The Pendulum Test for Evaluating Spasticity of the Elbow Joint

Chou-Ching Lin, MD, PhD, Ming-Shaung Ju, PhD, Chun-Wang Lin, MS

ABSTRACT. Lin C-C, Ju M-S, Lin C-W. The pendulum test for evaluating spasticity of the elbow joint. Arch Phys Med Rehabil 2003;84:69-74.

Objective: To develop a modification of the pendulum test to allow evaluation of elbow spasticity.

Design: The main difficulties of directly applying the conventional pendulum test to the elbow were the small inertia of the forearm and the uncomfortable posture. We designed an experimental apparatus similar to a clock pendulum and developed an elbow biomechanic model to measure objectively spasticity of the elbow joint. The model consisted of linear stiffness and damping and gravity contribution.

Setting: A referral medical center in Taiwan.

Participants: Eleven stable stroke patients and 11 able-bodied subjects.

Interventions: A custom-designed accessory apparatus to facilitate the pendulum test in elbow joints.

Main Outcome Measures: By using an optimization technique, we estimated parameters of the proposed elbow biomechanic model as the candidate indicators of spasticity.

Results: The stiffness constant remained relatively consistent in all groups. Both the damping coefficient and damping ratio increased in the affected side of stroke patients and tended to increase with the severity of spasticity. Damping ratio had marginally better differentiation capability than the damping coefficient.

Conclusions: The damping ratio derived from the proposed model differentiated spasticity from normotonus and increased as spasticity increased.

Key Words: Elbow; Muscle spasticity; Rehabilitation; Stroke.

SPASTICITY IS A COMMON SYMPTOM in spinal cord injury and stroke patients. There is no simple quantitative method to evaluate the severity of spasticity. The commonly used Ashworth Scale[1] is a semiquantitative tool. The pendulum test, developed initially by Wartenberg[2] for knee joint, is a simple test. The examiner first lets the subject lie on a test bed with the legs hanging off the bed. Then the examiner raises 1 of the subject's legs to the horizontal position and then lets the leg fall and swing freely. The number of swings is taken as an indicator of muscle tone. The pendulum test was later modified and refined by several researchers and thus has become a quantitative measurement. These researchers also proposed many different parameters for quantifying the severity of spasticity. The simpler parameters include calculating ratios of peaks and troughs[3,4] and angular velocity.[5] More complicated parameters are derived from estimating proprietarily proposed models of the knee joint. The proposed models range from simple linear models[6,7] to more complicated models incorporating structural and physiologic details[8,9] and time-varying parameters.[10] For the clinical applications, Katz et al[4] showed that the results of the pendulum test correlate well with clinical perception. Brown et al[11] showed that the relaxation index, derived from the pendulum test, can be used for differentiating between spasticity and rigidity.

The advantages of the quantitative pendulum test are simplicity, reproducibility,[12] safety, and quantitative description. However, the test is only natural for the knee joint, and it is cumbersome to apply the test to the elbow joint. The main difficulties include the smallness of forearm inertia and the uncomfortable posture. Usually, the subjects have to be prone with their forearm extending out of the test bed, and the test is started from the extension position. The result may be unsatisfactory even in this uncomfortable posture, because the more spastic-prone muscles of the elbow joint (the flexors) are stretched at the beginning of the test.

In this study, we designed a simple accessory apparatus, which solved both of the above-mentioned difficulties, to assist in performing the pendulum test with the elbow joint. We also propose a biomechanic model of the elbow by using the pendulum test and formulate parameters for quantification. The model is a linear additive stiffness-damping model with an additional term that represents the gravity effect. Model parameters are estimated by using optimization techniques.

METHODS

Participants

Eleven stroke patients with spasticity (mean age, 57.7 ± 16.1y) and 11 able-bodied, right-handed subjects (mean age, 59.5 ± 11.8y) were recruited (table 1). All of the subjects were men. The diagnosis of stroke was confirmed initially by both history and image studies. The medical and neurologic conditions of the patients were stable for at least 6 months. We intentionally chose patients without elbow contracture. No patient was taking antispasticity medication at the time of experiment. Able-bodied subjects were age-matched volunteers. The study protocol was approved by the local ethics committee. Before an experiment, the purpose, the potential hazard, and the procedure were fully explained to each subject, and a written permission form signed. Muscle force and muscle tone were graded manually by 2 qualified neurologists with semiquantitative measures, the Medical Research Council (MRC) Scale[13] and the Modified Ashworth Scale[14] (MAS), respectively (tables 2, 3). The body weight and forearm length were measured for later estimation of mass, center of mass, and

From the Department of Neurology, University Hospital (C-C Lin); and Department of Mechanical Engineering, National Cheng Kung University (Ju, C-W Lin), Tainan, Taiwan.

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Reprint requests to Chou-Ching Lin, MD, PhD, Dept of Neurology, National Cheng Kung University, 138 Sheng Li Rd, Tainan, Taiwan 701, e-mail: *cxl45@mail.ncku.edu.tw*.

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Table 1: Basic Data of Able-Bodied Subjects and Stroke Patients

Group	Age (y)	Body Weight (kg)	Muscle Force (MRC)		Spasticity (MAS)	
			Right	Left	Right	Left
N1	51	61	5	5	0	0
N2	74	59	5	5	0	0
N3	73	65	5	5	0	0
N4	46	62	5	5	0	0
N5	60	73	5	5	0	0
N6	73	50	5	5	0	0
N7	64	60	5	5	0	0
N8	58	62	5	5	0	0
N9	55	64	5	5	0	0
N10	64	72	5	5	0	0
N11	37	70	5	5	0	0
S1	42	68	4+	5	1	0
S2	53	62	5−	4−	0	1
S3	59	69	4	5	1	0
S4	82	55	5	3	0	1
S5	71	50	5	4	0	2
S6	74	55	5	5−	0	2
S7	51	55	5	4	0	3
S8	34	75	3	5	3	0
S9	74	65	5	0	0	3
S10	37	53	5−	5	2	0
S11	58	56	5	5−	0	2

Abbreviations: N, able-bodied; S, stroke.

inertia of the forearm.[15] The test was performed on both sides for all the subjects, except in 2 patients who refused to do the test on the healthy side.

Experimental Setup

The whole experimental setup (fig 1) included 3 parts: the accessory apparatus, electromyography, and a data acquisition system. The accessory apparatus consisted of a shaft, a weight, and a part that fastens to the wrist. The steel shaft, 107.5cm in length, was connected at the midpoint to the test bed through a pure rotary joint. An electronic goniometer at this joint measured the elbow joint angle. A weight (1.13kg) was fastened to the lower end of the shaft to increase the total inertia and counterbalance the weight of the forearm. The wrist-fastening part, a Teflon®-coated plate with soft cushion, was connected to the upper part of the shaft through a sliding-rotary joint. The additional degree of freedom in sliding was designed to avoid hindering movements encountered when the centers of rotation of the elbow joint and the accessory apparatus did not coincide. Electromyograms (EMGs) of biceps and triceps brachii were

Table 2: MRC Scale[13]

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Table 3: Modified Ashworth Scale[14]

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Fig 1. A subject with the accessory apparatus in the experiment.

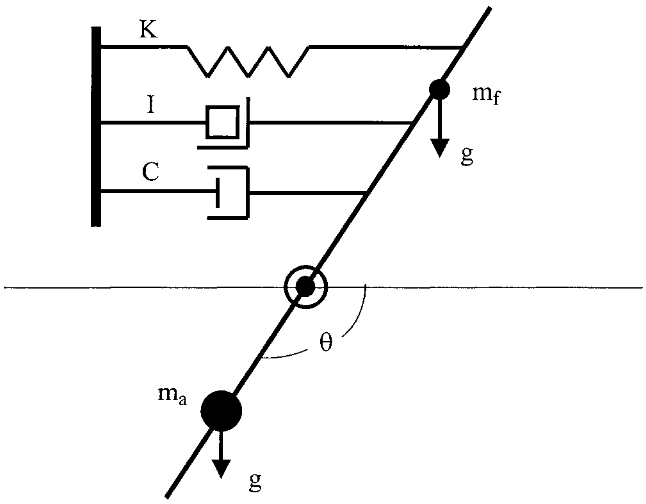

Fig 2. Biomechanical model of the elbow joint with the accessory apparatus (see text for description).

collected with 2 pairs of standard cup electrodes and amplified with an analog band-pass filter (1.59–300Hz) by using the Polygraph 360 system.[a]

All channels of data (1 channel of joint angle, 2 channels of electromyography) were sampled at 600Hz for 15 to 25 seconds, depending on the duration of swing, and stored in a personal computer for offline analyses. The data collection was accomplished with a program written in LabView.[b]

Experimental Procedures

The subject lay on the test bed in the supine position. The distal part of the forearm and the wrist were fixed with an elastic strip to the wrist-fixing part of the accessory apparatus. The electromyographic electrodes were attached to the motor points of biceps and triceps brachii. Several pretrial swings were performed to familiarize the subject with the sensation of pendulum motion. The upper part of the shaft was hooked to the test bed with a chain of predesigned length, such that the elbow joint angle was 130° (full extension, 0°). Because the muscle tone was history dependent, to eliminate the effects of previous movements, the tilted position was maintained for 2 minutes. The subject was told to relax as much as possible and not to interfere with the pendulum motion. The data collection was started and, about 5 seconds later, the chain was released swiftly without informing the subject. After the swing motion stopped, determined by visual inspection, the data collection was terminated.

During the test, EMGs and angle trajectory appeared on the computer screen in real time. If the experimenter observed large activity in the EMG at inappropriate times or an abnormal angle trajectory, for example, the swing becoming larger with time, the trial was discarded. Six qualified trials in total for each side were collected for offline analyses.

Elbow Joint Models for Analyzing the Pendulum Test

We constructed a biomechanic model of the elbow joint for data analyses (fig 2). The dynamic characteristics of the whole system was expressed in the following equations:

$$I\ddot{\theta} = -\tau_g - K(\theta - \theta_e) - C\theta \quad (1)$$

$$I = I_a + I_f \quad (2)$$

$$\tau_g = m_a g\ L_a \sin(\theta - \frac{\pi}{2}) - m_f\ g\ L_f\ \sin(\theta - \frac{\pi}{2}), \quad (3)$$

where θ is the elbow joint angle; τ_g is the torque caused by gravity; K is the stiffness constant; θ_e is the threshold angle; C is the damping coefficient (also called the coefficient of viscosity); g is the gravity constant; and I_a, m_a, L_a, I_f, m_f, and L_f are the inertia, mass, and length of the accessory apparatus, and forearm (including hand), respectively. L_a and m_a could be measured and I_a was estimated with the method proposed by Chen.[15] L_f and body weight were measured. I_f and m_f were calculated from L_f and body weight by anthropometric formulae.[16]

Parameter Estimation of the System

The angle trajectory was first low-pass filtered (cutoff frequency, 10Hz) with a fourth-order Butterworth filter. All the 6 trials were compared and those that were out 1 standard deviation (SD) were discarded. If fewer than 3 trials remained, the whole data set was discarded; if more than 3 trials remained, all the remaining trials were averaged. The method used for parameter estimation was similar to the method that Li et al[17] developed for the knee pendulum test; it is briefly summarized in appendix 1. The goodness of parameter estimation was evaluated with root-mean-square (RMS) error between the actual and estimated elbow angle trajectories.

Data Analyses and Statistics

In addition to the model parameters, we also calculated the damping ratio (η) of the biomechanical model as:

$$\eta = \frac{C}{2\sqrt{K \cdot I}}, \quad (4)$$

where C, K, and I were defined in equations 1 through 3. For each parameter, we first tested whether the difference between 2 sides of the able-bodied subjects was significant. Because there was no significant difference, we merged the results of 2 sides as a single normal group for comparison later. Then, we used analysis of variance to test for differences among the able-bodied group, and the healthy and affected sides of the stroke patients with an α value of .05.

RESULTS

Estimating the Stiffness and Damping of the Accessory Apparatus

Before the main experiment, trials without subjects were performed first to estimate the properties of the accessory apparatus. The estimated parameters (stiffness constant, damping coefficient, RMS error) were K=0N·m/rad, C=.008N·m·s/rad, and RMS error=5.77° for the first 10 seconds. The discrepancy became more prominent in the later part of the pendulum movement. Because it was expected that the magnitude of pendulum motion would decrease with time in the actual experiment, the effect of discrepancy would also decrease. Therefore, the stiffness and damping effects of the accessory apparatus were neglected in parameter estimation of the main experiments.

Qualitative Description of Typical Individual Results

Figure 3 showed typical results of subjects with different Ashworth grades. For the normal subject N4, the pendulum motion stopped after 4 swings, at about 7 seconds. As the spasticity became more prominent, the magnitude and number of swings decreased, and the total time of pendulum motion

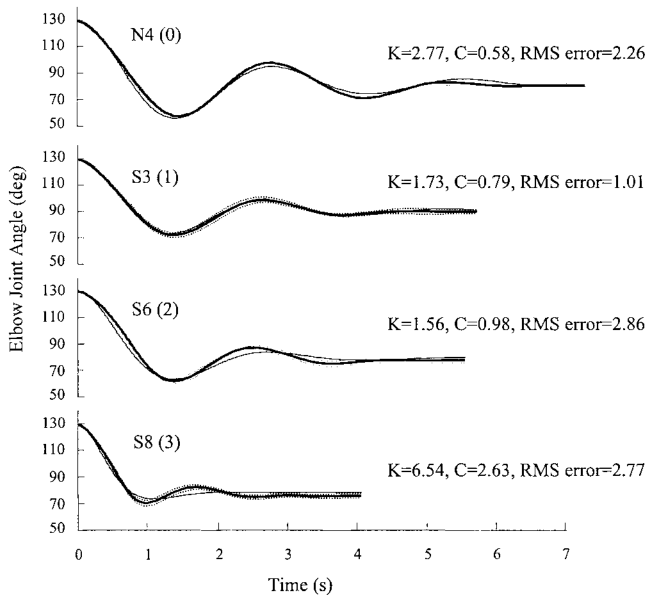

Fig 3. Angle trajectories of different severity of spasticity. The number in parenthesis after the subject number is the Ashworth grade. The thick line is the mean experimental results, dashed lines are 1 SD ranges, and the thin line is the estimated result.

also decreased. The swing also became more asymmetric with respect to the final angle and more nonlinear. C increased with increasing spasticity, whereas K and RMS error showed no particular trend. There was no point on the angle trajectory, suggesting the threshold angle of stretch reflex.

Parameter Estimation Results With the Biomechanic Model

The estimated angle trajectories fit the experimental results well. The mean RMS error ± SD for the able-bodied group, and healthy and affected sides of the stroke patients were $4.32°\pm1.86°$, $2.35°\pm0.68°$, and $2.07°\pm0.66°$, respectively. The group means and SDs of the estimated model parameters (K, θ_e, C, η) are shown in figure 4. There was no significant difference among 3 groups for K ($F_{2,39}=1.19$, $P=.315$) or for the threshold angle (θ_e) ($F_{2,39}=1.16$, $P=.325$). The difference between able-bodied subjects and healthy side of the stroke patients alone for K was marginally significant ($F_{1,29}=4.54$, $P=.042$). On the contrary, damping coefficient (C) and damping ratio (η) differed significantly among the 3 groups. Both C ($F_{2,39}=31.73$, $P<.001$) and η ($F_{2,39}=31.50$, $P<.001$) of the able-bodied subjects were smaller than those of either side of the stroke patients. When comparing C ($F_{1,18}=13.92$, $P=.0015$) and η ($F_{1,18}=10.92$, $P=.0039$) of 2 sides of the stroke patients alone, the difference was also significant.

DISCUSSION

Comparison of Estimated Parameters With Experimentally Derived Constants

Because the properties of the elbow joint are not linear, the values of experimentally derived constants depend on the assumptions, the test conditions, and the range of motion (ROM). It is difficult to compare results from different studies. The estimated stiffness constant of the normal subjects (1.01–4.42N•m/rad) in our current study accorded well with the published data, which range from 0.74 to 2.2N•m/rad.[18-20] The estimated damping coefficient (.11–.79N•m•s/rad) was larger than the values reported in other studies (eg, 0.1–0.3N•m•s/rad).[20] The additional weight of the forearm may increase friction in the apparatus joint. Yet, in our preliminary tests, the damping coefficient only increase .01N•m•s/rad when we added a .25-kg standard weight to the upper part of the bar. Misalignment of the centers of rotation between the apparatus and the elbow may also contribute to the increased damping coefficient. Finally, different settings of experiments may have a large influence on estimating the damping coefficient.

Decreased Stiffness Constant on the Intact Side of Stroke Patients

A decreased stiffness constant on the healthy side of the stroke patients is unexpected. We thought the stiffness constant for both sides in stroke patients would increase because of their relatively sedentary life style, which might cause increased content of fibrous tissue in both connective tissue and muscle. Fowler et al[21] argued that the soft-tissue change may fully explain the results of pendulum test in the knee joints of stroke patients. We performed similar analyses (fig 5), which showed significant difference in the angle of reversal ($F_{2,39}=13.03$, $P<.001$) between normal subjects and stroke patients. Maximal angular velocity also showed a similar trend, although the difference did not reach statistic significance ($F_{2,39}=2.36$, $P=.108$). Because no patient in our study had a limitation in ROM, the argument that joint angle limitation had an important impact on the results of pendulum test is not applicable. On the contrary, our results indicate that the changes in the mechanical properties of soft tissue cannot be the sole explanation for the results in the elbow pendulum test. This suggestion does not argue against the proposal that limitation in ROM has a major role for the knee pendulum test. Because passive properties have more influence on the overall performance of the knee joint in the normal subjects, it is possible that passive properties may also have a larger influence on pendulum test of this joint.

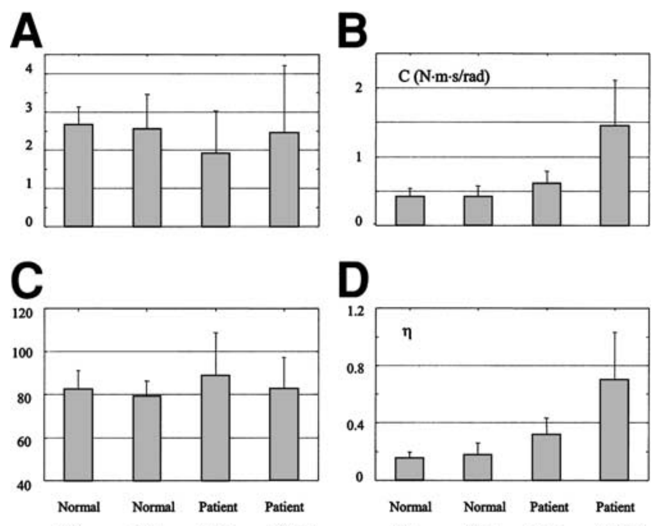

Fig 4. Results of parameter estimation. (A) K of the healthy side of the stroke patients is smaller than K of able-bodied groups. (B) No difference in θ_e was observed among groups. (C) C and (D) η of the affected side of stroke patients are significantly larger than the corresponding numbers of the other 3 groups.

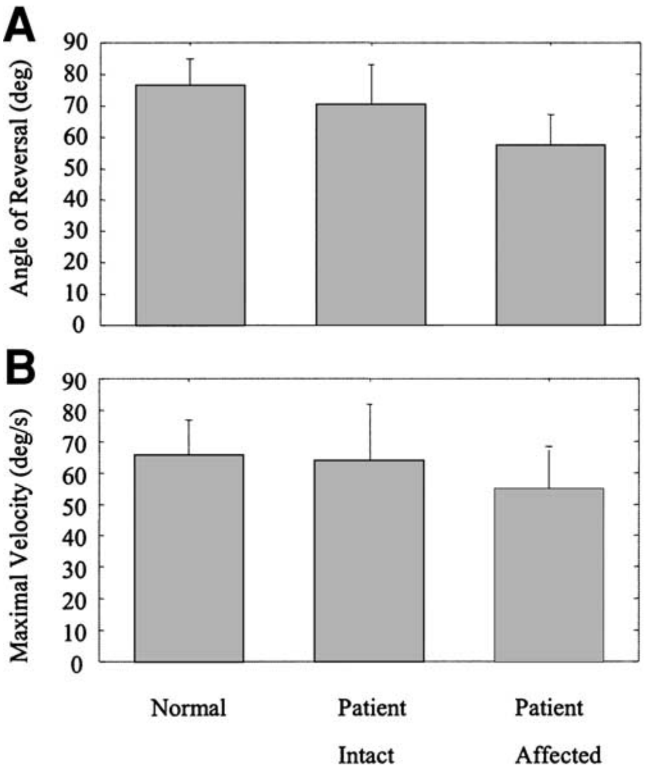

Fig 5. Comparison of (A) angle of reversal and (B) maximal swing velocity in normal subjects and stroke patients. Both parameters are large for normal subjects, medium for the healthy side, and small for lesion side of stroke patients. Only the difference in angle of reversal had statistical significance.

There are 3 possible reasons for the decrease in the stiffness constant. One possibility is diabetic neuropathy, which occurs more frequently in stroke patients and may decrease afferent inputs and joint stiffness. Second, it may be the result of decreased body weight or disuse atrophy. Although the mean body weights of both stroke and normal groups were similar (63.5kg, 60.3kg, respectively), the muscle mass of the 2 upper limbs may become asymmetric. Third, the decrease of stiffness constant may reflect an underestimation of inertia. We used the same anthropometric formulae for estimating the inertia of the forearm in both groups studied. The mass distribution may differ for each group.

Increased Damping Coefficient in the Affected Side of Stroke Patients

The increased damping coefficient and damping ratio in the affected side of stroke patients is the most important finding in our study, which agrees with the common concept that spasticity is mainly a velocity-dependent phenomenon. One previous study,[22] using ramp-and-hold stretches, also showed similar dependence of resistance on stretch velocity. Powers et al[23] showed that the increased resistance in spastic patients during constant velocity stretch is mainly caused by decreased threshold angle for stretch reflexes. In our study, we did not observe any corresponding point representing the threshold angle. We think the difference in experimental setup is the reason why. We used the pendulum motion as the external perturbation, which imposes neither constant velocity nor constant torque stretch on the elbow joint. The resistance is smoothed out in these kind of stretches, instead of manifesting a sharp transition, like that found with the threshold angle. We do not argue against the concept of threshold angle, but emphasize that the increased resistance in spastic elbows is multifactorial.[24] The main implication of our results is that spasticity in the elbow joint can be quantitatively evaluated with a simple apparatus and a simple indicator.

Severity of Spasticity

The SDs of parameters are relatively large on the affected side of stroke patients, indicating larger variability in this group (fig 4). There are 2 possible explanations. The first is the inherent fluctuation in the severity of spasticity. It is well known that the level of spasticity may change with the history of previous movements, concomitant movement of other parts of body, and the general physical conditions. We tried to reduce the effects of previous movements by holding the upper limb at the fixed initial position for 2 minutes and to minimize the effects of movement of other body parts by instructing the subjects to relax as much as they could. However, the fluctuation in general physical conditions cannot be easily controlled. We did not have a good indicator for this variable.

The second possibility is the grouping of patients with variable severities of spasticity. The stroke patients in our study had spasticity graded from 1 to 3 (table 1). We did not recruit patients with grade 4 spasticity, because the spasticity becomes so strong that we would have needed to add more weight to the accessory apparatus to perform the test. Figure 6 shows the relationship between the severity of spasticity, graded with the MAS, and the damping coefficient (C) and damping ratio (η). There is a tendency for both C and η to increase with the grade. The fitness of linear regression was modest (r=.80 for C, r=.78 for η). More cases are needed to firmly establish the relationship.

Choice of Parameters as Indicators of Spasticity

Both the damping coefficient and the damping ratio increase with spasticity. The damping ratio incorporates the effect of inertia. Because the mean body weights were similar in the 2

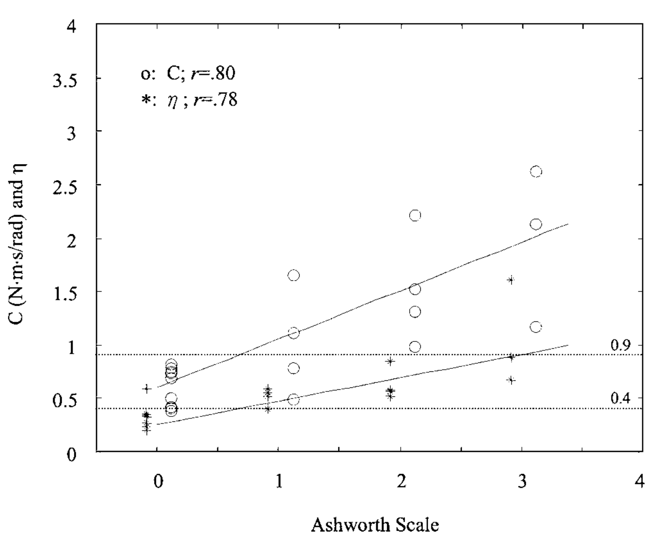

Fig 6. The relationship of Ashworth grade to C and η in stroke patients. Both C and η increased with Ashworth grade. The solid lines represent the results of linear regression. The dashed lines are the suggested cut levels, respectively, of C and η for indicating spasticity.

groups, it was expected that the results of the damping coefficient and ratio would be similar. In reality, although the mean results were similar, the conversion made the differentiation between the intact and affected sides marginally better (fig 6). If we take a damping coefficient greater than 0.9 as the cutpoint for spasticity, all the cases except 2 elbows with grade I spasticity are compatible with the prediction. With a damping ratio greater than 0.4 as the cutpoint for spasticity, all the cases except 1 control elbow were compatible with the prediction. Therefore, we recommend using the damping ratio as an indicator of muscle tone in the upper limb. The time needed for an average personal computer to do optimization calculation is less than 30 seconds in most cases.

CONCLUSION

The results our current study showed that the custom-designed accessory apparatus facilitates performing the pendulum test in the elbow joints. The damping ratio derived from a simple model can be used as an indicator of spasticity. It is possible to perform the test and to automate the analysis procedures for broad clinical applications.

APPENDIX 1: OPTIMIZATION ALGORITHM

The parameter estimation problem was formulated as a constrained optimization problem, that is, to minimize the performance measure J(z) under the constraints $g(z) \leq 0$, where

$$J(z) = \sum_{i=1}^{N} \sqrt{(\theta[1] - \hat{\theta}[1,z])^2/N}\,, \tag{A1}$$

the time series $\{\hat{\theta}\}$ was the simulated data using the design vector z, and the time series $\{\theta\}$ was the experimentally acquired data. The design vector z was chosen as:

$$z = [\theta_e \ K \ C]\,, \tag{A2}$$

and the constraint vector g(z) was:

$$g(z) = -[\theta_e \ K \ C]\,. \tag{A3}$$

The purpose of the constraints was to make the model stable, because negative stiffness or damping would turn the model into an energy-generating system. The Kuhn-Tucker conditions for constrained optimization problem could be written as:

$$\nabla J(z^*) + \sum_{i=1}^{6} \lambda_i^* \nabla g_i(z^*) = 0$$

$$\nabla g_i(z^*) = 0, \quad \text{where } i = 1,2,\ldots,m_e$$

$$\lambda_i^* \geq 0, \quad \text{where } i = m_e + 1,\ldots,6 \tag{A4}$$

where λ is the Lagrange multiplier. The sequential quadratic programming (SQP) method was employed to solve the problem.

To estimate the optimal parameters (z^*), an initial set of z values was chosen simply by guessing or based on previous simulation results. The angle trajectory $\{\hat{\theta}(i,z)\}$ was calculated from equations 1 to 4 in the main text by using parameter vector z and applying the Runge-Kutta method for integrating the equation of motion. In the next step, $\{\hat{\theta}(i,z)\}$ was fed to equation A1 to obtain J(z). Then, the SQP method was used to solve the constrained optimization for a better estimate of z in the current iteration. The refined z was used to calculate $\{\hat{\theta}(i,z)\}$ again in the next iteration. In this manner, $\{\hat{\theta}(i,z)\}$ and z were calculated recursively, until the final optimal z^* and $\hat{\theta}(i,z)^*$ were obtained.

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