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Identification of Patellar Tendon Reflex Based on Simple Kinematic Measurement*

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Abstract

This paper presents a method for quantifying tendon reflex dynamics addressing kinematical characteristics of the patellar tendon reflex. The method uses a limb-mounted three-dimensional motion sensor and an instrumented hammer to assess input–output relations of the patellar tendon reflex. A healthy adult male subject participated in our experiment. A simple rigid-body physical model was introduced to obtain viscoelastic and inertial responses of kinetic motion of the lower leg. This model is used to estimate knee extension torque by indicating the reflex responses of the muscle. A system identification method was then applied to describe the reflex responses to the hammer tapping by considering a second-order mathematical model with a delay term. Iterative prediction-error minimization was applied to the cascaded data for three tapping conditions: weak, medium, and strong. Good consistency was obtained between the analysis from the model and the measurement results. The results suggest that the proposed method was sufficiently feasible to characterize the reflex responses with a few characterized system parameters, which will be useful to provide additional quantitative assessment capability for neuromuscular diagnosis.

Key words : Stretch Reflex, Diagnosis, Motion Analysis, Inertial Measurement, System Identification, Kinematics, Biomechanics

1. Introduction

Assessment of the patellar tendon reflex (PTR) is a widely used clinical practice for the diagnosis of neurological disorders. Actually, PTR is elicited by tapping the muscle tendon below the kneecap. A sudden muscle stretch induced by a sharp tendon tap activates a trigger signal from a muscle spindle, which provides rapid contraction of the quadriceps muscle. This reflex response is normally observed as an extension of the lower leg with kicking movement. Absence of the reflex indicates, for instance, muscle and peripheral nerve disorders; hyperactive reflex suggests upper motor neuron disorder. Most clinical evaluations of PTR are qualitative, relying on subjective visual observations of the limb motion. Accordingly, obtaining quantitative and accurate parameterization of the reflex dynamics might enhance its utility and reliability in daily clinical practice. Recently, some advanced methods have been

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proposed to characterize tendon reflex or stretch reflex dynamics, using mechanical instruments and systemic identification techniques. Zhang *et al.* used a multi-axial torque sensor and a machine-controlled instrumented hammer, then presented a proper identification technique described with impulse response relating the reflex torque to the tapping force(1). Ju *et al.* presented a machine-controlled testing system to quantify the level of spasticity according to the resistant torque generated by the muscle stretch reflex subjected to change at various constant stretch velocities(2). These methods were aimed at presenting an accurate clinical measure of spasticity and muscle stretch reflex dynamics with precisely designed experimental equipment. Rymer, Jun, Hase *et al.* proposed practical identification methods for spasticity assessment in accordance with the clinical pendulum test procedure in which the leg is suddenly dropped from an initially extended position(3)−(5). The method considered kinematics and modeling of the lower leg oscillations as measured using a potentiometer or a video motion capture system. Mamizuka *et al.* presented a simple quantification method for clinical PTR examinations using a limb-mounted miniature accelerometer and an instrumented hammer(6). This kinematic-based approach using portable equipment enhances the most attractive feature of the PTR examination with tendon tapping, which is its applicability without elaborate laboratory equipment.

The purpose this study was to improve utility of the PTR examination in clinical practice by quantification of the reflex responses. This article presents a novel quantification procedure that can be used feasibly for clinical diagnosis with the PTR examination. The method characterized kinematic features of the PTR responses with simple mechanical system parameters. A simple instrument using a limb-mounted motion sensor and an instrumented hammer was developed to investigate input–output relations between tapping stimulations and kinematical PTR responses. Both intrinsic and reflex contributions on the kinetic dynamics of lower leg were evaluated to identify knee torque responses, represented as a consequence of muscle activities induced by the PTR.

2. Method

2.1. Experimental Procedure

Measurements were made of a normal male adult (age 40 yr, height 172 cm) who was free of all disease and who had no history of neurological disorders. The subject gave signed informed consent. The subject was seated comfortably and was asked to relax fully during tendon tapping trials with their legs hanging naturally over the seat edge. The trunk and other extremities were not restrained. Only the ankle joint was supported with a short leg brace as one rigid body for testing the lower leg. The brace also enables stable mounting of sensors. A brief rest of about 10 s was taken between tapping trials.

As instruments, we used a three-dimensional motion sensor (3DM-GX1; Microstrain, VT, USA) and a piezoelectric impact hammer (5800B4; Dytran Instruments Inc., CA, USA). The motion sensor was attached to the plantar position of the brace to obtain postural dynamics of the lower leg in PTR. The motion sensor was a united sensor combining three angular rate sensors (gyros), three orthogonal DC accelerometers, three orthogonal magnetometers, a 16-bit A/D converter, and an embedded micro-controller. The motion sensor measures three-dimensional orientations in the full 360 deg of motion on all three axes. It can also provide calibrated data from each of nine orthogonal sensors. The data transfer rate was equivalent to 100 Hz via RS-232. Total weight of the limb-mounted instrument including the ankle brace was 168 g, which has a negligibly small effect on the reflex dynamics.

The patellar tendon reflex was elicited by tapping the tendon with the instrumented hammer. The hammer was mounted on a free pendulum with an adjustable frame. Magnitudes of the tapping force were controlled by the initial release angles of the hammer-mounted pendulum. The tapping target had been adjusted previously to the most sensitive spot on the patellar tendon. The tapping force was measured using a sampling frequency of 3000 Hz. Three levels of tapping force were examined in the experiment: *Weak* (30 N), *Medium* (70 N), and *Strong*

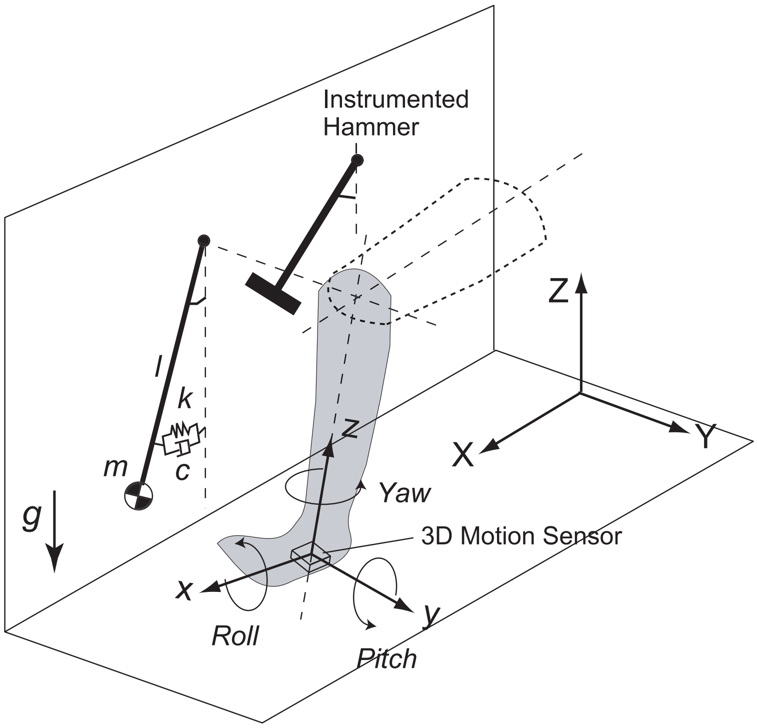

Fig. 1 Schematics of measurement setups and a mechanical model for reflex dynamics on the sagittal plane. The motion sensor was attached to the plantar position of the brace to obtain postural dynamics of the lower leg in PTR. Magnitudes of the tapping force were controlled by the initial release angles of the instrumented hammer mounted on the pendulum frame.

(120 N). All instruments were synchronized. Data were recorded using a personal computer.

2.2. Modeling of Kinematics

A major difficulty in identifying reflex dynamics from the lower leg motion is that the pure contribution of reflex activity is only slightly distinguishable from those produced by intrinsic dynamics of the lower leg. For that reason, we first identified the intrinsic dynamics of the lower leg; then we estimated the knee extension torque as a contribution of the tendon reflex responses.

Our pre-experiment result suggested that internal–external rotation was negligible. Therefore, a three-dimensional measure of the kicking motion was projected to the sagittal plane. It was then simplified to a flexion–extension motion of the knee joint. Dorsiflexion and plantarflexion of the ankle joint were also negligible because of fixation by the brace. A free single pendulum was considered as a mechanical model for the kinetic response, which consisted of a torsion spring, a damper, and a mass, as portrayed in Fig. 1. Assuming quasi-linear conditions of the system, the dynamic equation is described as shown by Eq. (1).

$$I\ddot{\theta} + c\dot{\theta} + (k - mgl)\theta = \tau \tag{1}$$

Therein, the notation m represents the whole mass of the lower leg (shank and foot); I is the whole inertial moment when the center of rotation was assumed to be located at the center of the knee joint. The weight of the limb-mounted instrument was accounted in the mass of the lower leg. Parameter k represents the spring (elastic) factor, c is the damping (viscous) coefficient, and τ is the knee joint torque. Furthermore, θ is the angular displacement of the leg motion in sagittal plane which is calculated as the pitch rotation in three-dimensional space, as depicted in Fig. 1. The angular displacement θ is the relative rotation angle from the initial posture of the lower leg hanging downward naturally. Consequently, the center of the mass

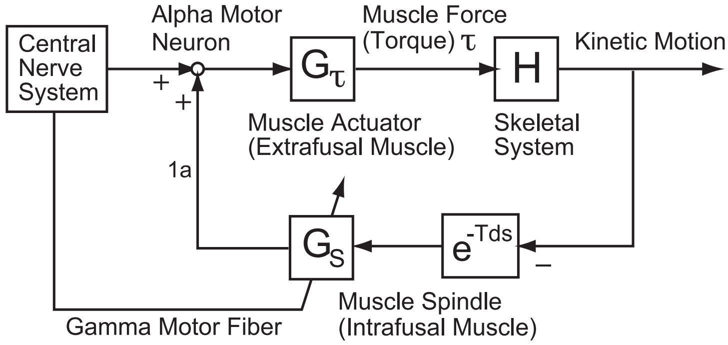

Fig. 2 Schematics of the reflex system in motor control.

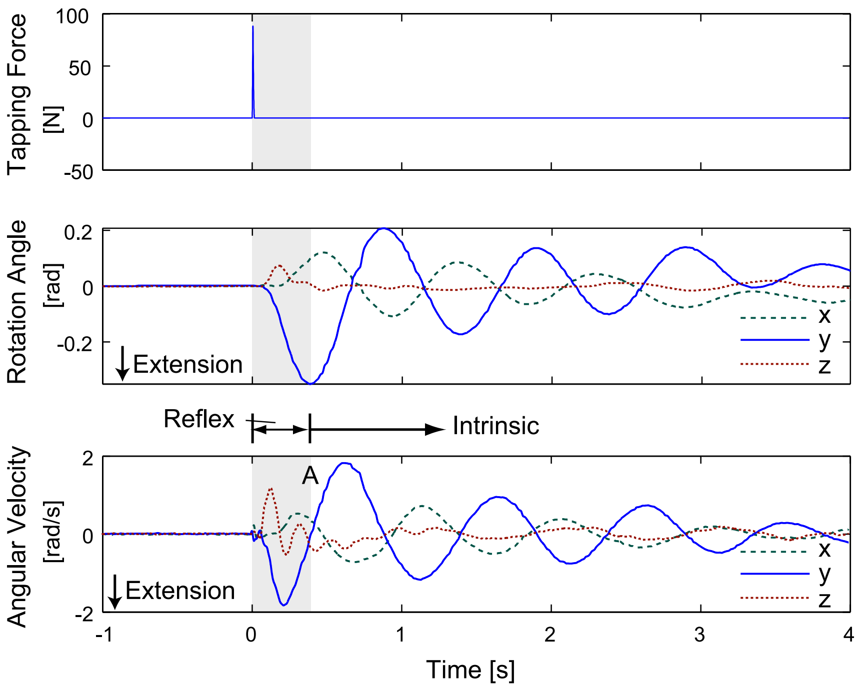

Fig. 3 Typical results of the tapping force (top) measured using the instrumented hammer. Also shown are the reflex motion dynamics as the rotation angle (middle) and angular velocity (bottom) using the leg-mounted motion sensor.

of the lower leg is directed to the gravity direction in the initial posture, indicating that θ is zero before tapping. The angular acceleration is calculated using algebraic manipulation of the gyro measurements. The notation g represents the acceleration of gravity. The link and body parameters such as mass of the segment, position of the center of the mass, and radial of gyration of the segment were estimated using published anthropometric data[7].

Before estimation of the activated knee torque τ by PTR, the unknown viscoelastic and inertia parameters were identified for the subject. The parameters were identified from data showing damped free vibration in the lower leg oscillation, which reflect intrinsic characteristics of the model. The s-domain transfer function $H(s)$ of the system was derived from the dynamic equation presented as Eq. (2). Subsequent frequency domain identification estimated the unknown coefficients α, β, and γ, in which the estimated transfer function performs the best fit to the measured in the frequency response[8], as shown below:

$$H(s) = \frac{s}{\alpha s^2 + \beta s + \gamma}, \tag{2}$$

$$\alpha = \frac{I}{ml}, \quad \beta = \frac{c}{ml}, \quad \gamma = \frac{k - mgl}{ml}. \tag{3}$$

Table 1 Obtained values of the viscoelastic and inertial parameters of the lower leg for a healthy male subject.

Parameters	Mean (SD)
Mass of the lower leg. m (kg)	4.65 (n/a)
Radius of gyration. (m)	0.19 (n/a)
Inertial moment. I (kgm^2)	0.26 (n/a)
Spring coefficient. k (Nm/rad)	19.84 (0.57)
Damping coefficient. c (Nms/rad)	0.27 (0.07)

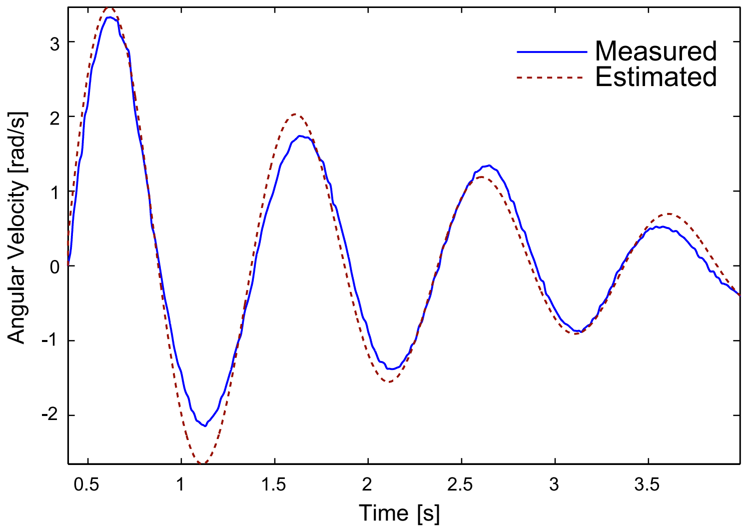

Fig. 4 Comparison between the measured and the model simulated angular velocity in the passive dynamics of the lower leg (after point A in Fig. 3.)

2.3. Identification of the Patellar Tendon Reflex

Schematics of the reflex system, as shown in Fig. 2 portray the feedback system in a motor control. In this study, the muscle response by the patellar tendon reflex was assumed to be an open loop, which is characteristic of the feedback system. Based on that viewpoint, the transfer function of the PTR reflex dynamics is considered as presented in Eq. (4).

$$G_{rf} = -G_\tau G_s e^{-T_d s} \tag{4}$$

The reflex response was parameterized as an input–output relation between the tapping force and the knee extension torque using a low-order time process model consisting of a standard second-order system with a delay term, as shown in Eq. (5).

$$G_{rf}(s) = \frac{K'}{s^2 + as + b} e^{-T_d s} \tag{5}$$

In standard form,

$$G_{rf}(s) = \frac{K}{1 + 2\xi T_\omega s + T_\omega^2 s^2} e^{-T_d s}, \tag{6}$$

$$a = 2\xi\omega_n, \quad b = \omega_n^2, \quad K' = K\omega_n^2, \quad \omega_n = 1/T_\omega, \tag{7}$$

where T_ω is a time constant, ξ is a damping coefficient, K is a static gain, and T_d is a time delay. Those parameters of the transfer function were estimated using iterative prediction-error minimization method (PEM), in which nonlinear minimization was used to determine the best second-order fit for all delays of 50–80 ms. The time delay T_d is possibly detectable for each trial by identifying the onset instant of the kicking movement in time domain, as presented in earlier study(6). We applied the model fitting method by PEM with multiple tapping data for identifying T_d to provide better comprehensive parameterization of reflex dynamics, which is close to the averaged characteristics of each response.

Table 2 System parameters obtained from the Prediction Error Minimization method: T_ω, Time constant; ξ, Damping coefficient; K, Static gain; T_d, Delay; τ_p is mean and SD value of the maximum knee extension torque for each tapping strength.

Tapping strength	τ_p [N]	T_ω	ξ	K	$T_d[ms]$
Weak	2.68(0.56)	0.064	0.949	1.401	45.83
Medium	4.48(0.49)	0.057	0.929	1.267	46.65
Strong	4.72(0.74)	0.061	0.811	0.897	54.76

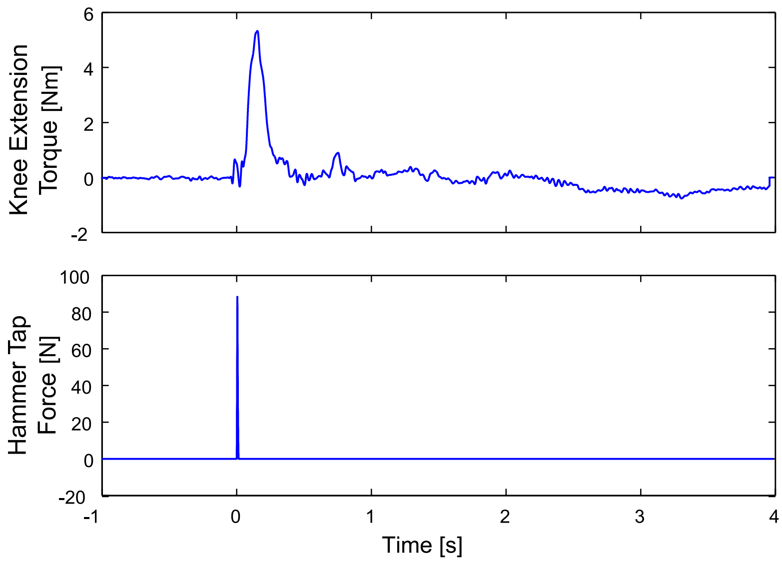

Fig. 5 Representative results of the tapping force and knee reflex torque estimated using Eq. (1).

3. Result

Figure 3 shows typical measurement results of the tapping force by the instrumented hammer (top), rotation angle (middle), and angular velocity (bottom) of the lower leg using the limb-mounted motion sensor. The rotation angles and angular velocities are expressed for each axis. The former part of the signals immediately after the hammer tapping indicates a fast kicking movement, which is apparently affected strongly by reflex activities. Subsequently, the limb motion exhibits a damping free oscillation after the maximum peak of knee extension (minimum of knee flexion) marked as point A in the figure.

First, link parameters of the rigid body model were estimated using published anthropometric data. Then the values of unknown viscoelastic and inertia parameters of Eq. (1) were identified from the latter damping free vibration part of the data using the frequency domain identification method. The mean values of estimated parameters are presented in Table 1. The values were reasonable, as confirmed through reference to the results of prior studies for the knee stiffness evaluation(9). As presented in Fig. 4, good agreement was obtained between experimental and simulation result obtained using estimated parameters in the time domain, demonstrating the reliability of the intrinsic and inertial parameter estimation. The result of knee extension torque calculated from dynamic equation (1) is portrayed in Fig. 5.

Subsequently, input–output relations between the tapping force and the knee extension torque were identified as a standard second-order system as Eq. (6). The parameters were estimated from edited records, in which only variable periods of the data were cascaded with multiple taps, as depicted in Fig. 6. It is apparent that the knee torque presented higher values for the stronger tapping forces. Identification using PEM was applied to the cascaded data containing four taps each for the *weak*, *medium*, and *strong* taps. Thereby, the system parameters were identified as mean values for four taps trials each for the tapping strength. The delay term T_d was also obtained by PEM calculation as a representative value of the

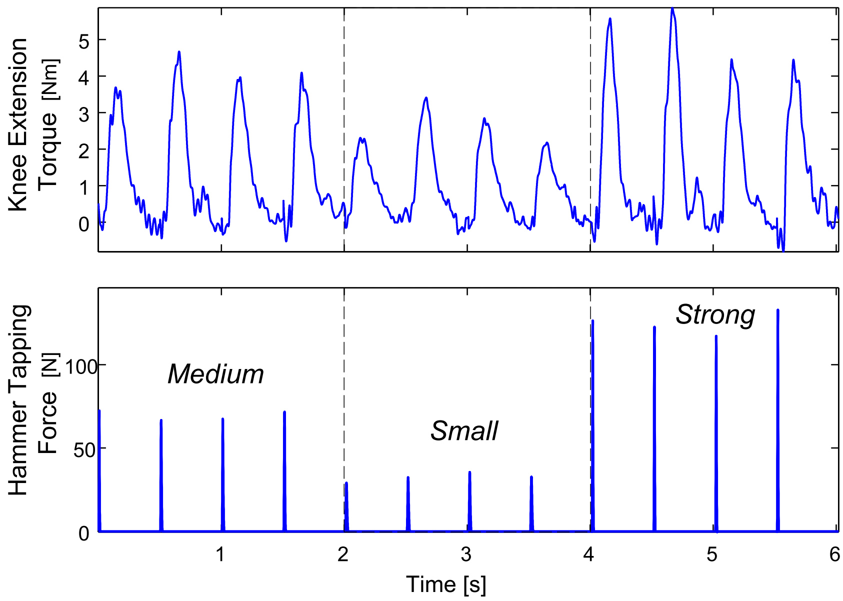

Fig. 6 Input–output signals for the transfer function identification Eq. (5), cascaded with edited signals of each tap (0.5 s long after the hammer taps were used.)

four responses. The results of all estimated parameters are presented in Table 2. Mean and SD values of the maximum knee torque τ_p for each tapping strength are also presented in the table. The maximum knee torque τ_p increased by the tapping strength. This correlation between the reflex amplitude and the tapping strength was supported by the results of previous studies[(1),(6)]. The delay term T_d was consistent for the subject in variation of tapping strength. The values showed good agreement with that reported by Zhang *et al.* who evaluated the onset instant of reflex knee torque as measured using the mechanical knee torque sensor[(1)]. Both the static gain K and the damping coefficient ξ decrease slightly in accordance with the hammer tap strength.

We compared simulation and experimental results for the subject in the time domain, as portrayed in Fig. 7. The simulation results were estimated as PTR responses using the identified transfer function. They were simulated from other independent tapping trials that were not used for system identification. Although a discrepancy was observed between the simulation and the experiment, it is both expected and acceptable. The noise and the modeling error that were included in measured input and output were causes of the discrepancy.

4. Discussion

In most cases of clinical PTR examination, intensity of the reflex response was assessed by simple scale grading based on subjective visual observations on the amplitude and velocity of the reflex responses, which uses qualitative expressions of the movement such as *absent*, *sluggish*, *normal*, *exaggerated*, and *markedly hyperactive*[(10),(11)]. The proposed system provides quantitative measures of intensity of reflex responses from kinematic movement, then promises more reliable characterization than the conventional clinical scale grading.

It has been generally reported that robustness of PTR examination is rather low because of its poor reproducibility of manual tapping stimulation. The PTR examinations normally show a large variation even for a single subject because of errors resulting from tapping disturbances[(12)]. In this study, utilization of a free pendulum stem with a rigid frame provided highly reproducible tendon tapping. Thereby, the variation of reflex responses was lessened to a tolerable degree. The coefficient of variance ($SD/Mean \times 100\%$) of maximum knee torque by reflex τ_p was as high as 20.9%. Further improvement of the reflex response variability might enhance the reliability of identification. On the other hand, modeling error in kinematics that is attributable to the simplification of rigid body dynamics might be a major source

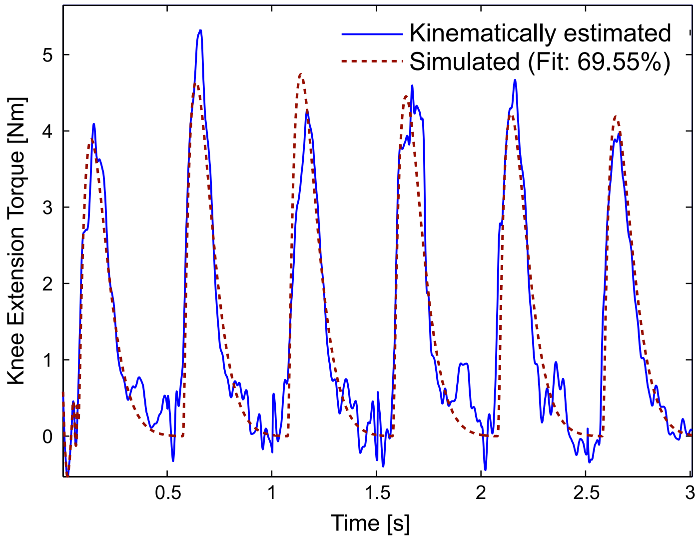

Fig. 7 Results of the system identification in comparison with simulated (dashed line, by Eq. (5)) and measurement-based knee extension torque (solid line, estimated by Eq. (1)).

of the discrepancy between the simulated and measured responses in Fig. 7. The estimated viscoelastic parameters were constants as averaged values for the entire free leg oscillation, although the parameters actually indicate nonlinear behaviors in accordance with joint angles, angular velocities, or muscle contractions. The estimated spring (elastic) parameter k became larger than that reported by anthropometric regression models and prior studies(3). This might result from involuntary muscle contractions and a wide range of knee angular oscillations.

In this preliminary stage of this research, a healthy subject was investigated to demonstrate the feasibility of the quantification procedure in the PTR examination. Although only one subject was examined in the experiment, the method characterized stretch reflex responses using a few simple system parameters and the transfer function from several measures of the reflex responses to the hammer tapping force. Reportedly, spastic patients present significantly higher amplitude in reflex response than healthy subjects(13). They also produce much stronger and more repeatable reflex responses in weaker tapping strength than healthy subjects do. It is anticipated that the presented method can be used to quantify changes in reflex responses associated with spasticity by quantifying gain characteristics of the reflex responses under tapping strength variation. However, the presented mode of modeling based characterization assuming only knee reflex movement might be inappropriate for some abnormal hyperactive disease cases showing strong reflex responses in both the knee joint and hip joint. The proposed methods should therefore be considered as possible indicators of the knee reflex dynamics by PTR in the present stage. Further investigations are necessary to discuss the reliability and neurological significance of the identified parameters with a larger number of healthy subjects and patients.

5. Conclusions

In this article, we described an alternative method for measuring and characterizing PTR dynamics. Effortless measurement using a limb-mounted three-dimensional motion sensor and an instrumented tapping hammer were developed to investigate the kinematical feature of the PTR responses. A simple mechanical body model was introduced to estimate both intrinsic and reflex contributions on the kinematical responses, then knee joint torque was obtained as a consequence of the muscle stretch reflex. System identification technique was applied to describe the relation between tapping stimulus and reflex responses with approximation to a second order transfer function with a delay term. Experiment with a healthy subject demonstrated that the gain, damping, and time delay parameters of the PTR responses were characterized quantitatively using the proposed method; they are not easily assessed using conventional scale grading. The positive aspect of the proposed method was its usability following the clinical PTR test procedure using simple equipment, which can be implemented with portable equipment. The method might be useful in routine clinical practice to quantify changes in the tendon reflex associated with neurological conditions. Further clinical investigations of the reliability and utility of the proposed method for clinical diagnoses will be subjects of our future work.

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