SPASTICITY EVALUATION THROUGH PHYSIOLOGICAL FEEDBACK
SYSTEM MODEL IDENTIFICATION

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ABSTRACT
The quantitative and objective evaluation of spasticity has been desired by the clinical staff in rehabilitation and orthopaedics where subjective evaluations such as modified Ashworth scale (MAS) were mostly used. There have been a lot of studies for the evaluation of spasticity, but only few of them suggested quantitative evaluation with doubtful practicality for use in clinic. Data from the simple pendulum test can be used for quantitative evaluation of the spasticity with the help of biomechanical modeling. The spasticity of a knee joint can be modeled as nonlinear feedback of muscle lengthening velocity and the muscle length. Through the optimization of the modeling error, the feedback parameters can be determined which can be used as the quantitative evaluation criteria of the spasticity. This paper shows the principle and the clinical possibility of the method.

KEY WORDS
spasticity, pendulum test, feedback, muscle length, lengthening velocity

1. Introduction

Spasticity is known as a hypersensitive, velocity-dependent response to passive muscle stretch [1]. In many patients with stroke, spinal cord injury, cerebral palsy patients, spasticity reflects damage to the central nervous system that limits their ability to walk and perform fine movements needed for work and recreation [2].

Most traditional clinical assessments of spasticity for evaluation of the therapy effect have been qualitative in nature. Measures such as the Modified Ashworth Scale (MAS) [3], tendon taps, and stretch reflex tests rely on the judgment of examiner in assigning a value to the individual’s degree of spasticity [4].

Recently, mathematical approaches to this assessment have been suggested [5-11]. Among these, pendulum test [8-11] is one of the simplest measurements for clinical application, so that this paper deals with the quantitative assessment of the spasticity with pendulum test.

The main feature of this paper is twofold. One is the suggestion of spastic joint torque model as the delayed feedback of muscle length and lengthening velocity, which is thought to be physiologically feasible. The other is to model the knee joint as the sum of intrinsic (mechanical) part and spastic part and then identify each part in series. Through this, we could find the parameters of the spastic feedback system, which can reproduce the experimental pendulum trajectory of patients.

2. Method

The spasticity of two stroke patients was tested, whose anthropometric data is shown in table 1. As shown in Fig. 1, the subjects were set in supine position. EMG from quadriceps muscle and the knee joint angle, as well as the video data for the reference of data analysis, were measured during the pendulum test. EMG was measured with sampling frequency $f_s=2kHz$, amplification ratio $A_v=150$ and 12-bit resolution in AD conversion. The knee joint angle was calculated from the 3-dimensional coordinate data measured by magnetic sensor system (Liberty, Polhemus Co.). The whole experimental setup is shown in Fig. 2.

As the initial posture of the pendulum test, 0, 15, 30 degrees from full extension were used.

Table 1. The information of the subjects

<table>
<thead>
<tr>
<th>Subject</th>
<th>Age</th>
<th>Sex</th>
<th>Weight (Kg)</th>
<th>Height (cm)</th>
<th>MAS</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>66</td>
<td>F</td>
<td>81</td>
<td>171</td>
<td>1</td>
</tr>
<tr>
<td>B</td>
<td>57</td>
<td>F</td>
<td>45</td>
<td>158</td>
<td>1</td>
</tr>
</tbody>
</table>

Fig. 1 Block diagram experimental setup
The joint torque of knee joint is modeled as summation of the intrinsic one and the spastic one as in Fig. 3 and eq. (1). The intrinsic torque is composed of those coming from gravity, nonlinear damper, and nonlinear spring as in eq. (2) [12, 13]. The parameters to be estimated from the experimental joint angle trajectory are $G$, $D$, $k_1$, $k_2$, $n$, $\theta_{ROM}$. The moment of inertia ($I$) was estimated from the regression equation of height and weight [14] of each subject.

The spastic torque is modeled as the delayed feedback of the muscle length and the lengthening velocity as in eq. (3). This equation is based on the feedback properties of intrafusal Ia fiber in muscle spindle that its firing rate depends both on the length and the rate of change of length [15].

We made a daring hypothesis that the muscle length’s role in spastic reflex is only setting a threshold and the relative magnitude of the spastic torque is modulated by the muscle lengthening velocity. In the actual formulation of eq. (3), the muscle length is substituted by the knee joint angle and the lengthening velocity is substituted by the joint angular velocity for simplicity. The muscle delay indicates the delay in neuromuscular transmission through the feedback system.

$$\dot{\theta} = \left( T_{int} + T_s \right) / I$$  \hspace{1cm} (1)

$T_{int}$: intrinsic torque  
$T_s$: spasticity torque  
$I$: moment of inertia

$$T_{int} = G \sin \theta - D \dot{\theta} + k_1(e^{-x_{spastic}} - 1)$$  \hspace{1cm} (2)

$G$: Gravity coefficient  
$D$: Damper coefficient  
$k_1, k_2$: spring coefficient  
$n$: nonlinear damper index

$$T_s(t - \tau_{spastic}) = A_s \cdot u(\theta_{th} - \theta) \cdot r(\theta_{th} - \theta)$$  \hspace{1cm} (3)

$A_s$: spasticity gain  
r( ): ramp function  
u( ): unit step function  
$\theta_{th}$: angle threshold

To identify the parameters of the knee joint model, we designed two-step method to save the searching time. The first step is the identification of the intrinsic model parameters using only the latter part of pendulum data where no spastic EMG is shown as is clear in Fig. 4. The second step is to identify the spastic reflex parameters and fine tuning of the intrinsic parameters using the whole range of pendulum data. As the cost function of the parameter search, normalized RMS error between the experimental and simulated joint angle trajectories was used.

In both two identification steps, cost function was set to be the error between the experimental and simulated joint angles as in eq. (4). All the data of 3 different initial joint angles were integrated in the error calculation to enhance the generalization capability of the knee joint model.

$$\text{NRMSE} = \sqrt{\frac{\sum_{i=1}^{N} (\theta_i^{exp} - \theta_i^{sim})^2}{\sum_{i=1}^{N} (\theta_i^{exp})^2}}$$  \hspace{1cm} (4)
3. Result and Discussion

Fig. 5 and Fig. 6 show the simulation result with identified model parameters in subject A and B. The simulated joint angle trajectories match well with the experimental ones. The figures also show the spastic joint torque appearing at the end of the first falling of lower leg. These spastic torque pattern matches well with the EMG activity in quadriceps muscle in the lower graphs of the figures. The EMG activity precedes the spastic torque in the figure, which is justified by the neuromuscular delay in the muscle.

The neuromuscular delay $\tau$ was 0.086s in Subject A and 0.110s in Subject B. This result agrees well with the literature that the total neuromuscular delay is about 0.1s including the muscular delay of 0.03s and the muscle activation delay of 0.04s.

The identified model parameters are shown in Table 2. The muscle length thresholds are similar in both patients. The muscle lengthening velocity threshold of subject B is about 1.5 times of subject A, which indicates the spasticity of subject A is more severe than that of subject B. The spastic gain of each subject is not thought to represent the spastic degree, because the difference in maximum muscle force in each patient might affect the spastic gain. Therefore, we suggest the muscle lengthening velocity threshold as the index of each individual’s degree of spasticity.
4. Conclusion

In this study, we designed a new model of spasticity and identified the spastic torque of two stroke patients. Though the MAS of two patients were same, the identified spastic parameters differed. Especially, the muscle lengthening velocity threshold was suggested as the good index of the spastic degree.

Acknowledgements

This work was supported by Regional Industrial Technology Development Project, Ministry of Commerce, Industry and Energy, Korea (Grant No.IH-3-41).

References

