FES Knee Bending and Stretching System with RISE-based Tracking Control for Human Limb

Yoshihiro Kushima, Kotaro Kawataka, Hiroyuki Kawai, Yasunori Kawai, and Warren E. Dixon

***Abstract*— This paper considers a functional electrical stimulation (FES) knee bending and stretching system on robust integral of the sign of the error (RISE)-based tracking control for human limb. The knee bending and stretching motion on rowing exercises is modeled as an Euler Lagrange system by using a closed-chain mechanism. Considering the scleronomic holonomic constraint, the equation of motion expressed in terms of the seat position is obtained. The human thigh model can be divided into three pairs of antagonistic muscles. The torque of the thigh consists of a combination of torque produced by the Quadriceps and Hamstrings muscle groups. The Quadriceps and Hamstrings are contracted by the electrical stimulation, and the control of stimulation is designed applying a RISE-based control framework. Experiment results are shown on seven healthy individuals by using a rowing exercise machine. The results confirm that the proposed muscle stimulation method can realize the knee bending and stretching motion similar to voluntary tracking.**

I. Introduction

Coordinated firing of motor neurons activate skeletal muscles which generate torques about the body's joints. However, neurological diseases or injury can cause paresis or paralysis and impaired motion. In particular, upper motor neuron disorders such as stroke, spinal cord injury (SCI), Parkinson's disease lead to movement disorders that affect functional activities such as standing, walking, cycling, rowing etc. Functional electrical stimulation (FES) is a means to artificially fire the motor neurons to yield some functional movement (e.g., walking [1], standing [2], rowing [3], etc.).

For individuals that suffer from disease or injury, physical rehabilitation can lead to a recovery of some functional motion. Low impact exercise like cycling and rowing are popular means to yield rehabilitation. Often these exercises employ electrical stimulation to aid the limb motion and counter the effects of the disease or injury that limit limb motion. In [3], a manual controller has been considered to alternate the delivery of maximal constant-level electrical stimulation to knee bending and stretching motion. Davoodi and Andrews [4] proposed a fuzzy logic controller to drive the state of the rowing cycle. Miyawaki $et\ al.$ [5] developed a FES-rowing machine for elderly and paraplegic people to use for safe and effective rehabilitation exercise. Such results illustrate the potential for FES-based rowing, but they do not provide a stabilizing controller for knee bending and stretching. However, some recent studies such as [6]–[8] have focused on the development of robust integral of the sign of the error (RISE)-based FES controllers and the associated analytical stability analysis for leg extension, or cycling.

Y. Kushima, K. Kawataka, and H. Kawai are with Department of Robotics, Kanazawa Institute of Technology, Ishikawa 921-8501, Japan hiroyuki@neptune.kanazawa-it.ac.jp

Y. Kawai is with Department of Electrical Engineering, National Institute of Technology, Ishikawa College, Ishikawa 929-0392, Japan

W. E. Dixon is with Department of Mechanical and Aerospace Engineering, University of Florida, Gainesville, FL 32611-6250, USA

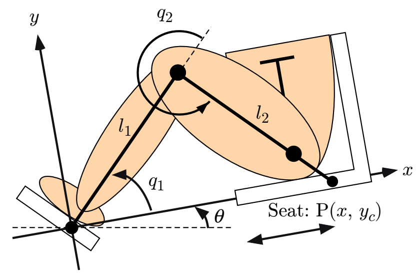

Fig. 1. Knee bending and stretching model on rowing exercises.

In this paper, we consider a FES knee bending and stretching system for RISE-based tracking control of a human limb. Knee bending and stretching is modeled as an Euler Lagrange system for applying RISE-based control framework [8]. Experiment results are shown to confirm the validity of the proposed method in seven healthy individuals.

II. Knee Bending and Stretching Model

In a rowing-based knee bending and stretching motion, the motion occurs along one axis. In this paper, we consider that the seat slides along an axis that is parallel with the x-axis as shown in Fig. 1. Then, a knee bending and stretching model can be regarded as the closed-chain mechanism [9]. Let a 2 degree of freedom (DOF) holonomic mechanical multibody system Σ' consist of a collection of rigid bodies described by the following differential equation

$$\Sigma' : \ M'(q)\ddot{q} + C'(q,\dot{q})\dot{q} + g'(q) = 0, \qquad (1)$$

where $q = [q_1 \ q_2]^T \in \mathcal{R}^2$ is the joint angles, $M' \in \mathcal{R}^{2\times 2}$ represents the inertia matrix, $C'(q,\dot{q})\dot{q} \in \mathcal{R}^2$ is the centrifugal and Coriolis terms, and $g'(q) \in \mathcal{R}^2$ is the gravity term.

From Fig. 1, the scleronomic holonomic constraint is given by

$$\mathcal{C} : \ \phi(q) = l_1 S_1 + l_2 S_{12} - y_c = 0 \qquad (2)$$

where l_1 is the length from the knee to the heel, l_2 is the length from the knee to the seat through the hip joint, and y_c is a constant seat position on the y-axis. S_{ij} and C_{ij} denote $\sin(q_i + q_j)$ and $\cos(q_i + q_j)$, respectively.

Assumption 1: The knee and ankle angles are constrained to the regions $\pi < q_2 < 2\pi$ and $\pi < q_1 + q_2 < 2\pi$ from (2) and from natural physiological constraints.

Since it is easy to measure the seat position x in the knee bending and stretching motion by using a linear encoder, a parameterization for the seat position x is developed as

$$q \longmapsto x = \alpha(q) = l_1 C_1 + l_2 C_{12}. \tag{3}$$

Using the constraint in (2) and the parameterization in (3), we define

$$\psi(q) := \begin{bmatrix} \phi(q) \\ \alpha(q) \end{bmatrix} = \begin{bmatrix} 0 \\ x \end{bmatrix}. \tag{4}$$

Differentiating (4) with respect to time yields

$$\psi_q(q)\dot{q} = \begin{bmatrix} 0 \\ 1 \end{bmatrix} \dot{x} \tag{5}$$

where

$$\psi_q(q) := \frac{\partial \psi(q)}{\partial q} = \begin{bmatrix} l_1 C_1 + l_2 C_{12} & l_2 C_{12} \\ -l_1 S_1 - l_2 S_{12} & -l_2 S_{12} \end{bmatrix}.$$

Then, the joint velocity $\dot{q}$ can be represented as

$$\dot{q} = \mu(q)\dot{x}, \tag{6}$$

$$\begin{aligned} \mu(q) &= \psi_q^{-1}(q) \begin{bmatrix} 0 \\ 1 \end{bmatrix} \\ &= \frac{1}{-l_1 l_2 S_2} \begin{bmatrix} -l_2 S_{12} & -l_2 C_{12} \\ l_1 S_1 + l_2 S_{12} & l_1 C_1 + l_2 C_{12} \end{bmatrix} \begin{bmatrix} 0 \\ 1 \end{bmatrix} \\ &= \frac{1}{l_1 l_2 S_2} \begin{bmatrix} l_2 C_{12} \\ -l_1 C_1 - l_2 C_{12} \end{bmatrix}, \end{aligned} \tag{7}$$

where $\det(\psi_q) = l_1 l_2 S_2 \neq 0$ except when $q_2 = n\pi,\ n \in \mathcal{Z}$. Thus, there exists $\psi_q^{-1}(q)$ by Assumption 1, i.e., the knee joint angle q_2 never equals $n\pi,\ n \in \mathcal{Z}$.

From (2) and the parameterization in (4), the following relation can be obtained

$$x^2 + y_c^2 = l_1^2 + l_2^2 - 2 l_1 l_2 \cos(q_2 - \pi). \tag{8}$$

Thus, q_2 is represented as functions of x as

$$q_2 = \cos^{-1}\left(\frac{l_1^2 + l_2^2 - x^2 - y_c^2}{2 l_1 l_2}\right) + \pi, \tag{9}$$

and, q_1 can be represented as

$$q_1 = \cos^{-1}\left(\frac{l_1^2 + x^2 + y_c^2 - l_2^2}{2 l_1 \sqrt{x^2 + y_c^2}}\right) + \tan^{-1}\left(\frac{y_c}{x}\right), \tag{10}$$

by using the law of cosines as

$$\begin{aligned} l_2^2 = {} & l_1^2 + x^2 + y_c^2 \\ & -2 l_1 \sqrt{x^2 + y_c^2} \times \cos\left(q_1 - \tan^{-1}\left(\frac{y_c}{x}\right)\right). \end{aligned} \tag{11}$$

The expressions in (9) and (10) yield the parameterization

$$q = \sigma(x). \tag{12}$$

Therefore, the equation of motion of the constrained system expressed in terms of the seat position x is obtained by combining

$$\begin{cases} M(q)\ddot{x} + C(q,\dot{q})\dot{x} + g(q) = 0 \\ \dot{q} = \mu(q)\dot{x} \\ q = \sigma(x) \end{cases}, \tag{13}$$

where

$$\begin{aligned} M(q) &= \mu(q)^T M'(q) \mu(q) \\ &= \frac{C_{12}^2}{l_1^2 S_2^2}(m_1 l_{g1}^2 + m_2 l_1^2 + \tilde{I}_1) \\ &\quad - \frac{2 m_2 l_{g2} C_1 C_2 C_{12}}{l_2 S_2^2} + \frac{C_1^2}{l_2^2 S_2^2}(m_2 l_{g2}^2 + \tilde{I}_2), \end{aligned} \tag{14}$$

$$\begin{aligned} C(q,\dot{q}) &= \mu(q)^T C'(q,\dot{q})\mu(q) + \mu(q)^T M'(q)\dot{\mu}(q,\dot{q}) \\ &= -\frac{1}{l_2^2 S_2^3}\left(S_1 C_1 S_2 \dot{q}_1 + C_1^2 C_2 \dot{q}_2\right)\left(m_2 l_{g2}^2 + \tilde{I}_2\right) \\ &\quad + \frac{m_2 l_{g2}}{l_2 S_2^3}\left(S_2 C_2 (S_1 C_{12} + C_1 S_{12})\right)\dot{q}_1 \\ &\quad + \frac{m_2 l_{g2}}{l_2 S_2^3}\left(C_1 C_{12} + C_1^2 C_2\right)\dot{q}_2, \end{aligned} \tag{15}$$

$$\begin{aligned} g(q) &= \mu(q)^T g'(q) \\ &= \frac{g}{l_1 l_2 S_2}\Big[m_1 l_2 l_{g1} C_{1\theta} C_{12} \\ &\qquad + m_2 l_1 l_2 C_{1\theta} C_{12} - m_2 l_1 l_{g2} C_1 C_{12\theta}\Big], \end{aligned} \tag{16}$$

to yield

$$\Sigma:\ M(x)\ddot{x} + C(x,\dot{x})\dot{x} + g(x) = F_x, \tag{17}$$

where $F_x \in \mathcal{R}$ is the force at the seat along the x-axis, $\mu(q)$ is defined in (7), and $\sigma(x)$ is the parameterization defined in (12). In (16), θ means the angle between the x-axis and horizontal direction as shown in Fig. 1. By using (14) and (15), $M(x) > 0$ and $\dot{M}(x) - 2C(x,\dot{x}) = 0$ can be confirmed.

III. Muscle Contraction Method

A. Muscle Contraction

The human thigh model can be divided into three pairs of antagonistic muscles as depicted in Fig. 2, where two groups consist of antagonistic mono-articular muscles and one group consists of antagonistic bi-articular muscles. The antagonistic mono-articular muscles that span the hip joint consist of three extensor muscles denoted by e_{m1} and two flexor muscles denoted by f_{m1}. The antagonistic mono-articular muscles that span the knee joint consist of a flexor muscle denoted by f_{m2} and three extensor muscles denoted by e_{m2}. Antagonistic bi-articular muscles span both the hip and the knee joint and consist of fe_{m3} and ef_{m3} where fe_{m3} flexes the hip and extends the knee, and ef_{m3} extends the hip and flexes the knee.

As shown in Fig. 2, the directions of $\vec{F}_{f_{m1}}$ and $\vec{F}_{e_{m1}}$ coincide with the direction of the shank, the direction of $\vec{F}_{f_{m2}}$ and $\vec{F}_{e_{m2}}$ pass through the hip and the ankle joint, and the directions of $\vec{F}_{fe_{m3}}$ and $\vec{F}_{ef_{m3}}$ are parallel to the thigh. Because the foot should be fixed for the knee bending and stretching motion on rowing exercises, the force direction at the foot is opposite at the seat by the action-reaction law. Although it is ideal that the induced force direction always parallels the movable direction, i.e., x-axis, for the rowing exercises, such a configuration would be difficult for the following reasons. While healthy individuals may be able to activate individual muscles during voluntary contractions,

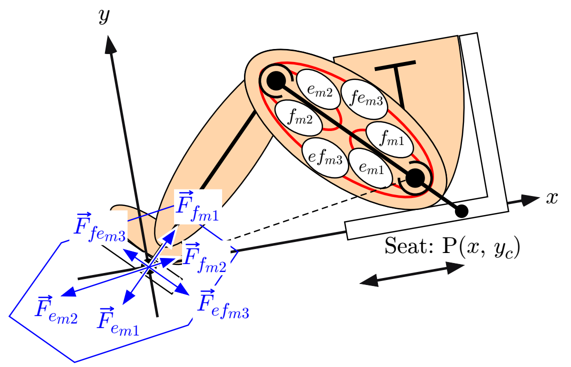

Fig. 2. Human thigh model. (i) Antagonistic mono-articular muscles spanning the hip joint consist of three extensor muscles e_{m1}, i.e., gluteus maximus, gluteus medius and gluteus minimus, and two flexor muscles f_{m1}, i.e., psoas major and iliacus. (ii) Antagonistic mono-articular muscles spanning the knee joint consist of biceps femoris short head f_{m2} and three extensor muscles e_{m2}, i.e., vastus intermedius, vastus lateralis and vastus medialis. (iii) Antagonistic bi-articular muscles spanning both the hip and the knee joint consist of rectus femoris fe_{m3} and three muscles ef_{m3}, i.e., biceps femoris long head, semimembranosus and semitendinosus. fe_{m3} flexes the hip and extends the knee, and ef_{m3} extends the hip and flexes the knee.

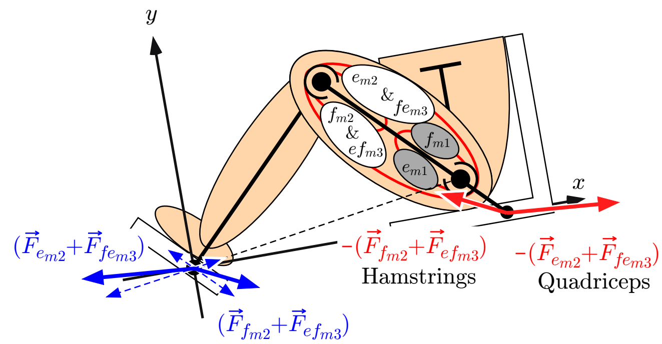

Fig. 3. Generated force by quadriceps and hamstrings at the seat.

it is difficult to selectively activate individual muscles during external FES with transcutaneous electrodes if the muscles are in close proximity to each other (e.g., the pair of vastus intermedius, vastus lateralis and vastus medialis e_{m2} and rectus femoris fe_{m3}, the pair of biceps femoris short head f_{m2} and biceps femoris long head, semimembranosus and semitendinosus fe_{m3}.). Therefore, we consider the quadriceps femoris muscle group which contains e_{m2} and fe_{m3}, and the hamstrings muscle group which contains f_{m2} and ef_{m3} to obtain the force which is as parallel as possible to the movable direction and has enough strength as shown in Fig. 3.

The torque produced at the joint(s) the muscle spans is defined as

$$\tau_i := \Omega_i u_i, \quad \Omega_i := \zeta_i \eta_i \cos(a_i), \qquad i \in \mathcal{T}, \quad \mathcal{T} := \{e_{m2}, f_{m2}, ef_{m3}, fe_{m3}\}, \tag{18}$$

where $\zeta_i \in \mathcal{R}$ denotes a positive moment arm that changes with the seat position [10], [11], $a_i \in \mathcal{R}$ is defined as the pennation angle between the tendon and the muscle which changes with the seat position [6], $\eta_i \in \mathcal{R}$ is an unknown function that relates the applied voltage to muscle fiber force which changes with the seat position and velocity, and $u_i \in \mathcal{R}$ is the control voltage input applied across each muscle group.

Assumption 2: For each bi-articular muscle, the torque acting on each of the two joints is assumed to be equal.

The force at the seat along the x-axis, denoted by F_x, is related to the joint torque $T = [T_1 \ T_2]^T$ as

$$F_x = \mu(q)^T T + M_e(q) + M_v(q) - d, \tag{19}$$

where $M_e(q) \in \mathcal{R}$ and $M_v(\dot{q}) \in \mathcal{R}$ are elastic [12] and viscous moments [13] defined as

$$M_e(q) := \mu(q)^T \begin{bmatrix} -k_{11} e^{-k_{12} q_1}(q_1 - k_{13}) \\ -k_{21} e^{-k_{22} q_2}(q_2 - k_{23}) \end{bmatrix}, \tag{20}$$

$$M_v(\dot{q}) := \mu(q)^T \begin{bmatrix} b_{11} \tanh(-b_{12}\dot{q}_1) - b_{13}\dot{q}_1 \\ b_{21} \tanh(-b_{22}\dot{q}_2) - b_{23}\dot{q}_2 \end{bmatrix}, \tag{21}$$

where $k_{11}, \cdots, k_{23} \in \mathcal{R}$ and $b_{11}, \cdots, b_{23} \in \mathcal{R}$ are unknown constants, and d is an unknown bounded disturbance from unmodeled dynamics. Moreover, the joint torques can be represented as

$$\text{Quadriceps} \begin{cases} T_1 = \tau_{fem3} \\ T_2 = \tau_{em2} + \tau_{fem3} \end{cases}, \tag{22}$$

$$\text{Hamstrings} \begin{cases} T_1 = -\tau_{efm3} \\ T_2 = -\tau_{fm2} - \tau_{efm3} \end{cases}. \tag{23}$$

Then, the total joint torque consists of the combination of the torque by the Quadriceps and Hamstrings as follows:

$$T = \chi T_{\text{Quad}} + (1-\chi) T_{\text{Ham}}, \tag{24}$$

$$T_{\text{Quad}} := \begin{bmatrix} 0 & 1 \\ 1 & 1 \end{bmatrix} \begin{bmatrix} \Omega_{em2} \\ \Omega_{fem3} \end{bmatrix} u, \tag{25}$$

$$T_{\text{Ham}} := -\begin{bmatrix} 0 & 1 \\ 1 & 1 \end{bmatrix} \begin{bmatrix} \Omega_{fm2} \\ \Omega_{efm3} \end{bmatrix} u, \tag{26}$$

where $u \in \mathcal{R}$ is the control input, and $\chi \in \{0, 1\}$ is a switching parameter.

Combining (17) and (24) yields

$$M(x)\ddot{x} + C(x, \dot{x})\dot{x} + g(x) = \Omega u - d + M_e(x) + M_v(\dot{x}) \tag{27}$$

where

$$\Omega := \mu(q)^T \left(\chi \begin{bmatrix} 0 & 1 \\ 1 & 1 \end{bmatrix} \begin{bmatrix} \Omega_{em2} \\ \Omega_{fem3} \end{bmatrix} - (1-\chi) \begin{bmatrix} 0 & 1 \\ 1 & 1 \end{bmatrix} \begin{bmatrix} \Omega_{fm2} \\ \Omega_{efm3} \end{bmatrix} \right). \tag{28}$$

B. RISE-based Tracking Control

The control objective is to develop a stimulation strategy so that a person's legs push and pull, and the rowing seat follows a desired trajectory. To quantify the control objective, the position error is defined as

$$e_1 = x_d - x \tag{29}$$

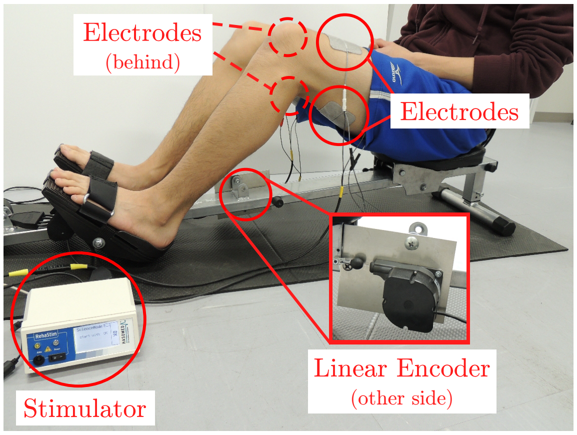

Fig. 4. FES knee bending and stretching system.

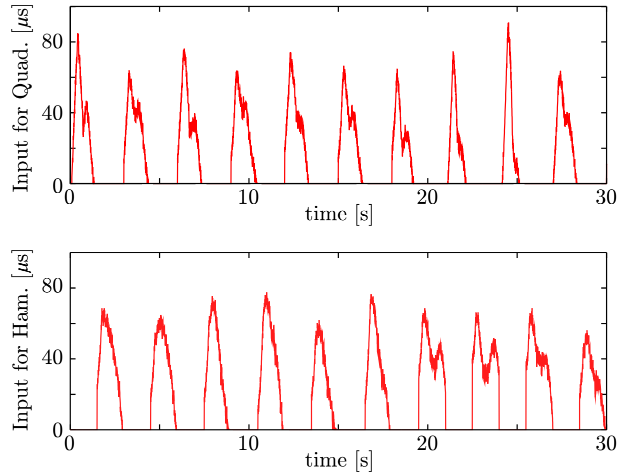

Fig. 5. Input for Hamstrings and Quadriceps.

where x_d is the desired seat position which is designed such that x_d, $x_d^k \in \mathcal{L}_\infty$, where x_d^k denotes the kth time derivative of x_d for $k = 1, 2, 3, 4$. To facilitate the subsequent analysis, auxiliary tracking errors e_2, $r \in \mathcal{R}$ are defined as

$$e_2 = \dot{e}_1 + \alpha_1 e_1 \tag{30}$$

$$r = \dot{e}_2 + \alpha_2 e_2 \tag{31}$$

where $\alpha_1, \alpha_2 \in \mathcal{R}$ are selectable positive constants. By using (29)–(31), the knee bending and stretching dynamics in (27) can be written as

$$M(x)r = M(x)(\ddot{x}_d + \alpha_1 \dot{e}_1 + \alpha_2 e_2) + C(x, \dot{x})\dot{x} \\ - M_e(x) - M_v(\dot{x}) + g(x) + d - \Omega u. \tag{32}$$

For the knee bending and stretching system in (32), the following control input [8] is used

$$u = (k_s + 1)(e_2 - e_2(0)) + \nu \tag{33}$$

$$\dot{\nu} = (k_s + 1)\alpha_2 e_2 + \beta \mathrm{sgn}(e_2), \;\; \nu(0) = \nu_0 \tag{34}$$

where ν is the generalized Filippov solution to $\dot{\nu}$, ν_0 is some initial condition, k_s, $\beta \in \mathcal{R}$ are positive, constant control gains, and $\mathrm{sgn}(\cdot)$ denotes the signum function. For the knee bending and stretching system in (32), the control input in (33) yields semi-global asymptotic tracking, i.e. $|e_1| \rightarrow 0 \;\; as \;\; t \rightarrow \infty$ [8].

IV. Experiments

A. Experimental Setup

The experiment is shown in Fig. 4. A rowing exercise machine was modified by attaching a linear encoder. Currents are sent by a RehaStim™ current-controlled stimulator with the constant frequency 40Hz. The pulse width is regarded as control input with the constant current, because a RehaStim™ can change a current from 0 to 126mA in 2mA steps, and a pulse width from 20 to 500μs in 1μs steps. Two pairs of 3” by 4” rectangle PALS® electrodes are placed over the quadriceps and the hamstrings. Control programs are written in MATLAB and SIMULINK, and implemented on a digital signal processor (DSP) from dSPACE using the Real-Time Workshop. Gain parameters were selected as $k_s = 200$, $\alpha_1 = 0.4$, $\alpha_2 = 3.6$, and $\beta = 25$ for all individual participants. The desired position and velocity were $x_d = -0.15 \cos\left(\frac{2}{3}\pi t\right) + 0.7$[m] and $\dot{x}_d = 0.1\pi \sin\left(\frac{2}{3}\pi t\right)$[m/s], respectively. From the practical point of view, we select χ as $\chi = 1(\dot{x}_d > 0)$ and $\chi = 0(\dot{x}_d < 0)$, respectively.

TABLE I

RMS Seat Position Error

Sub.	RMS[m]		Max RMS Error[m]	
	RISE	Volu.	RISE	Volu.
A	0.0351	0.0372	0.0436	0.0772
B	0.0296	0.0381	0.0371	0.0827
C	0.0406	0.0403	0.0487	0.0893
D	0.0328	0.0407	0.0422	0.0884
E	0.0416	0.0217	0.0491	0.0535
F	0.0364	0.0294	0.0472	0.0570
G	0.0376	0.0302	0.0463	0.0750

TABLE II

RMS Seat Velocity Error

Sub.	RMS[m/s]		Max RMS Error[m/s]	
	RISE	Volu.	RISE	Volu.
A	0.0881	0.0784	0.1204	0.1501
B	0.0956	0.0739	0.1334	0.1683
C	0.1215	0.0766	0.1490	0.1644
D	0.0907	0.0880	0.1147	0.1935
E	0.1024	0.0495	0.1300	0.0917
F	0.0923	0.0654	0.1098	0.1474
G	0.0905	0.0666	0.1175	0.1540

Seven healthy males, who are aged 21 to 38 years, participated in this trial. All participants provided informed consent under the supervision of the Institutional Review Board (IRB) at Kanazawa Institute of Technology. Participants do knee bending and stretching motion for 2 minutes by both the proposed muscle stimulation method and voluntary exercise. All individual participants check the position and velocity error on a monitor for reducing tracking error during voluntary exercise, while they try not to contract muscles by themselves and do not see the position and velocity error during knee bending and stretching motion by the proposed muscle stimulation method.

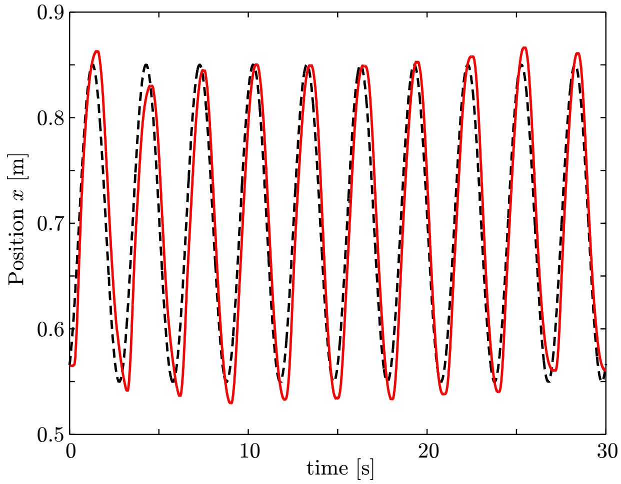

Fig. 6. Seat position on the proposed muscle stimulation method. Dashed line denotes the desired seat position.

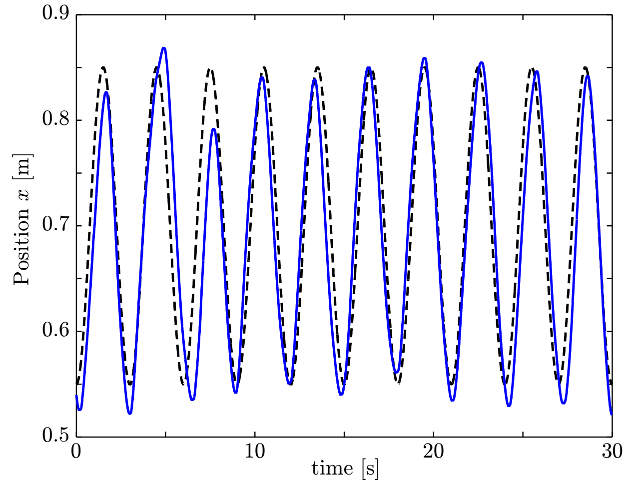

Fig. 8. Seat position on the voluntary tracking. Dashed line denotes the desired seat position.

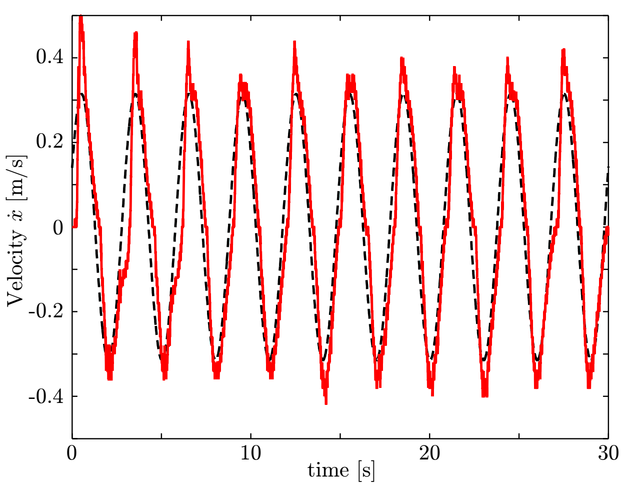

Fig. 7. Seat velocity on the proposed muscle stimulation method. Dashed line denotes the desired seat velocity.

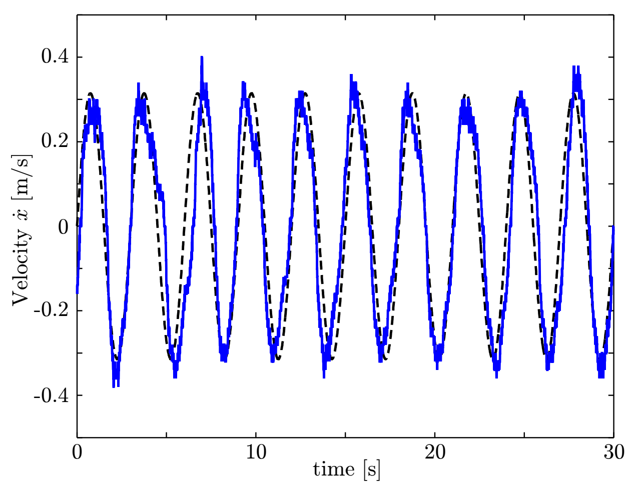

Fig. 9. Seat velocity on the voluntary tracking. Dashed line denotes the desired seat velocity.

B. *Experimental Results*

Experimental results are shown in Figs. 5–13. Figs. 5–7 illustrate the control input, seat position, and seat velocity for 30 seconds for one of the participants by the proposed muscle stimulation method, respectively. Figs. 8 and 9 are seat position and seat velocity by voluntary tracking for the same participant. Though there are small errors, we can confirm that the position and the velocity track desired ones by the proposed method similar to the voluntary tracking from Figs. 6–9.

Figs. 10–13 show the seat position error and the seat velocity error by the proposed method and by the voluntary tracking, respectively. In these figures, both the proposed method and the voluntary tracking have similar errors. Comparatively speaking, the proposed method has the periodic errors, and voluntary tracking has the irregular errors. Tables I and II show average of root mean square (RMS) error per cycle and maximum RMS error for seven individual participants. All individual participants reduced the maximum RMS position error. While the voluntary tracking is better than the proposed method in RMS velocity error for all individual participants, the maximum RMS errors are improved by the proposed method except for subject E. As a result, we confirm that the proposed muscle stimulation method can realize the knee bending and stretching motion similar to voluntary tracking. While for the results need to be determined through clinical trials in disease specific communities of people to fully evaluate the impact of this work, these results indicate a strong potential for successful clinical implementation.

V. Conclusions

This paper examines a FES-induced knee bending and stretching system using a RISE-based tracking controller. The knee bending and stretching motion is modeled as a 1DOF Euler Lagrange system based on the closed-chain mechanism for RISE-based control framework. Antagonistic bi-articular muscles are considered to decide the generated force direction explicitly. From the experiment results, it confirmed that the proposed muscle stimulation method can

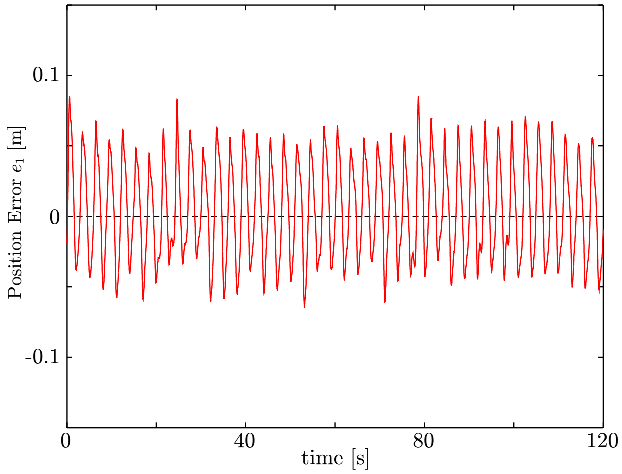

Fig. 10. Seat position error on the proposed muscle stimulation method.

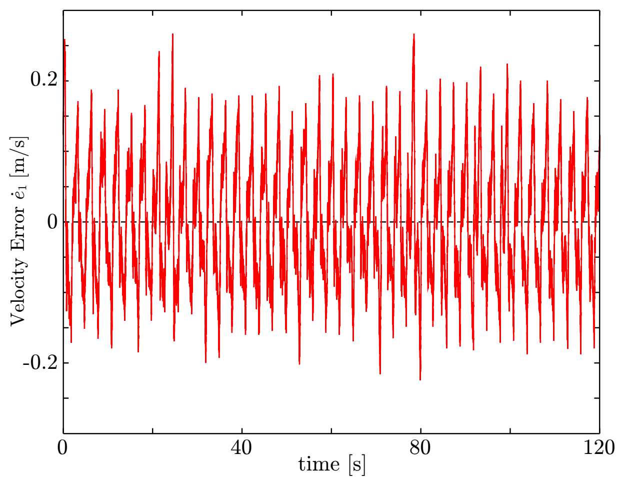

Fig. 11. Seat velocity error on the proposed muscle stimulation method.

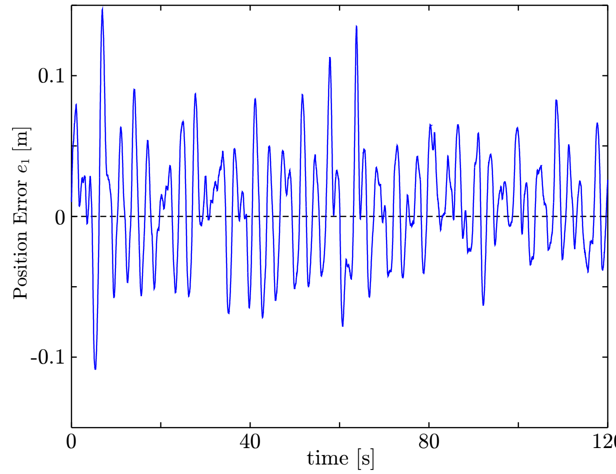

Fig. 12. Seat position error on the voluntary tracking.

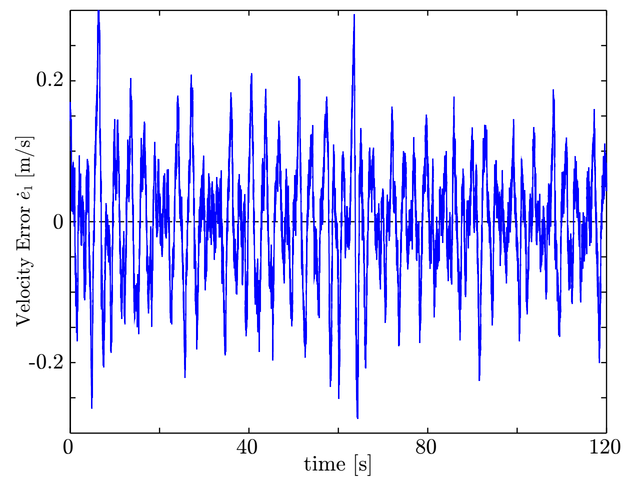

Fig. 13. Seat velocity error on the voluntary tracking.

realize the knee bending and stretching motion similar to the voluntary tracking.

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