG signals are direct EMG biofeedback control [14] and modeled-based EMG biofeedback control [2, 3]. Direct EMG biofeedback control uses EMG signals as feedback information and EMG signals are included into the control loop directly. Modeled-based EMG biofeedback control is designed to augment human motions using estimated biology torque; therefore, it needs to estimate human torque by using EMG signals and the control loop requires other torque sensing information, such as an SEA mechanism or a torque sensor. Direct EMG biofeedback control can be used to reduce human muscle torque in some tasks. The aim of EMG biofeedback control is to control the EMG signal to equal to the command. On the other hand, the aim of modeled-based EMG biofeedback control is to amplify the estimated human torque using EMG signals and other biofeedback signals. The aim of modeled-based EMG biofeedback control is to control the actuator torque to equal to the amplified estimated human torque. Both controls seem effective in helping humans to reduce muscle exertion; however, it still has not been proven if users are comfortable when they wear the exoskeleton and perform some of the tasks. Some articles have proposed that the assist-as-needed method is more suitable for human use [15]. However, the authors of these articles do not consider how the exoskeleton system influences the user's limb dynamics.

Moreover, although direct EMG biofeedback control can be used to assist humans to do any task, its mechanism is not similar to that of human muscles. Muscles will become weaker under this control model because muscles degenerate without using. One solution is to give a specific trajectory command for a specific task [14], but this requires a great deal of data based on all kinds of tasks.

In this paper, a new backdrivable torsion spring actuator (BTSA) with hybrid control that switches between direct EMG biofeedback control and zero impedance control is proposed to provide a new rehabilitation training and walking assistance mechanism for humans and to overcome aforementioned problems. Namely, the proposed control can provide human assistance as needed and make effects of the BTSA disappear when the human subject feels assistance is not needed. The design concept and dynamic properties of the proposed system are addressed in Section II. Modeling and hybrid control of the BTSA are discussed in Section III. The simulation and experimental performance of the BTSA with hybrid control are derived and discussed in Section IV. Finally, the conclusion is presented in Section V.

II. DESIGN OF A BACKDRIVABLE TORSION SPRING ACTUATOR

A new BTSA system is constructed using a simple torsion spring, bevel gears, and an actuator. The soft stiffness of the BTSA provides mechanically intrinsic safety and measures the torque between the human and the actuator. The detailed working principle and practical design will be addressed in this section.

Figure 1 shows an exploded view of the BTSA—the actuator is serial with bevel gears and the output bevel gear is serial with a torsion spring. Two potentiometers are used. Inside, one potentiometer is inserted into the spring and used to measure the input displacement of the torsion spring. The knee angle is measured by the other potentiometer via belt transmission between the output joint and the input shaft of the potentiometer. The deflection of the spring is equal to the difference between outputs of two potentiometers, which can be used to calculate output torque via the Hook's law. Therefore, the intelligent controller can use the two sensors and EMG signals to estimate human intention. The backdrivable torsion spring actuator was constructed, as shown in Figure 2. The specifications of the BTSA are shown in Table II.

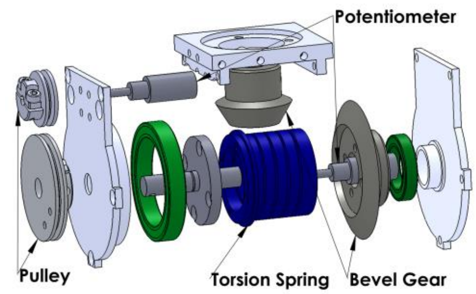

Fig. 1 Exploded view of the proposed backdrivable torsion spring actuator

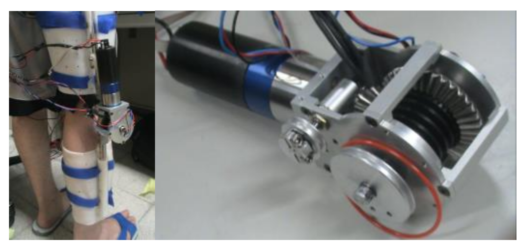

Fig. 2 Backdrivable torsion spring actuator

TABLE II SPECIFICATIONS OF THE BTSA

Weight (including the motor)	835 g
Length*Width*Height	62×50×187 mm^3
Reduction Ratio of Bevel Gear	2:1
Reduction Ratio of Motor Gear Head	43:1
Stall Torque	87.5 Nm
No-Load Speed	404 deg/sec
Spring Stiffness	40 Nm/rad

*The input motor used in this design is a Faulhaber DC-micromotor 3863H024CR with gear head 38/2 S (43:1).

III. HYBRID ZERO IMPEDANCE CONTROL AND EEG BIOFEEDBACK CONTROL

Since the BTSA approach is a general concept involving a series with a force input and a corresponding system output, the effects of the applied system should be carefully investigated. In our case, the applied system is the human knee joint. Therefore, a human-robot interaction model, proposed in [2], was used to investigate system properties and stability during physical human-robot interaction.

A. A Simple Human-Robot Interaction Model

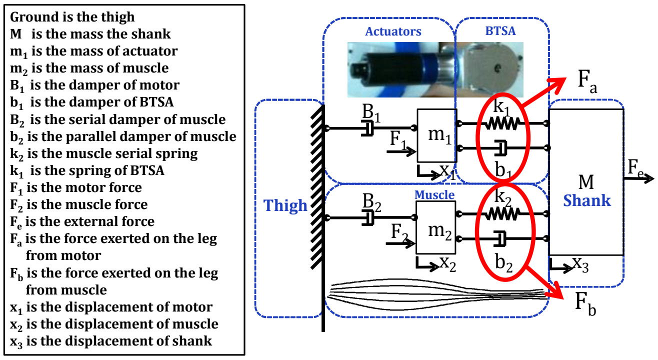

Fig. 3 Human-robot interaction model

The human-robot interaction model [2] is shown in Figure 3. The system governing equations are given as:

$$\begin{cases} m_1\ddot{x}_1 &= k_1(x_3 - x_1) + b_1(\dot{x}_3 - \dot{x}_1) - B_1\dot{x}_1 + F_1 \\ m_2\ddot{x}_2 &= k_2(x_3 - x_2) + b_2(\dot{x}_3 - \dot{x}_2) - B_2\dot{x}_2 + F_2 \\ M\ddot{x}_3 &= k_1(x_1 - x_3) + k_2(x_2 - x_3) + b_1(\dot{x}_1 - \dot{x}_3) + b \end{cases} \quad (1)$$

and the transfer functions are as follows:

$$\begin{bmatrix} \hat{X}_1(s) \\ \hat{X}_2(s) \end{bmatrix} = \begin{bmatrix} G_1(s) \\ G_2(s) \end{bmatrix} U(s) \quad (2)$$

$G_1(s)$ and $G_2(s)$ are 3×1 matrices

$\hat{X}_1(s) = X_1(s) - X_3(s)$

$\hat{X}_2(s) = X_2(s) - X_3(s)$

$U(s) = [F_1(s) \quad X_2(s) \quad F_e(s)]^T$

In the model, the inputs are motor displacement (x_1), actuator force (F_1), and external force (F_e), while the outputs are muscle spring displacement, BTSA spring displacement ($\hat{x}_1$), and knee joint displacement. muscle displacement ($\hat{x}_2$). Zero impedance control, direct EMG biofeedback control, and hybrid control of impedance control and direct EMG biofeedback control will be discussed in the following. Three control block diagrams are shown in Figure 4, Figure 5, and Figure 6, respectively.

B. Zero Impedance Control

The inputs of the zero impedance control are the displacement from the contractile element of the muscle (x_2), the actuator (F_1), and external force (F_e). The control method is to make the force (F_a) near zero or $\hat{x}_1$ near zero, as the damping term is very small. The force (F_a) exerted on the arm from the BTSA needs to achieve zero output force under simple PD control; namely, the reference input of the PD controller is zero. The block diagram is shown in Fig. 4. The control results in the user not feeling any resistance from the mechanism.

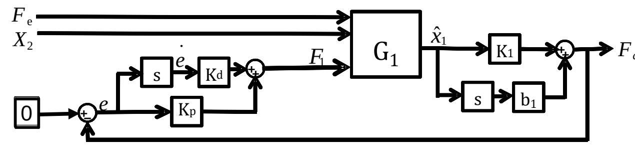

Fig. 4 Block diagram of zero impedance control

C. Direct EMG Biofeedback Control

One of the applications of the BTSA is to assist human walking rehabilitation. Therefore, direct EMG biofeedback control is designed to assist human motion and make the EMG signal track a given trajectory under PD control. Here, without losing generality, we simply define the gain between the EMG signal and the force (F_b) exerted on the leg from the muscle as 1, and thus use force (F_b) feedback signal as the EMG signal. If a more precise model is needed, a muscle mode-based method can be used to achieve nonlinear mapping [3].

The objective of direct EMG biofeedback control is that the force (F_b) needs to track some user defined trajectories. Here, the trajectory is defined as a constant value C, which will be set as zero in the following simulations. The block diagram is shown in Fig. 5.

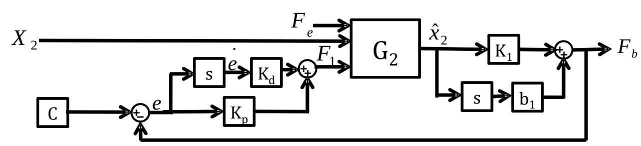

Fig. 5 Block diagram of direct EMG biofeedback control

D. Hybrid of Zero Impedance Control and Direct EMG Biofeedback Control

The concept of the proposed hybrid of zero impedance control and direct EMG biofeedback control is that the control mode is EMG biofeedback control when the user needs assistance and it is zero impedance control when the user does not need assistance. The deterministic principle is the EMG signal threshold θ.

$$\begin{cases} \text{Control Mode = direct EMG biofeedback control),if } |EMG_{sum}| \geq \theta \\ \text{Control Mode = zero impedance control)} \qquad \text{,if } |EMG_{sum}| < \theta \end{cases} \quad (3)$$

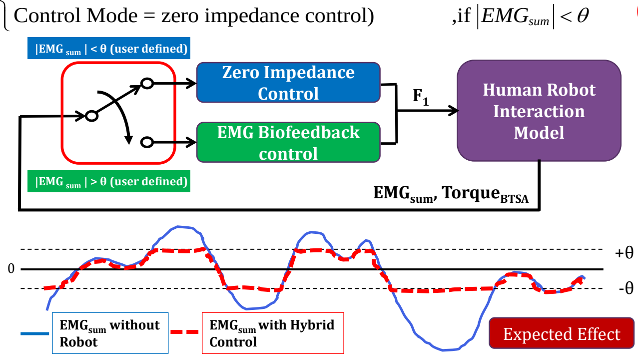

Fig. 6 Block diagram of hybrid control

EMGsum is equal to the normalized estimated EMG of the knee extensor minus the normalized estimated EMG of the knee flexor. θ is a constant threshold. If |EMGsum| is greater than or equal to the threshold, it is determined as the EMG biofeedback control mode. Conversely, if |EMGsum| is less than the threshold, it is determined as zero impedance control mode. Namely, the human moves as if the exoskeleton robot does not exist under the zero impedance control mode. Whereas, the robot assists the human with a small force to finish the task under the EMG biofeedback control mode.

E. Simulation and Experimental Setting

The simulation environment is achieved by a GUI-based simulation interface, SimMechanics from MATLAB/ Simulink. In the experiment the subject sat in a relaxed position on a chair; and then the subject was asked to extend the knee joint and then return it to the original position. The knee angle is defined as zero degrees when the thigh and shank are perpendicular, and it is defined as 90 degrees as the knee fully extends. The subject is a healthy 23-year-old man.

IV. Results and Discussions

A. *Zero Impedance Control Simulation Results*

The results of zero impedance control are shown in Fig. 7. The target of zero impedance control is to control the displacement of the torsion spring to zero. The input command of muscle displacement x_2 is a trajectory of $\sin(2\pi t)$. It is shown that the displacement of the torsion spring is approximated to zero. The use of a suitable PD controller can make the algorithm work. This also means that the subject should feel no resistance from the mechanism under the control method.

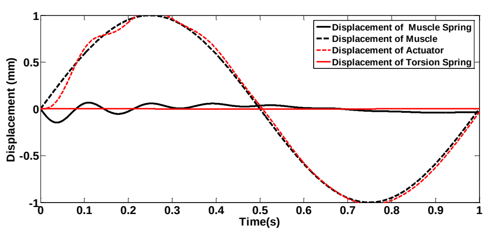

Fig. 7 Zero impedance control simulation results

B. *Direct EMG Biofeedback Control Simulation Results*

The direct EMG biofeedback control simulation results are shown in Fig. 8. The aim of direct EMG biofeedback control is to control the displacement of the muscle spring as user-defined displacement. Here, the control command of the muscle spring is set as zero and the input of the muscle displacement (x_2) is a trajectory of $\sin(2\pi t)$. It is shown that the displacement of the muscle spring is approximated to zero. The use of a suitable PD controller can cause the algorithm to work well. This represents that the subject will be able to easily move his leg without any muscle effort. However, the simulation model is a simple human-robot model and the relationship between EMG and torque is assumed as linear. The gain of (EMG/torque) in the simulation is set as 1. In reality, the relationship between EMG and torque is a non-linear relationship. The ideal result in this simulation is not quite so easy to achieve using EMG biofeedback control. Therefore, in the following experiment, the EMG signal is filtered and normalized. Then, the difference of the flexion EMG signal and the extension EMG is directly used as the feedback control signal. The dynamics are not exactly the same as this simple human-robot model, but, this model can still be used to prove and explain the control concept.

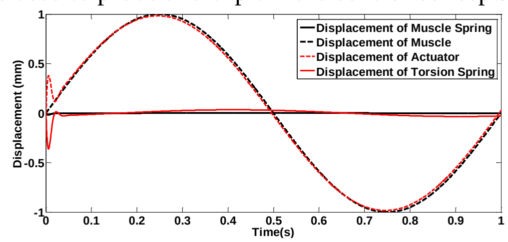

Fig. 8 Direct EMG biofeedback control simulation results

C. *Hybrid Control Simulation Results*

The results for the hybrid control simulation are shown in Fig. 9. The aims of hybrid control are to control the displacement of the muscle spring equal to a user-defined threshold when the displacement of the muscle spring is over the threshold and to control the displacement of the torsion spring equal to zero when the displacement is under the threshold.

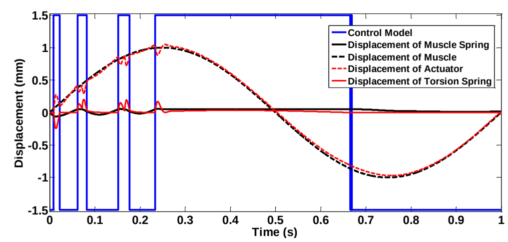

Fig. 9 Hybrid control simulation results

Here, the command for the torsion spring is set as zero mm in the zero impedance control and the command for the muscle spring is set as 0.05 mm in the direct EMG biofeedback control. The input command for muscle displacement x_2 is a trajectory of $\sin(2\pi t)$. The control model is defined as “1.5, EMG biofeedback control” and “-1.5, zero impedance control.” It is shown that the displacement of the muscle spring is always under 1.5 mm. Compared to Fig. 8, this result is more suitable for humans because the user will feel some muscle exertion, more closely approximating natural human muscle. The other benefit is that it can overcome the resistance in the dead zone of the EMG biofeedback control. The noise of the EMG signal is larger than the other torque sensor and it usually needs a wide dead zone to make the system more stable. However, in the dead zone, the user will waste energy to overcome the resistance of the mechanism. The zero impedance control can be used to solve this problem. Therefore, a hybrid of these two controllers, using two suitable PD controllers and a hybrid controller can make the subject more comfortable than using only one controller. The discontinuous displacement curve of the torsion spring is obvious. In the future, this may be solved by using certain fuzzy techniques. When using hybrid control for human assistance, the user will be able to easily feel when to use their own muscle powers and when to let the device take over.

D. *Experimental results for the BTSA with hybrid control*

In this work, a knee exoskeleton system was built, employing the proposed BTSA actuator, as shown in Fig. 2. In order to satisfy the individual needs of the elbow assistive exercise, a level arm with a shank holder was designed to move with the subject's shank, and a reference with a thigh holder was designed to allow the subject to fix the BTSA on his leg. The results of the three control methods are shown and discussed below.

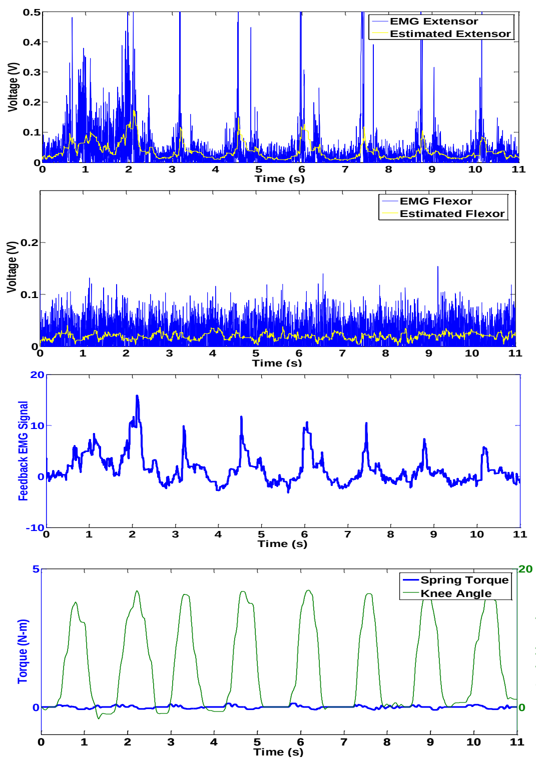

Fig. 10 Experimental results for the zero impedance control

The results for the zero impedance control are shown in Fig. 10. The first and second figures are the EMG signal and the estimated signal, respectively. The EMG signal is a rectified raw EMG signal and the estimated EMG signal is estimated by the Kalman filter proposed in [3]. The third figure is a normalized signal, calculated by a normalized extensor minus a normalized flexor. The normalizing method fixes the motor position as the user slowly extends and flexes his knee joint, then the torsion spring torque information is used and the extensor and flexor are estimated to find the coefficients for the normalized extensor and flexor. The fourth figure is the spring torque and the knee joint angle. The spring torque is measured using an inside potentiometer and the knee angle is measured using an external potentiometer. The spring torque represents the actuator torque exerted on the shank.

The spring torque is near zero and it reveals that this control method can cause the displacement of the spring to be near zero mm. Under this control, the subject needs to assist his limb and the resistance from the BTSA will disappear. Therefore, the EMG extensor and flexor are larger than the other two controls.

The results for the EMG biofeedback control are shown in Fig. 11. The normalized EMG signal is not really near zero and does not track the zero command perfectly. This is because the EMG noise is larger than the potentiometer and the EMG model is a non-stationary and non-linear signal; a linear normalization method may not be enough. But the normalized EMG signal is smaller than the other two methods and this reveals that the EMG biofeedback control is still effective in helping the user to exercise with a small EMG signal.

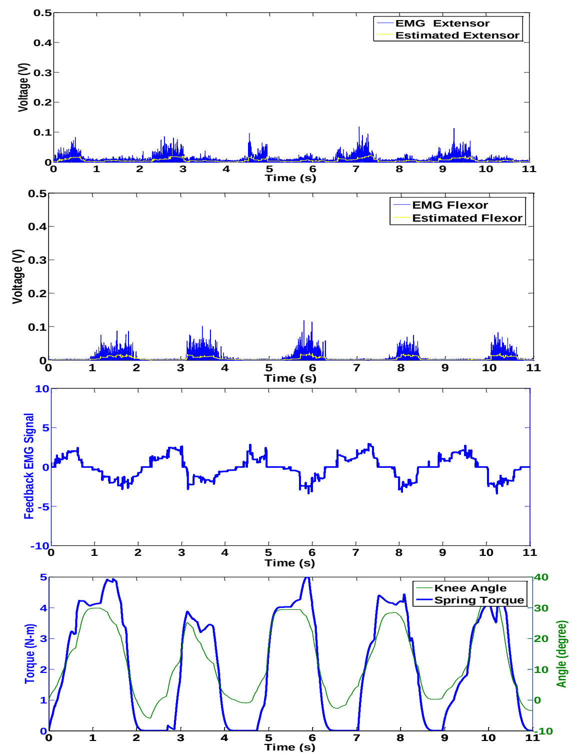

Fig. 11 Experimental results for the EMG biofeedback control

The results for the hybrid control are shown in Fig. 12. The threshold in this experiment is set at 1. It does not track perfectly, as in the EMG biofeedback results. But the hybrid control still produced the effect of helping the user to exercise with smaller torque than in the zero impedance control. Comparing the torsion torques between the hybrid control and EMG biofeedback control, the torsion torque of the hybrid control revealed that the BTSA assists the user as needed, but the torsion torque of the EMG biofeedback control revealed that the BTSA always assists the user as the user contracts his muscle. Compared to zero impedance control, the torsion torque of the hybrid control is zero only when the subject does not contract his muscle. When the feedback EMG signal is larger than the threshold, the EMG feedback control will start.

Three kinds of controls were implemented. However, the tracking errors for feedback EMG signals are small but not approximate to zero. More complex muscle models and non-linear controllers are needed to rectify the problem. Moreover, the task in this model was too simple to prove the effectiveness of this mechanism in human walking rehabilitation. In future, we need to experiment further with human walking tests, rehabilitation tests, or tests with external loads. The other issue is whether zero impedance control can be actually achieved in this study. The answer is that only approximately zero impedance was achieved. Other gravity terms, inertia terms, and BTSA damping terms need to be calculated to achieve zero impedance control.

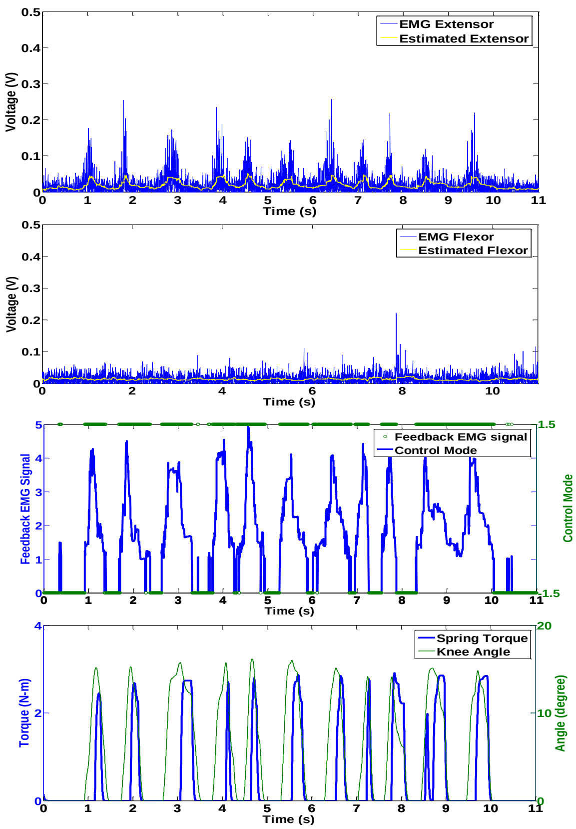

Fig. 12 Experimental results for the hybrid control

V. Conclusions

In this paper, a new BTSA with a hybrid control approach was proposed as a new method to assist humans in walking and rehabilitation. Considering mechanisms of human muscles, the proposed control can deal with possible muscles degeneration while successfully assisting human motion. Namely, this mechanism not only considers the assistance method when humans need it, but also considers control methods when assistance is not needed. The proposed system combines intrinsic safety with performance, and provides flexibility for users with different movement abilities by establishing different thresholds.

In the future, more human experiments should be conducted and stability analysis also needs to be addressed. The optimization of BTSA stiffness is also an important issue for the performance and safety tradeoff. In summary, the proposed BTSA approach with hybrid control is a good choice to help patients with weak muscle ability, the elderly, and even those with normal abilities.

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