2014 5th IEEE RAS & EMBS International Conference on
Biomedical Robotics and Biomechatronics (BioRob)
August 12-15, 2014. São Paulo, Brazil

FEXO Knee: A Rehabilitation Device for Knee Joint Combining Functional Electrical Stimulation with a Compliant Exoskeleton

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Abstract— **This paper presents the design and control of a novel assistive system, FEXO Knee, which combines functional electrical stimulation (FES) with a compliant exoskeleton for better physical rehabilitation of knee joint. The exoskeleton and FES work together in a synergetic manner that attempts to allow arbitrary torque allocation via regulating a tunable gain. The study focuses on controlling human rhythmic movements, i.e., the swing of shank, to demonstrate the assistance efficiency of the hybrid FES-exoskeleton rehabilitation. Two muscle groups (Vasti and Hamstrings) are stimulated to produce active torque for knee joint. The reference trajectories of the exoskeleton and FES are provided by central pattern generator that acts as a phase predictor to deal with unexpected phase confliction between human shank and exoskeleton. The modulated pulse width of FES stimulator is controlled by a model-based feed-forward controller. The elastic cable-driven actuator of knee exoskeleton allows safe interaction with the patients and avoids abruptly large torque shocks, which is more important than pure position tracking in robotic-assisted rehabilitation. The motion of the knee exoskeleton is controlled by a proportional-integral-derivative controller. The joint angle is the only feedback signal that needs to be measured in the control frame. The mutual torque is also measured during the swing but it is merely for the purpose of performance evaluation. Four healthy subjects participate in the initial evaluation experiments and the results show good performance of the hybrid FES-exoskeleton system.**

I. INTRODUCTION

Functional electrical stimulation (FES) has been demonstrated as an effective technique to help paraplegic patients caused by spinal cord injury (SCI) and stroke restore lower extremities mobility. Several FES systems have been developed attempting to realize gait compensation by using either surface or implanted percutaneous intramuscular electrodes in the past decades [1]. Even though the significant progress in closed-loop control and multichannel selection of muscles is achieved, it is still a complicated and tough problem to control FES in assisting paraplegic individuals to move in a natural manner due to the nonlinearity and time variability of human musculoskeletal system. Two significant drawbacks have severely hindered the widespread use of FES. The first one is muscle fatigue caused by continuous stimulating muscles and the second one is poor controllability resulting in insufficient joint torque to implement reliable limbs movement and body support [2]-[4]. To address these limitations, hybrid systems combining FES with exoskeletons have arisen as a promising approach.

Exoskeletons for physical rehabilitation have been utilized for retraining the neural system of people suffering paraplegia [5]-[8]. Exoskeletons use mechanical actuators to help patients generate gait patterns, providing functional benefits to the users. However, these devices can only provide passive assistance for users without exciting the muscles, especially for those who lose most of motion capacity after SCI or stroke. Therefore, it is a natural idea to use a powered exoskeleton as a torque compensator for FES, merging as a hybrid rehabilitation system that brings both functional and physiological benefits for patients. In previous studies, orthoses have been combined with FES as hybrid systems [9]-[10]. Even though some of these devices can manage muscle fatigue, the hybrid orthoses are still short of the capability to compensate insufficient joint torque produced by FES to generate reliable gait patterns. Besides, they do not take into account the human-machine interaction, so FES and the controlled orthoses may have phase conflict during walking. The hybrid system developed by Quintero et al. implemented a powered lower extremities exoskeleton combined with FES to enhance hip extension [3]-[4]. With the help of FES, the hip joint power of exoskeleton was reduced during periodic walking. However, the modulation of FES was based on an on-off control strategy without sufficient feedback information to regulation the contribution allocation between FES and the exoskeleton.

In this work, we introduce a novel hybrid device (FEXO Knee) combining FES with a compliant exoskeleton that is driven by a serial elastic actuator (SEA), to take full advantage of these two techniques. The reference position trajectory generator is based on a biologically-inspired control strategy, central pattern generator (CPG). CPGs are neural circuits in central nervous system of animals, which have been demonstrated to possess the functionality of regulating rhythmic locomotion without receiving sensory feedbacks or brain inputs [12]. The paper is organized as follows. Section II describes the design of knee exoskeleton, the control architecture is presented in Section III, and the results of preliminary evaluation experiment and relevant discussion are reported in Section IV. Finally, Section V discusses the results and draws the conclusions.

II. KNEE EXOSKELETON PLATFORM

This section presents the design of knee exoskeleton and attached sensors, which is an important part of FEXO Knee. It is consisted of a base rack to hold the actuator and

This work was supported by the State Key Laboratory of Robotics and System (SKLRS-2012-ZD-04).
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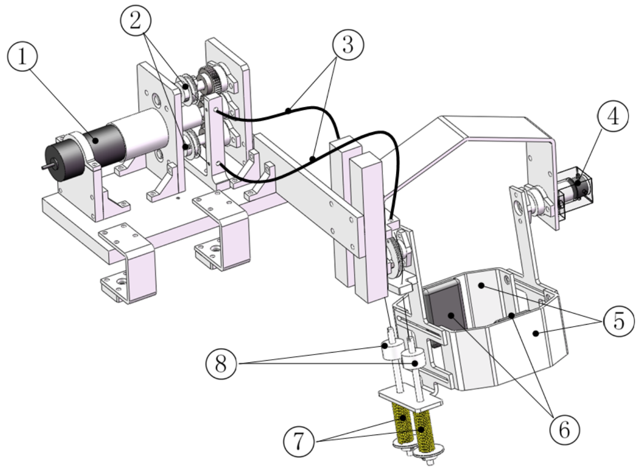

Fig. 1. The CAD model of exoskeleton in FEXO Knee displaying the main components: (1) DC servo motor, (2) driven pulleys, (3) Bowden cables, (4) encoder, (5) shank wraps, (6) interactive force sensors, (7) linear springs, (8) tension sensors.

a rotatory mechanism to realize the rhythmic swinging of human shank. The exoskeleton also contains position and force sensors for feedback control.

A. Mechanical Design and Actuation

The overall structure of knee exoskeleton is shown in Fig. 1. The material of main mechanical frame is aluminium. This prototype uses a DC servo motor with a maximum angular velocity of 7750 rpm, 273 W rated power, and a nominal torque of 0.4Nm. The planetary reducer combined with the motor has a speed ratio of 64:1. The servo motor drives two cable pulleys via three identical spur gears (the module is 1 and the number of teeth is 40). The diameter of the pulleys is 32mm. The motor actuates the exoskeleton joint by two Bowden cables rolled and fixed on the pulleys. The cables are serially connected with linear springs of 32N/mm, which turns into an SEA. Two adjustable wraps clamp and propel the human shank. The range of motion (ROM) of the joint is $\pm 80°$ for knee extension and flexion, and we consider the naturally drooping state is the zero position.

B. Sensory System

An absolute encoder is fastened coaxially with the joint to measure the rotatory angle with resolution of 0.09°. Two tension sensors combined with voltage amplifiers are set in series with the cables to measure transmission force. Two calibrated force sensing resistors (FSR 406, Interlink Electronics, USA) covered with square sponges are pasted on inner sides of shank wraps to measure the interactive force, which will be used to calculate the mutual torque via multiplying the force arm. FEXO Knee uses a data acquisition card (USB-6343, National Instrument, USA) to receive information from these sensors for real-time control, and the sampling frequency is 1 kHz.

III. CONTROL ARCHITECTURE

This section elaborates the synergetic control method we proposed for the hybrid FES-exoskeleton system. As previously discussed, The synchronization and interaction between the FES-actuated shank and the compliant knee exoskeleton are the most important issues. The control frame mainly contains three parts [see Fig. 2]: 1) the reference trajectory generator integrating the actual knee angle signal for online phase regulation, 2) the feed-forward controller for FES, and 3) the proportional-integral-derivative (PID) feedback controller for the exoskeleton. Each part will be detailed as follows.

A. CPG Model

We adopt the coupled Matsuoka oscillators as an artificial neural network to simulate the CPG [13], which generates the rhythmic patterns as desired trajectories for FES and exoskeleton respectively. The network is composed of four "neurons", and each neuron can be represented by:

$$T_r\dot{x}_i(t) + x_i(t) = -\sum_{j=1}^{4} w_{ij}y_j(t) + s_i + e_i(t) - br_i(t) \quad (1)$$

$$T_a\dot{r}_i(t) + r_i(t) = y_i(t) \quad (2)$$

$$y_i(t) = max\{0, x_i(t)\} \quad (3)$$

where T_r and T_a are time constants; $x_i(t)$ originally denotes the ith neuron's membrane potential, yet can be seen as a state variable here; w_{ij} denotes the weight of inhibitory effect from the jth neuron to the ith neuron, which specifies the inner coupling of the network; s_i denotes the tonic input of the ith neuron, set as a constant; $e_i(t)$ denotes the sensory feedback signal of the ith neuron; $r_i(t)$ denotes the neuron's adaptation rate and can also be seen as a state variable; b is a constant of the steady-state firing rate; $y_i(t)$ denotes the output of the ith neuron. The CPG module consists of a collection of nonlinear differential equations, and the modified Euler method is used in real-time numerical computation. The reference trajectories are given by:

$$\vartheta_{r,FES}(t) = \alpha(y_1(t) - y_2(t)) \quad (4)$$

$$\vartheta_{r,exo}(t) = \alpha(y_3(t) - y_4(t)) \quad (5)$$

where α denotes an amplitude constant of the trajectories. For online phase regulation, the neurons used for generating exoskeleton trajectory will not receive any feedback signals, i.e., $e_3(t) = e_4(t) = 0$, and the FES reference trajectory will coincide with the measured angle via the following learning method:

$$e_1(t) = \epsilon \cdot max\{0, \vartheta_a(t) - \vartheta_{r,FES}(t)\} \quad (6)$$

$$e_2(t) = -\epsilon \cdot min\{0, \vartheta_a(t) - \vartheta_{r,FES}(t)\} \quad (7)$$

where $\vartheta_a(t)$ denotes the measured actual angle by the encoder, and ϵ is the learning rate that determines the degree of phase and amplitude synchronization. In the initial experiments, ϵ is set at 5.0. As a result, the CPG module will act as a phase predictor that coordinates the phase of reference trajectories (FES and exoskeleton) and the actual angle through intrinsic network coupling and external sensory feedback.

B. Feedforward Controller for FES

We use an inverse muscular model to obtain the modulated stimulation pulse width as the output of FES. This inverse model is based on the Hill-type musculotendon actuator that accounts for the activation and contraction properties of muscles [16]-[17]. The model-based control method proposed by Ferrarin et al. used a piecewise linear recruitment function to describe the activation dynamics, a Gaussian function to describe the torque-angle relation, and a linear function to approximate the torque-angular velocity relation [17]. Therefore, the inverse model can be computed by

$$a(t) = \tau_{act}(t) \cdot \exp\{(\frac{\vartheta(t) + \pi/2 - \lambda_1}{\lambda_2})^2\} \cdot (1 - \lambda_3 \dot{\vartheta}(t))^{-1} \tag{8}$$

$$u(t) = \frac{a(t)(u_{sat} - u_{thres})}{u_{sf}} + u_{thres} \tag{9}$$

where $a(t)$[Nm] denotes the muscle activation; λ_1[rad], λ_2[rad] and λ_3[rad^{-1}s] are muscle and joint specific parameters used in the contraction dynamics; $u(t)$[µs] is the stimulation pulse width; u_{sat}[µs], u_{thres}[µs] and u_{sf}[µs] denote, respectively, the threshold, the saturation, and the scaling factor. Even though the relevant parameters, i.e., the contraction specific constants and the pulse width factors, are individually variable and depend on physiological measurement, we just consider them as tunable ones in our control system.

An inverse dynamics model (IDM) of the knee movement viewed as a pendulum swing is implemented to estimate the desired torque in the feed-forward controller

$$\tau(t) = I\ddot{\vartheta}(t) + B\dot{\vartheta}(t) + K\vartheta(t) + mgl_c \sin\vartheta(t) \tag{10}$$

where I[Nms^2/rad], m[kg] and l_c[m] are the segment (shank and foot) inertia, mass and equivalent length, respectively; B[Nms/rad] and K[Nm/rad] are the knee viscous damping and stiffness coefficients; $\vartheta(t)$[rad], $\dot{\vartheta}(t)$[rad/s] and $\ddot{\vartheta}(t)$[rad/s^2] denote the knee angular position, velocity and acceleration, using the desired trajectory and its first and second derivatives; $g = 9.81\text{m/s}^2$ is the gravity constant; $\tau(t)$[Nm] denotes the knee torque. The IDM for predicting torque produced by FES only consider the human knee dynamics, and the exoskeleton's inertia, mass, damping and stiffness are initially compensated by the DC servo motor.

In the feed-forward controller for FES, the angle signal in (8) and (10) is FES reference trajectory generated by the CPG module, i.e., $\vartheta_{r,FES}(t)$. The activation torque in the inverse muscular model is only a fraction of the total required torque

$$\tau_{act}(t) = (1 - \mathscr{G}) \cdot \tau_{r,FES}(t) \tag{11}$$

where $\mathscr{G}$ determines the degree of torque assistance provided by the exoskeleton. For better adaptation performance, the

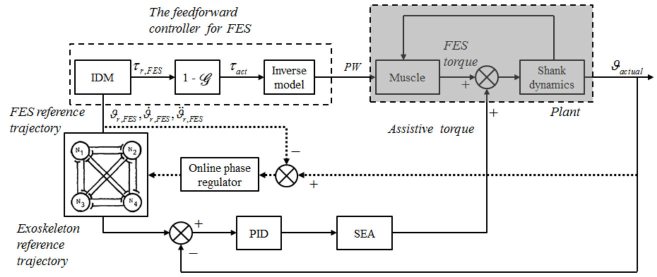

Fig. 2. Block diagram of the control architecture.

gain is not set as a constant but an online adjustable parameter, which is given by

$$\mathscr{G} = \delta \cdot (1 + K_p \parallel \vartheta_{r,exo}(t) - \vartheta_a(t) \parallel) \tag{12}$$

where δ denotes a allocation constant that represents approximately how much effort the exoskeleton should generate, and K_p is an adaptive coefficient. Equation (12) means that the assistance torque produced by the exoskeleton will be larger if the absolute angle error get greater. In the experiments, K_p is set at 0.2.

C. PID Feedback Controller

FEXO Knee controls the movement through a classical PID controller, in which the error signal is the difference between the reference trajectory of exoskeleton and the measured knee joint angle. We should point out that even though this closed loop is for position control, the absolute accuracy of angle tracking is not our primary concern, and a slight amplitude reduction is acceptable. In fact, the phase synchronization is more important. The phase delay of measured knee angle due to structural compliance and software communication latency will be solved via the aforementioned CPG method, that is why we call it a "phase predictor". The parameters k_p, k_i and k_d are tuned at 40.0, 1.0, and 0.3 by trial and error, which not only obtains good performance on compensating the exoskeleton's inertia, mass, damping and stiffness, but also provides stable torque assistance and motion pattern for the FES-actuated leg.

IV. Experimental Evaluation

A. Subjects

Four healthy male subjects participated in the preliminary evaluation experiments (age = 20.5 ± 0.6 years (mean ± s.d.); body mass = 60.3 ± 6.1 kg; height = 1.72 ± 0.07 m). The anthropometric parameters in IDM estimated for each subject were derived from [18]. The mass of the segment (shank and foot) was estimated as 6.1% of the total body weight, the length as 28.5% of the total body height, the center of mass as 60.6% of the segment length, and the radius of gyration as 73.5% of the segment length. The inertia is calculated as the product between the segment mass and the square of the segment radius of gyration, i.e., $I = m(0.735l)^2$[Nms^2/rad]. We did not consider the variations in stiffness K and the damping coefficient B during knee

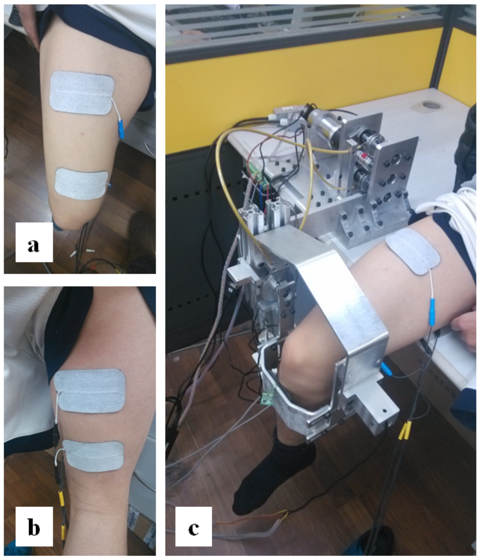

Fig. 3. Experimental setup: a) FES surface electrodes placed over the anterior thigh; b) FES surface electrodes placed over the posterior thigh; c) one subject wearing FEXO Knee.

flexion and extension movement, on account of that IDM in the control strategy is only a rough model of the actual knee dynamics. We assumed that the damping ratio η and the natural frequency ω could be tuned as constants for all subjects. The stiffness and damping coefficients were calculated by the following equations: $K = \omega^2 I - mgl/2[\text{Nm/rad}]$, $B = 2\eta\omega I[\text{Nms/rad}]$ [19]. In IDM, η and ω were respectively tuned at 0.2 and 5Hz, causing an underdamped knee motion.

B. Experimental Setup and Protocol

The FEXO Knee includes the self-developed knee exoskeleton and a commercial FES stimulator (RehaStim 2, Hasomed, Germany). The frequency of stimulation pules was set at 40 Hz. The pulse amplitude $I_S[\text{mA}]$ is a subject-specific parameter. Subjects sat on a desk and wore FEXO Knee on their right leg, two-channel surface electrodes were placed over the anterior and posterior thigh, as shown in Fig. 3. All subjects were told not to perform any voluntary movement during the procedure. None of them ever experienced the hybrid FES-exoskeleton protocol. The experiment was approved by the Ethics Committee of Shanghai Jiao Tong University, China. All subjects were volunteers and signed an informed consent before experiment.

In the formal experiments, each subject underwent three levels of stimulation according to the allocation constant ($\delta = 0.3, 0.5, 0.7$). Each stimulation session lasted for 120s. The amplitude and frequency of swing were determined by tuning parameters of the CPG module, which was designed in a symmetric structure. In this experiment, we set those parameters to obtain a constant waveform for all subjects with a peak amplitude of $\pm 30°$ and a frequency of 0.36Hz. Before the formal experiments, we did some pilot tests about FES on subjects to tune their specific parameters for the FES feed-forward controller. The relevant parameters of the controller and CPG are shown in Table I and II.

TABLE I
SUBJECT-SPECIFIC PARAMETERS

Subject	1	2	3	4
u_{sat}^{HAM}[µs]	200	300	500	150
u_{sat}^{VAS}[µs]	300	500	500	155
u_{thres}^{HAM}[µs]	100	100	100	100
u_{thres}^{VAS}[µs]	180	200	180	100
I_S[mA]	23	23	23	20

TABLE II
PARAMETERS OF CONTROLLER FOR ALL SUBJECTS

Parameter	Value	Parameter	Value	Parameter	Value
T_r	0.72	T_a	0.50	b	2.5
α	0.6	$s_{1\sim4}$	3.28	w_{12}	2.0
w_{21}	2.0	w_{34}	2.0	w_{43}	2.0
w_{13}	0.3	w_{24}	0.3	w_{32}	0.3
w_{41}	0.3	w_{14}	0.4	w_{23}	0.4
w_{31}	0.4	w_{42}	0.4	λ_1[rad]	0.87
λ_2[rad]	1.13	λ_3[rad^{-1}s]	0.04	u_{sf}[µs]	15

C. Data Processing and Experimental Results

We extracted the data during the relatively steady trials (11s- 100s) for evaluation and analysis, because the CPG module will have an unsteady "start-up" state in the first few seconds. In order to evaluate the assistive efficiency, a total torque of shank swing estimated from the actual knee angle data is necessary. From (10) we can see that torque estimation needs the information of angular position, velocity, and acceleration. However, raw angle data measured by an encoder always contain noise signals, which makes the estimation very rough. Therefore, state estimation is crucial. In our analysis, we used a method of adaptive phase oscillators to obtain approximated angular position, velocity, and acceleration, and then get a relatively smooth estimation of total torque by (10) [20]-[21]. After that, element-wise product of estimated total torque and the gain $\mathscr{G}$ can be computed to represent the desired assistance torque that the exoskeleton should provide.

The raw data were separated into trials, and each trial contains a complete cycle of knee extension and flexion. Some statistical values were calculated within each trial for further analysis: 1) the averaged error between measured knee angle and FES reference trajectory; 2) the absolute maximum angle amplitude and the duration of each trial in order to evaluate the task fulfillment of knee swing; 3) the averaged error between the desired assistance torque and measured mutual torque; 4) the absolute maximum mutual torque of knee extension and flexion. We chose the first 30 trials of data during steady state for computation. Each trial starts from the zero position. This statistic analysis gives an overview about the hybrid system's performance under the predefined experimental protocol. The results are shown in Fig. 5. We used MATLAB (MathWorks, Natick, MA) for data processing and statistical analysis.

The online phase regulator made FES reference trajectory keep in synchrony with the actual joint angle without delay, and the measured mutual torque basically conformed to

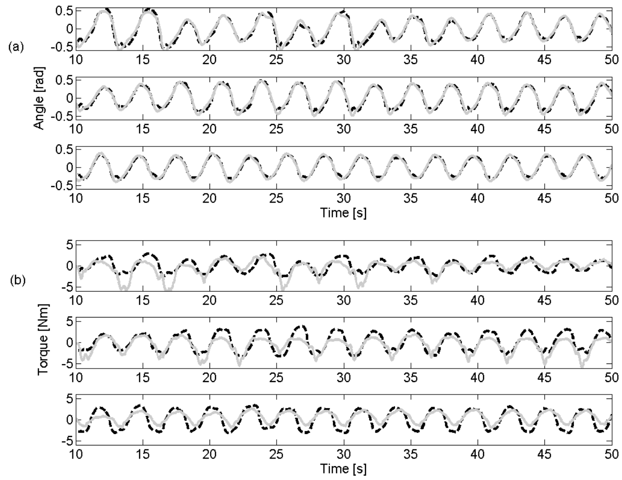

Fig. 4. Raw movement data of Subject #1 (11s- 50s): (a) comparison between the measured knee angle (dashed) and the FES reference trajectory (gray) (upper: $\delta = 0.3$, middle: $\delta = 0.5$, lower: $\delta = 0.7$), (b) comparison between the estimated desired assistance torque and the measured mutual torque (upper: $\delta = 0.3$, middle: $\delta = 0.5$, lower: $\delta = 0.7$).

the expected assistance torque under different experimental conditions [see Figs. 4, 5(d)]. The mean absolute error between measured knee angle and FES reference trajectory [see Fig. 5(a)] was 0.0022 rad in the condition $\delta = 0.3$, 0.0040 rad in the condition $\delta = 0.5$, and 0.0101 rad in the condition $\delta = 0.7$. This fact demonstrates that the online phase regulation of the CPG model makes FES reference trajectory keep in synchrony with the actual knee angle without delay. The good synergetic performance effectively reduces unexpected confliction between the machine and the human leg. One-way ANOVAs were used on performance evaluation of these trials considering the allocation ratio variation as the unique factor. ANOVA of duration showed no significance ($F(2, 87) = 2.19, p > 0.1$), revealing that the three sessions had the same duration. The absolute maximum angle in the first session ($\delta = 0.3$), however, was slightly higher than the others, but the other two sessions had the same angle amplitude ($F(1, 58) = 0.25$, $p > 0.1$). It means that if the FES actuators take too much effort, the swing range of the human leg wearing FEXO Knee may exceed the desired maximum trajectory, which would cause unexpected torque confliction.

The absolute maximum mutual torque of knee extension and flexion within each trial was used to assess the assistive effects of the three sessions, and ANOVAs showed statistical significance: $F(2, 87) = 6.22$, $p < 0.01$, for extension; $F(2, 87) = 25.05$, $p < 0.01$, for flexion. However, the cross-subject SEM were relatively large [see Fig. 5 (e), (f)] because of individual difference, especially for knee flexion. Besides, due to different stimulation response of Vasti and Hamstrings, the mutual torque varied in knee extension and flexion movement during each trial. The values of absolute maximum mutual torque (mean $\pm$ s.e.m.) for knee extension across all subjects and all trials were 2.059 ± 0.468 Nm ($\delta = 0.3$), 2.061 ± 0.277 Nm ($\delta = 0.5$), and 2.353 ± 0.343 Nm ($\delta = 0.7$). The results conform with the desired assistance levels. For knee flexion, the values were 2.256 ± 0.580 Nm ($\delta = 0.3$), 3.228 ± 0.379 Nm ($\delta = 0.5$), and 2.812 ± 0.613 Nm ($\delta = 0.7$).

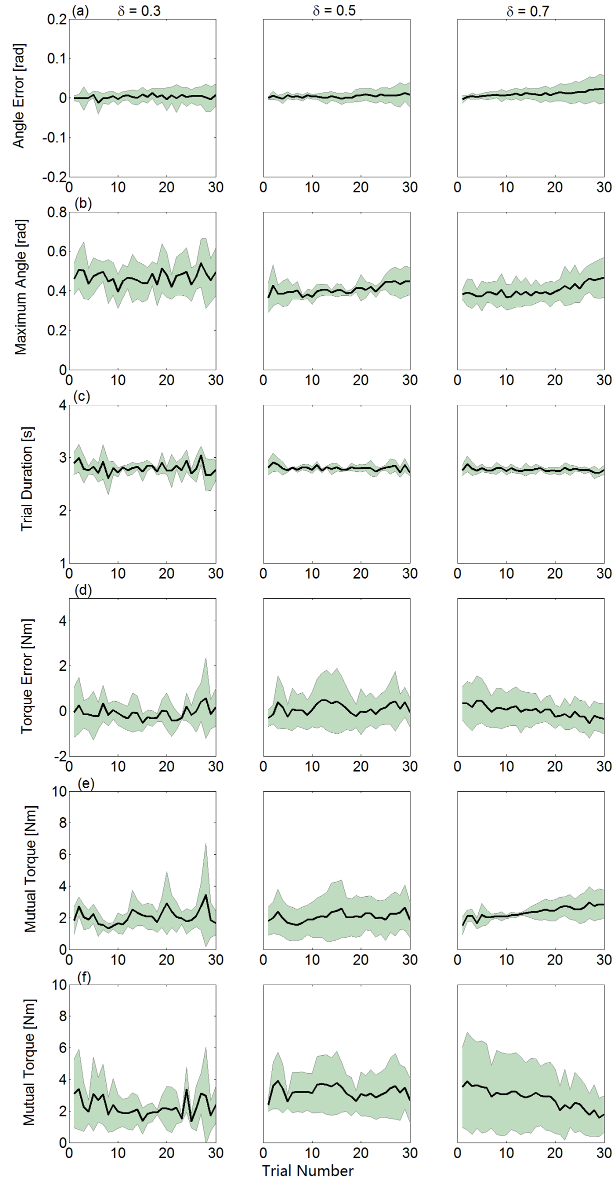

Fig. 5. Experimental results under variable assistance conditions: (a) averaged error between measured knee angle and FES reference trajectory, (b) absolute maximum angle amplitude, (c) trial duration, (d) averaged error between the desired assistance torque and measured mutual torque, and the absolute maximum mutual torque of knee (e) extension and (f) flexion. Solid lines represents the mean values across all subjects and shaded regions represent the standard error of the mean ($\pm$s.e.m) between subjects.

V. Discussion and Conclusion

This paper introduces a novel device FEXO knee for hybrid FES-exoskeleton rehabilitation, and a synergetic control frame is proposed. Some previous studies on robotic-assisted gait training show that the robot's compliance will bring about phase confliction problems [11]. Therefore, we use a CPG network with online phase regulation to avoid the possible out-of-phase trouble. The CPG module can produce robust rhythmic patterns due to its intrinsic oscillation. Besides, the output of CPG will synchronize sensory feedback signals to achieve mutual coupling, which is known as "entrainment". We can use this property to constrain the FES reference trajectory and the feedback position signal to act in synchrony. There are a lot of methods to simulate CPGs. In our control frame, an artificial neural network containing four mutually coupled Matsuoka oscillators is used to generate reference trajectories for FES and the knee exoskeleton.

For synergetic torque assistance of FES, we introduce a compliant knee exoskeleton. It is driven by a serial elastic actuator, which has been widely used in legged robots and rehabilitation robots. Compared with stiffer actuators, compliant actuators such as SEAs have some specific merits, e.g., large force shock tolerance, stable force control, the capability of energy storage and release, etc., to allow a safer human-robot interaction [14]-[15]. Although it is harder to control SEAs to precisely track the desired trajectory, they are more suitable for rehabilitation robots.

Our present work just explore some fundamental issues on interactive mechanism of hybrid FES-exoskeleton assistance. We choose a simple rhythmic movement (the swing of shank) to study, and the motion pattern of knee angle and torque is not similar to that in normal human gait. Besides, the total torque produced by FES and the exoskeleton is also smaller than that during natural gait, because it does not need to support body weight.

A preliminary experiment is conducted for performance evaluation on FEXO Knee. We can draw some important conclusions from the results. Firstly, the allocation constant δ should not be too small for practical consideration. Secondly, the knee extension and flexion should be separately considered in further experiments considering the slightly abnormal result when $\delta = 0.7$ caused by individual difference of Hamstrings stimulation response.

In future research, more experimental protocols considering variable motion amplitude and frequency will be carried out and more subjects, especially paraplegic patients, will be involved in the evaluation experiments. Besides, we will pay more attention on the improvement of structural design and practical usage of this device, to make it a wearable and light-weight system for real clinical implementation. The hybrid FES-exoskeleton rehabilitation would be a very promising method for patients to restore their motor capacity of extremities.

Acknowledgment

The authors would like to thank the voluntary subjects for participating in the experiments.

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