Proceedings of the 28th IEEE
EMBS Annual International Conference
New York City, USA, Aug 30-Sept 3, 2006

Evaluation of Spasticity in the Ankle of a Hemiplegic Subject Using a Step-like Response

Daisuke Murayama,[1] Takanori Uchiyama,[2] and Ryusei Uchida[3]
[1]School of Fundamental Science and Technology, Graduate School of Keio University, Yokohama, Japan
[2] Department of Applied Physics and Physico-Informatics, Faculty of Science and Technology, Keio University, Yokohama, Japan
[3] Department of Rehabiliation, Kanto-Rosai Hospital, Kawasaki, Japan

Abstract- **The objective of this study is to propose a new indices to evaluate spasticity in the ankle joint of hemiplegic patients. Each subject sat on a bed with one foot supported with a jig, which was used to measure the response of the ankle joint angle. The subject was instructed to relax and not to generate voluntary force. A step-like load was applied to dorsiflex the ankle joint. The ankle joint angle and electromyograms of the soleus and tibialis anterior muscle were recorded. First, the step-like response was approximated with a mathematical model, which is based on musculoskeletal and physiological characteristics using the least squares method in order to estimate net inertia and the elastic and viscous coefficients of the foot. The torque generated by the elastic component was then estimated. The normalized elastic torque was approximated with a dumped sinusoid using the least squares method. The time constant and frequency of the normalized elastic torque were calculated. We propose two indices estimated from the relationships between the time constant and the frequency. One of the indices reflected the step-like load dependency. The other reflected the difference from healthy subjects. Both indices increased as the Ashworth scale increased.**

Key words — **spasticity, elasticity, ankle, viscosity**

I. INTRODUCTION

Spasticity is an aftereffect of stroke. Stretch reflex runs high, when a joint is moved passively. That reflex made resistance force, and joint vibration. The grade of spasticity is generally mainly measured with the Ashworth scale, which is simple and does not require any equipment. It is based on a relationship between the range of motion and resistance force during extension/flexion on the joint. Therefore, it is a subjective index that sometimes fluctuates depending on the skill of the doctor. In the ankle joint, in particular, the range of motion is restricted, and ankle clonus easily presents. Therefore, diagnosing the grade of spasticity in the ankle joint with the Ashworth scale is difficult.

We have proposed a method to quantitatively evaluate the grade of spasticity in an upper limb by calculating parameters with the aid of a mathematical model [1]. New indices are proposed in this study for evaluating spasticity in the ankle joint of hemiplegic patients using a similar method.

II. METHOD

A. Subjects and Experimental Setup

Three healthy subjects and eight hemiplegic subjects participated. All gave informed consent (Table 1).

Table 1. Subjects and Ashworth scale

Subject	Age	AS
A	23	—
B	23	—
C	22	—
D	53	2
E	77	0–1
F	67	0–1
G	47	2–3
H	68	3
I	55	0–1
J	61	2
K	72	2

We measured the step-like response in the dorsal flexion of the ankle joint. Figure 1 is a schematic illustration of the experimental setup. The ankle joint angle and the electromyograms (EMGs) of the soleus and tibialis anterior muscles were stored into a computer through an analog-to-digital converter (ADC). The ankle joint was dorsiflexed by the gravitational force of a load.

Each subject sat on a chair. The foot, with the shoe on, was placed and fixed on a jig. The subject was instructed to relax and not to generate voluntary force. The starting angle was a rest angle. The pulley block was locked at the resting angle, and a load was placed on the pulley block. The pulley block was then released. Load weights of 1, 2, and 3 kg were used. The response of the ankle joint angle was measured with a potentiometer. The ankle angle was considered to be zero degrees when the subject's foot was in a natural position (when the dorsiflexion direction was positive). The EMGs were obtained with Ag-AgCl surface electrodes, 10 mm in diameter, taped to the skin 30 mm apart. EMG signals were full-wave-rectified and then smoothed with a second-order low-pass filter ($f_c = 2.6$ Hz) to obtain an integrated electromyogram (IEMG).

B. Mathematical Model

The equation of motion around the ankle joint is (1).

$$I\ddot{\theta}(t) + \eta\dot{\theta}(t) + k(t)\theta(t) = f(t), \qquad (1)$$

where I is the inertia of the foot and the measurement equipment attached to the foot, $\theta(t)$ is the relative ankle joint an-

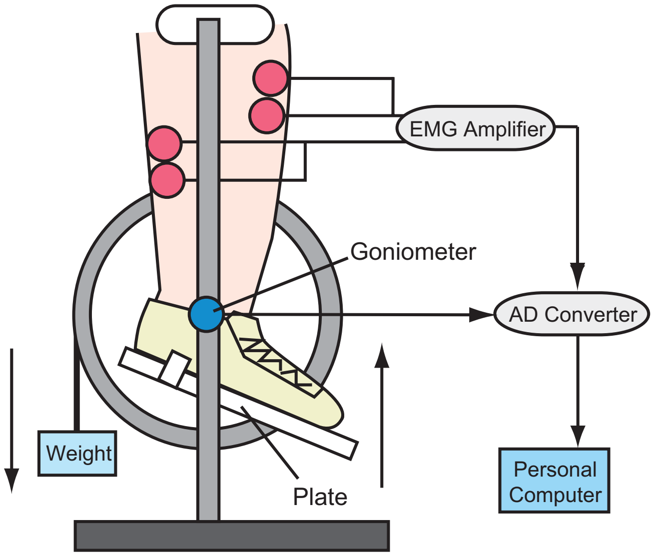

Fig. 1. Schematic illustration of the experimental setup.

gle, η is the viscous coefficient and k is the elastic coefficient. $f(t)$ is the torque applied to the elbow joint by the weight.

We assumed that the elastic coefficients were proportional to both the relative ankle joint angle and muscle activities (IEMG), as shown in (2).

$$k(t) = k_0 + k_1\theta(t) + a_1e_1(t) + a_2e_2(t), \tag{2}$$

where $e_1(t)$ and $e_2(t)$ are IEMGs of the soleus and tibialis anterior muscle, respectively.

The inertia I, viscous coefficient η and elastic coefficient k were calculated using the least squares method. The relative ankle joint angle was then estimated using these calculated values with the Runge-Kutta method and compared with the observed ankle joint angle in order to evaluate the modeling performance.

When the ankle joint angle was approximated well with the proposed model, the normalized elastic torque shown in (3) was calculated.

$$h(t) = \frac{mgr - I'\ddot{\theta}(t) - \eta\dot{\theta}(t)}{mgr}, \tag{3}$$

where $I'(= I + mr^2)$ is the net inertia. If the elastic coefficient k were constant, (1) would be a linear second order differential equation and $\theta(t)$ would be calculated as shown in (4).

$$\theta(t) = z\left\{1 - \frac{1}{\sqrt{1-\zeta^2}}e^{-\zeta\omega_0 t}\sin\left(\sqrt{1-\zeta^2}\omega_0 t + \psi\right)\right\}$$

$$\tan\psi = \frac{1-\zeta^2}{\zeta}$$

$$\omega_0 = \sqrt{\frac{k}{I'}}, \zeta = \frac{\eta}{2\sqrt{kI'}}, z = \frac{mgr}{k}, \tag{4}$$

This means that the normalized elastic torque, $k\theta(t)/mgr$ ($= \theta(t)/z$), was approximated with a dumped sinusoid in (5).

$$A\exp(-t/\tau)\sin(2\pi f t + \phi) + 1.0, \tag{5}$$

where A is the relative amplitude, τ is the time constant of dumping, f is the frequency and ϕ is the phase. Therefore time constant and the frequency were evaluated.

We propose two indices estimated from the relationships, τ-f plots, between the time constant and the frequency. First, the centroids of the markers of the τ-f plots in each load were calculated. One of the indices, D_1 was the distances between the centroids (1 kgf to 2 kgf and 2 kgf to 3 kgf). The heavier the weight was, the faster the ankle was flexed. The spastic characteristics often emerge when the joint was flexed fast. Therefore the index D_1 can be regarded as the index of the velocity dependence. Moreover, in the case of the liner differential equation, τ was $2I'/\eta$ and the frequency f was $\sqrt{4kI' - \eta^2}/4\pi I'$. This implies that both τ and f vary due to the muscle activity. Another index, D_2, was the difference between healthy and spastic subjects and was the centroid between healthy and spastic subjects applying a load of 2 kgf. The index D_2 reflected the muscle activity in k although I' varied somewhat among subjects.

III. RESULTS AND DISCUSSION

Figure 2 represents a typical example of the angle, IEMGs and normalized elastic torque of subject G, whose Ashworth scale was two to three when the weight was 2.0 kg.

The top panel (a) shows the time course of the ankle joint angle. The red and green lines denote the observed and estimated angles, respectively. The blue line denotes the residuals. The estimated ankle joint angle agreed with the observed one. The proposed model as shown in (1) and (2) approximated the ankle joint angle well.

The middle panels (b) and (c) show the time course of the IEMGs of soleus and tibialis anterior muscle, respectively. In the dorsal flexion, the soleus muscle was elongated. The IEMGs of the soleus muscle increased at the local maximums of the angle. The IEMGs of the tibialis anterior muscle increased at 0.5 s, the local minimum of the angle.

The bottom panel (d) shows the time course of the normalized elastic torque. The red and blue lines denote the observed and approximated normalized elastic torques, respectively. The approximated torque agreed with the observed one.

Figure 3 represents the time constant and frequency of the normalized elastic torque of subjects D, E, and G and three healthy subjects. The red, green, blue, and white markers denote subjects E (AS 0-1), D (AS 2), and J (AS 2-3) and the healthy subjects, respectively. The circles, triangles and squares denote 1, 2, and 3 kgf of the load, respectively. As the Ashworth scale increased, the markers were distributed farther from those of the healthy subjects. Moreover, the markers shifted as the load was increased.

Figure 4 (a) represents the relationship between the Ashworth scale and D_1 of all subjects. Figure 4 (b) represents the

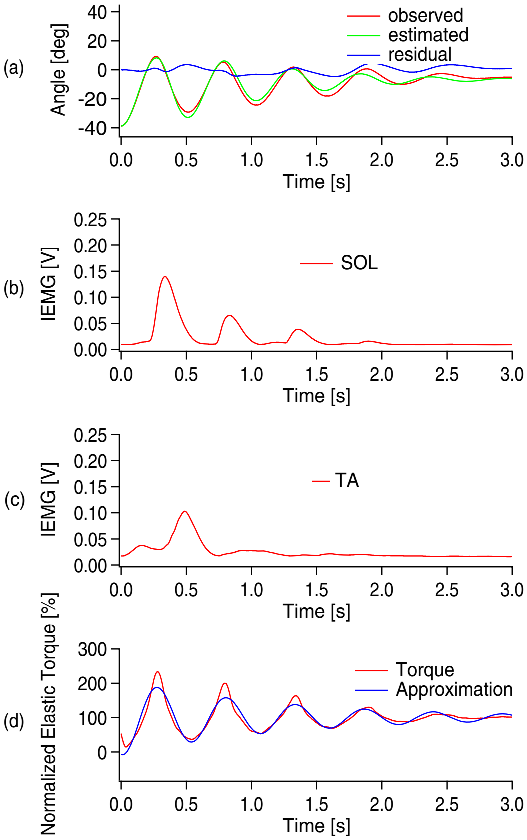

Fig. 2. Typical example of the observed and estimated data (subject G, 2 kg). (a) angle , (b) IEMG of the soleus muscle, (c) IEMG of the tibialis anterior muscle, (d) normalized elastic torque

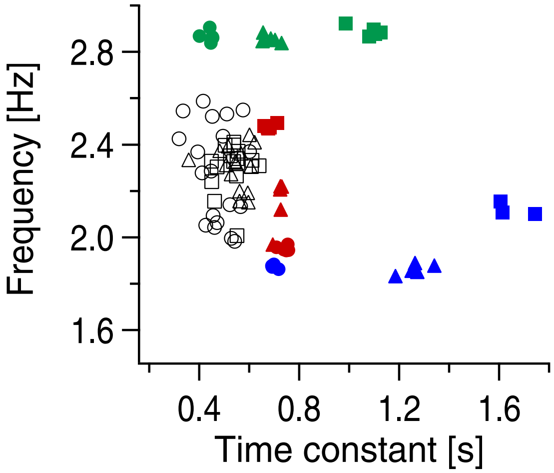

Fig. 3. Typical example of the relationship between the time constant and frequency of the normalized elastic torque. (circles: 1 kgf; triangles: 2 kgf; squares: 3 kgf; closed markers: healthy subjects; open markers: hemiplegic subjects; red: Subject E, AS 0-1; green: Subject D, AS 2; blue: Subject G, AS 2-3)

relationship between the Ashworth scale and D_2 of all subjects. D_1 and D_2 tended to increase as the Ashworth scale increased. However, in detail, there are a few exceptions. This means that both of D_1 and D_2 are necessary for a quantitative evaluation of the spasticity in the ankle joint.

IV. CONCLUSION

- The proposed method provided good approximation of the response in which a subject's ankle was passively dorsiflexed by a step-like torque.
- We proposed two indices reflecting a step-like load dependency and the difference from healthy subjects.
- The proposed indices, D_1 and D_2, increased as the Ashworth scale increased.

REFERENCES

[1] T. Uchiyama and R. Uchida: "Modeling the step-like response in the upper limbs of hemiplegic subjects for evaluation of spasticity," *IEEJ Trans. EIS,* 125, No.9, 1376-1387, 2005.

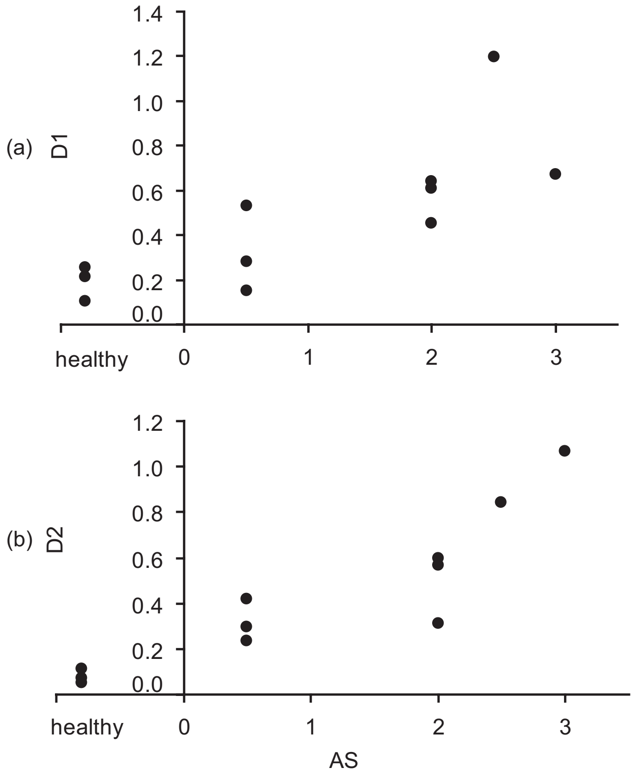

Fig. 4. Relationship between AS and (a)D1, (b)D2