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A prototype rehabilitation device with variable resistance and joint motion control

Shufang Dong^a, Ke-Qian Lu^a, J.Q. Sun^{a,*}, and Katherine Rudolph^b

^a *Department of Mechanical Engineering, University of Delaware, Newark, DE 19716, USA*

^b *Department of Physical Therapy, University of Delaware, Newark, DE 19716, USA*

Abstract

Resistance exercise has been widely reported to have positive rehabilitation effects for patients with neuromuscular and orthopaedic conditions. This paper presents the design of a versatile rehabilitation device in the form of a rotating joint arm mounted on the adjustable seat that provides passive resistance during strength training for muscles. The resistance is supplied by a magnetorheological damper. Intelligent controls are developed to produce resistance force based on the prescription of the therapist. The device provides both isometric and isokinetic strength training and is reconfigurable for several human joints. Special consideration has been given to the human-machine interaction in the adaptive control algorithms that can modify the behavior of the device to account for strength gains or muscle fatigue.

Keywords

Joint motion control; Rehabilitation; Magnetorheological; Fluid; Smart materials; Intelligent control

1. Introduction

Muscle strengthening or resistance exercise has been reported to have positive effects on the recovery of normal physiological function after neurological or traumatic injuries. Due to the high cost or inadequate performance of the available resistance exercise devices in the market for rehabilitation at home, we have developed a variable resistance exercise device in the form of a knee brace which can provide precise resistive force that varies across the range of knee motion during both the isometric and dynamic resistance exercise. This paper reports a new versatile rehabilitation device (VRD) that is light weight, portable, and cost-effective, and is multifunctional and can be configured to exercise muscles at the knee, ankle, hip, wrist, elbow, and shoulder.

A number of studies have shown that the resistance exercise results in improved mobility, reduced pain, and improved stability of elder people [1,2] and patients with knee osteoarthritis (OA) [3] or rheumatoid arthritis (RA) [4]. Resistance exercise also helps improve functions in patients with impaired motor activity after stroke or in people with cerebral palsy (CP) [5-8]. A common joint disorder, for which resistance exercise is often prescribed, is knee OA, which is estimated to occur in 10% of adults over 50 [9]. Resistance exercise is an effective non-surgical intervention and is an important component of post-operative management following the arthroscopic treatment or arthroplasty [10]. Studies of exercise in people with knee OA indicate that outcomes may be dependent on the mode of exercise that is used [11,12]: the isometric and isotonic exercise is suggested for relieving knee pain, and the isokinetic exercise

* Corresponding author. Tel.: +1 302 831 8686, fax: +1 302 831 3619. E-mail address: sun@me.udel.edu (J.Q. Sun).

is recommended to increase joint stability and walking endurance. To reverse the quadriceps weakness in patients after total knee arthroplasty, Stevens et al. [13] suggested that the resistance exercise with biofeedback may be necessary to facilitate muscle activation. Miyaguchi et al. found that isometric quadriceps resistance training results in reduced pain and improved function. They found significant changes in the biochemistry of the knee joint fluid, which may partially explain the analgesic effect of resistance exercise [14]. Ide et al. found that resistance training can improve function in people with shoulder joint instability [15]. Clearly, muscle strength training is beneficial and the existence of suitable low-cost exercise devices is vital to maintaining and improving physical function for many patients.

The development of this variable resistance device was motivated by the length tension relationship of skeletal muscle in which largest force is generated when muscle fibers are at their optimal length [16-18]. In vivo muscle fibers change length with joint angle so the force producing capacity of a muscle changes across the range of joint motion and is typically highest in the mid range of joint motion. Muscle force generation during concentric exercise is also influenced by the contraction velocity as described by a hyperbolic model relating the force and velocity during contractions [19]. As contraction velocity increases, muscle force decreases. Force during isometric contractions in which the muscle length does not change is higher than the force generated during concentric contractions. When exercising with resistance from cuff weights or bar bells, the amount of resistance is prescribed so the patient can move through the entire range of joint motion. If the muscle produces less force at the ends of the range of motion, the amount resistance that allows full motion would necessarily be lower than that in the mid range of motion where the fiber length is optimal. As such, strengthening might be improved by prescribing higher resistance in the mid range. However, this is difficult to realize with currently available exercise devices.

The new VRD is capable of regulating joint motion according to the needs of the particular patient and takes muscle mechanics into consideration and adapts to limb dynamics. In addition to precisely regulating resistance to provide isometric or variable resistance to dynamic muscle contractions, the control algorithms in the new VRD are capable of monitoring the human-machine interaction in real-time and to provide visual feedback of the information on torque, joint motion and angular velocity. Furthermore, the device could extract the dynamic characteristics of the patient, such as the effect of fatigue and motivation, to modify the resistance provided in an exercise session. The device will allow us to investigate the therapeutic effects of prescribed exercise protocols in patient's own home.

The paper is organized as follows. Section 2 describes the functions of the versatile rehabilitation device. Section 3 presents the models of damper force, human muscle force, and the control algorithm for the joint motion regulation. The simulation and experimental results of the device are presented in Section 4. Section 5 concludes the paper.

2. Description of versatile rehabilitation device

Fig. 1 shows a picture of the VRD in the form of a rotating lever fixed on an adjustable seat. A magnetorheological (MR) damper provides the variable resistance and passively resists the motion of a joint. Ideally, the off-state damper force should be small so that the patient can freely move the joint with little effort. On the other hand, the damper should be strong enough to resist the motion completely, thereby allowing isometric exercise and measurement. A linear displacement sensor is used to measure the rotation of the joint. Angular velocity and acceleration are obtained by low-pass filtering and differentiating the angle measurement. Strain gauges are used to measure the resistive force. The resistive force can be controlled by varying the electric voltage across the electromagnetic coil in the MR damper. The amount of voltage applied matches the needs of the patient based on the prescription of a clinician. The

rotation center of the VRD can be aligned to the joint rotation center by adjusting the height of a track and the angular position of the device. Different attachments are used to allow exercise of many joints of the body. In this paper, however, we will describe its use in exercising the muscles involved in knee flexion and extension.

The VRD works in three modes: isometric, variable controlled resistance, and variable controlled velocity. In the isometric mode, the patient moves the knee to a certain angle and the MR damper provides enough force to lock the knee. The patient provides a maximum voluntary isometric contraction (MVIC) during knee flexion and extension. The largest torque is recorded in each direction. The MVIC is repeated at different joint angles to generate a torque profile across a range of motion required by the therapist.

In the variable resistance mode, the clinician prescribes the effort that will be required of the patient, which can be a percentage of the isometric torque profile. Once programmed, the VRD will adjust the resistance to match the desired exercise level automatically with the help of an intelligent controller as the patient exercises. In the motion control mode, the patient tries the best to flex or extend in order to meet the prescribed goals that can be isokinetic variable velocity. Real-time visual feedback is shown to the patient. The controller dynamically adjusts the resistive force to compensate the joint motion. At the same time, when patient's performance declines, due to muscle fatigue, for example, the status of the patient is estimated and the controller can adjust the exercise prescription within the guidelines from the therapist.

3. Modelling and control of the rehabilitation device

Next, we consider the dynamic performance of the VRD in the force and motion regulation. We control the voltage input to the RL network of the excitation coil to reach the required current I . The coil current directly affects the intensity of applied magnetic field H , the viscosity and yield stress τ of the MR fluid, and the resistive force f of the MR damper. The passive resistive force is adjusted according to the active force generated by the muscles to achieve the prescribed joint motion. In the following, we assume that the MR fluid and iron core of the electromagnetic coil operate in the lower linear range of the $B-H$ curve with the applied magnetic intensity far away from saturation. This assumption agrees with the device working and design conditions.

3.1. Plant model

An RL circuit is used to model the MR damper coil [20,21]:

$$u = RI + L \frac{dI}{dt} \quad (1)$$

where L is the self-inductance of the coil, R the resistance of the coil, and u is the applied input voltage. Based on the experimental observations, we assume that the resistance R is slowly varying since the damper temperature varies during exercise, and that the self-inductance L is also slowly varying because the induction $B-H$ curve is not strictly linear. The absolute value of the total resistive force of the MR damper is $f = f_\tau + f_\eta$. The magnitude of the off-state MR damper force f_η without the current input to the coil can be expressed as $f_\eta = c_1 + c_2|\dot{x}|$, where c_1 represents a constant force due to static friction and $c_2|\dot{x}|$ is the viscous force. $\dot{x}$ is the joint angular velocity. The on-state MR damper force f_τ is related to the current in the coil, the turns of the coil, MR damper gap dimensions, magnetic flux cross area and the magnetic material induction curve. Here, we let the coefficient c_3 absorb all the effects of these parameters and rewrite the total force of the MR damper as:

$$f = c_1 + c_2|\dot{x}| + c_3I \quad (2)$$

In this formulation, we use the current I as the control input and assume that c_1 is constant and can be measured, c_2 and c_3 are unknown parameters to be estimated in real-time. Note that the direction of the damper force is always opposite to the limb velocity and we omit the absolute value notation on the current I since I is always positive.

The dynamics of the joint system includes limb dynamics, external torques, articular mechanics, and muscle mechanics. It is usually described by a second-order quasi-linear system [22]. In the present case, the dynamics are described by:

$$m\ddot{x} + c\dot{x} + kx = f_{\text{active}} - \text{sgn}(\dot{x})f, \quad (3)$$

where m represents the inertia, c represents the viscosity of ligaments and articular surface, k the elastic stiffness due to ligaments or muscle cocontraction, and f_{active} is the dynamic force of the muscle group. We assume that contributions from passive viscosity c and stiffness k of the joint are small and their effects are neglected in the model. Regarding the muscles moment model, there are numerous studies to relate the muscle electromyographic (EMG) signal to the resultant muscle force [23-26], but the complex algorithms make those models difficult to incorporate into real-time controls with limited computational power.

Here, we adopt a simple dynamic force model of the muscle as $f_{\text{active}} = \beta h(x) - \alpha \dot{x}$, where $h(x)$ is the isometric force profile, $-\alpha \dot{x}$ describes the trend of decreasing muscle force with the increase of shortening velocity [27-30]. $\beta(t)$, known as the fatigue or intent coefficient, is a slowly varying parameter to compensate the reduction of isometric force after several exercise cycles. Finally, Eq. (3) becomes:

$$m\ddot{x} + \alpha \dot{x} - \beta h(x) = -\text{sgn}(\dot{x})f. \quad (4)$$

3.2. Adaptive control

We separate the control design into three loops: the current loop, resistive force tracking, and joint motion regulation. The diagram of the control concept is illustrated in Fig. 2. Outside each inner control loop, we consider the inside part as an ideal control system provided that the inner loop has stable, fast, and accurate response for the given command input.

3.2.1. Current loop—First, we design the current loop compensator. Define the coil current tracking error $e_1 = I - I_d$, where I_d is the desired current level in the coil. Denote the estimate of the resistance R as $\hat{R}$ and that of L as $\hat{L}$. The control voltage is given by:

$$u = \hat{L}\dot{I}_d + \hat{R}I_d \quad (5)$$

Substituting Eq. (5) to Eq. (1), we have,

$$L\dot{e}_1 + Re_1 = \tilde{L}\dot{I}_d + \hat{R}I_d \quad (6)$$

where $\tilde{R} = \hat{R} - R$ and $\tilde{L} = \hat{L} - L$.

Under the assumption of slowly varying R and L , we propose the adaptive law to update the parameter estimation as [31],

$$\frac{d\hat{R}}{dt} = \eta_1 e_1 I_d, \quad \frac{d\hat{L}}{dt} = \eta_2 e_1 \dot{I}_d, \quad (7)$$

where η_1 and η_2 are positive adaptation gains.

Consider a Lyapunov function $V_1(e_1) = \frac{1}{2}e_1^2 + \frac{1}{2\eta_1 L} \tilde{R}^2 + \frac{1}{2\eta_2 L} \tilde{L}^2$.

$$\begin{aligned} \dot{V}_1 &= e_1 \dot{e}_1 + \frac{1}{\eta_1 L} \tilde{R} \frac{d\tilde{R}}{dt} + \frac{1}{\eta_2 L} \tilde{L} \frac{d\tilde{L}}{dt} \\ &= -\frac{R}{L} e_1^2 + \frac{1}{L} e_1 (\tilde{L} \dot{I}_d + \tilde{R} I_d) + \frac{1}{\eta_1 L} \tilde{R} (-\eta_1 e_1 I_d) \\ &\quad + \frac{1}{\eta_2 L} \tilde{L} (-\eta_2 e_1 \dot{I}_d) = -\frac{R}{L} e_1^2 < 0 \end{aligned} \quad (8)$$

According to the invariant set theorem [31], the closed loop current system is stable and the current tracking error e_I will converge to zero as $t \rightarrow \infty$.

3.2.2. Resistive force tracking—We assume that the compensated closed loop RL network is an ideal current source. To account for the slowly varying and uncertain coefficients c_2 and c_3 of the damper force in Eq. (2), we consider adaptive controls to realize the resistive force tracking. Without loss of generality, we shall present the adaptive control for the case of joint flexion with $\dot{x} > 0$. The case of joint extension with $\dot{x} < 0$ can be treated in the same manner.

Let $\hat{c}_2$ and $\hat{c}_3$ denote the estimated values of c_2 and c_3 . The desired resistive force is f_r . The control current for force tracking is given by:

$$I = \frac{f_r - c_1 - \hat{c}_2 |\dot{x}|}{\hat{c}_3} \quad (9)$$

Define the force tracking error as $e_f = f_r - f$. Substituting the control I in Eq. (2), we have,

$$e_f = \tilde{c}_2 |\dot{x}| + \tilde{c}_3 I, \quad (10)$$

where the parameter estimation error are $\tilde{c}_2 = \hat{c}_2 - c_2$ and $\tilde{c}_3 = \hat{c}_3 - c_3$.

Similarly, we propose the adaptive law to update the parameter estimation as:

$$\frac{d\hat{c}_2}{dt} = -\gamma_2 e_f |\dot{x}|, \quad \frac{d\hat{c}_3}{dt} = \gamma_3 e_f I, \quad (11)$$

where γ_2 and γ_3 are positive adaptation gains. Consider another Lyapunov function

$$\begin{aligned} V_f &= \frac{\tilde{c}_2^2}{2\gamma_2} + \frac{\tilde{c}_3^2}{2\gamma_3}, \\ \dot{V}_f &= \frac{\tilde{c}_2}{\gamma_2} \frac{d\tilde{c}_2}{dt} + \frac{\tilde{c}_3}{\gamma_3} \frac{d\tilde{c}_3}{dt} = -\frac{\tilde{c}_2 \gamma_2 e_f |\dot{x}|}{\gamma_2} - \frac{\tilde{c}_3 \gamma_3 e_f I}{\gamma_3} \\ &= -e_f (\tilde{c}_2 |\dot{x}| + \tilde{c}_3 I) = -e_f^2 < 0 \end{aligned} \quad (12)$$

According to the invariant set theorem [31], the force tracking error will converge to zero as $t \rightarrow \infty$.

3.2.3. Motion regulation—Let us proceed to the adaptive control design for motion regulation assuming that the force controller delivers the desired force level quickly and accurately. Let $x_r(t)$ be the desired motion profile and the motion tracking error be $e = x - x_r$. Define a sliding function $s = \dot{e} + \lambda e$, where $\lambda > 0$. Let $\hat{m}$, $\hat{\alpha}$, and $\hat{\beta}$ be the estimates of m , α , and β , respectively. Going through the steps of sliding mode control [31], we arrive at a control force as:

$$f = -\hat{m}(\ddot{x}_r - \lambda \dot{e}) - \hat{\alpha} \dot{x} + \hat{\beta} h(x) + k s, \quad (13)$$

where $k > 0$ is a control gain. Substituting the control to Eq. (4), we have,

$$m \dot{s} + k s = \tilde{m} (\ddot{x}_r - \lambda \dot{e}) + \tilde{\alpha} \dot{x} + \tilde{\beta} (-h(x)), \quad (14)$$

where $\tilde{m} = \hat{m} - m$, $\tilde{\alpha} = \hat{\alpha} - \alpha$, and $\tilde{\beta} = \hat{\beta} - \beta$.

We would like the parameter estimates to converge to the true values of the corresponding parameters. To achieve this, we apply the composite adaptation law. To prepare for this, we first multiply both sides of Eq. (4) by $\lambda_f/(p + \lambda_f)$, where p is the Laplace variable and λ_f is a known positive constant. We yield:

$$m v_1 + \alpha v_2 + \beta v_3 = w, \quad (15)$$

where

$$v_1 = \frac{\lambda_f p}{p + \lambda_f} \dot{x}, \quad v_2 = \frac{\lambda_f}{p + \lambda_f} \dot{x} v_3 = \frac{\lambda_f}{p + \lambda_f} (-h(x))$$

$$w = \frac{\lambda_f}{p + \lambda_f} f \quad (16)$$

Define the predicted value of w as $\hat{w} = \hat{m} v_1 + \hat{\alpha} v_2 + \hat{\beta} v_3$. The prediction error e_w is,

$$e_w = \hat{w} - w = \tilde{m} v_1 + \tilde{\alpha} v_2 + \tilde{\beta} v_3 \quad (17)$$

Introduce the vectors $\Theta = [m \ \alpha \ \beta]^T$, $\hat{\Theta} = [\hat{m} \ \hat{\alpha} \ \hat{\beta}]^T$, $\tilde{\Theta} = \hat{\Theta} - \Theta$, $\mathbf{U} = [v_1 \ v_2 \ v_3]^T$ and $\mathbf{U}_s = [(\dot{x}_r - \lambda \dot{e}) \dot{x} (-h(x))]^T$. Eqs. (14) and (17) are reduced to:

$$\dot{s} = -\frac{k}{m} s + \frac{1}{m} \tilde{\Theta}^T \mathbf{U}_s, \quad (18)$$

$$e_w = \tilde{\Theta}^T \mathbf{U}. \quad (19)$$

Combining the adaptation for motion tracking and for parameter prediction, we obtain the composite adaptation law as:

$$\frac{d\hat{\Theta}}{dt} = -\mathbf{P} [s \mathbf{U}_s + \xi(t) e_w \mathbf{U}], \quad (20)$$

where $\xi(t) > 1/2$ is the weight to balance the updating effect of tracking error and prediction error, and $\mathbf{P}$ is a symmetric positive definite gain matrix updated by the following equation from the exponentially forgetting algorithm [31]:

$$\frac{d}{dt} [\mathbf{P}] = \varphi(t)\mathbf{P} - \mathbf{P}\mathbf{U}\mathbf{U}^T\mathbf{P}, \quad (21)$$

$$\frac{d}{dt} [\mathbf{P}^{-1}] = -\phi(t)\mathbf{P}^{-1} + \mathbf{U}\mathbf{U}^T, \quad (22)$$

where $\varphi(t) > 0$ is the forgetting factor.

Consider a Lyapunov function $V = \frac{1}{2} \left[s^2 + \frac{1}{m} \tilde{\Theta}^T \mathbf{P}^{-1} \tilde{\Theta} \right]$, which includes the motion tracking error and parameter prediction error, differentiate it with time to get,

$$\begin{aligned} \dot{V} &= \frac{1}{2} \left\{ 2s\dot{s} + \frac{2}{m} \tilde{\Theta}^T \mathbf{P}^{-1} \frac{d\tilde{\Theta}}{dt} + \frac{1}{m} \tilde{\Theta}^T \frac{d}{dt} [\mathbf{P}^{-1}] \tilde{\Theta} \right\} \\ &= -\frac{k}{m} s^2 - \frac{\varphi(t)}{2m} \tilde{\Theta}^T \mathbf{P}^{-1} \tilde{\Theta} - \frac{\xi(t) - \frac{1}{2}}{m} e_w^2. \end{aligned} \quad (23)$$

Since $\xi(t) > 12$, $\dot{V}$ is negative definite and $\dot{V} = 0$ only when $s = 0$, $e_w = 0$, and $\tilde{\Theta} = 0$. By the invariant set theorem [31], the motion tracking error e and the prediction error e_w will converge to zero, and the estimated parameters will converge to their true values.

4. Simulation and experimental results

Before we present the simulation and experimental results, we will discuss the hardware and its limitations. The DC voltage applied to the coil is within the range of 0 to 12 V. The voltage amplifier for the digital controller has a dead zone of 0.2 V and a non-proportional gain. We linearize this path by using $u = 5.56(u_c - 0.2)$ based on the experimental result, where u is the applied voltage and u_c is the corresponding digital command. The current I in the coil is measured through the voltage across a 1 Ω resistor in serial with the coil. Prior to the experiments, the resistive force and flexion angle of the VRD are calibrated.

The angular velocity of the joint is obtained by differentiating a low-pass filtered angle measurement with 5 Hz cut-off frequency. The digital version of the velocity expression is given by,

$$\dot{x}_n = 0.8834\dot{x}_{n-1} + 31(x_n - x_{n-1}). \quad (24)$$

The sample frequency of the digital controller is 250 Hz in experiments.

The nominal parameters for the control simulations are as follows: $c_1 = 15$, $c_2 = 9$, $c_3 = 100$, $R = 7.5 + 2\cos(4t)$, $L = 0.15 + 0.05\sin(4t)$, $\alpha = 10$, $\beta = 1$, $m = 4$ and $h(x) = 100 + 30\sin(10x)$. In both simulations and experiments, adaptive gains are $\eta_1 = 500$, $\eta_2 = 10$, $\gamma_2 = 10$, $\gamma_3 = 150$ and $k = 30$. The low-pass filter gains are $\lambda = 1$, $\lambda_f = 20$ and $\varphi(t) = 1$. The composite adaptation weight $\xi(t) = 1$. The initial values for adaptive control in simulation are $\hat{R} = 7.5$, $\hat{L} = 0.15$, $\hat{m} = 6$, $\hat{\alpha} = 9$, $\hat{\beta} = 1.1$, $\hat{c}_1 = 15$, $\hat{c}_2 = 6$ and $\hat{c}_3 = 80$. These values are the same for the experiments except for $\hat{\alpha} = 30$ and $\hat{\beta} = 1$. The initial value of gain matrix is $\mathbf{P} = [100 \ 0 \ 0; 0 \ 50 \ 0; 0 \ 0 \ 0.1]$.

Fig. 3 shows the simulated coil current tracking, Fig. 4 shows the simulated damper force tracking and Fig. 5 shows the motion regulation together with the online parameter estimation. The current loop and force loop show the fast and accurate tracking performance. The motion control shows good regulation. The online parameter estimation converges quickly. Note that

we have converted the time signal to the corresponding sampling period in the current loop and angular position in the force and motion regulation figures.

We implemented the above controls in the experiments. The experimental result of the adaptive current control is shown in Fig. 6. We find the experimental result of the current loop is even better than the simulation probably because the nominal values of the inductance and resistance of the electromagnetic coil are much more oscillatory in the simulation than in the experiment. The experimental result of the force tracking is shown in Fig. 7. The actual force tracking is fast and accurate except that a small overshoot is observed at the initial stage, which may come from the time-delay of the mechanical system. Fig. 8 shows the experimental results of the motion regulation of the knee joint. The hamstring muscles force parameters are dynamically predicted online. The experiments are very satisfactory indeed.

5. Conclusions

We have presented the design of a versatile rehabilitation device with variable resistance and joint motion regulation. The MR damper is optimized as the passive resistive force generator. The coupled model of the human joint and the device is proposed for control design. A multi-loop intelligent adaptive control for regulating the resistive force of the damper and attaining the prescribed joint motion pro-file has also been studied. The adaptive control effectively deals with system uncertainties of the dynamic model, and achieves excellent force and motion tracking performance. The VRD will be evaluated in a clinical setting. Its rehabilitation effectiveness on various patients will be studied in the near future.

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Fig. 1.
A picture of the versatile rehabilitation device.

Fig. 2.
Configuration of the control system for the versatile rehabilitation device.

Fig. 3. Simulation of the adaptive current control. Solid line: the reference. Dashed line: the actual current.

Fig. 4. Simulation of the adaptive force control. The upper figure: the force tracking. The lower figure: the coil voltage input. Solid line: the desired force. Dashed line: the actual force.

Fig. 5. Simulation of the adaptive joint motion regulation. Solid line: the prescribed motion profile or the true parameters. Dashed line: the actual motion or the estimated parameters.

Fig. 6. Experimental result of the adaptive current control. Solid line: the reference. Dashed line: the actual current output. This figure compares well with Fig. 3.

Fig. 7. Experimental result of the adaptive force control. The upper figure: the force tracking. The lower figure: the coil voltage input. Solid line: the desired force. Dashed line: the actual force.

Fig. 8.

Experimental result of the knee joint motion regulation. Solid line: the pre scribed motion profile or the true parameters. Dashed line: the actual motion profile or the estimated parameters. In the lowest graph, solid line: the isometric force profile; dashed line: the isokinetic force.