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Precise Measurement System for Knee Joint Motion During the Pendulum Test Using Two Linear Accelerometers

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1. Introduction

The pendulum test is a means to evaluate the knee joint reflex from the pendulum motion induced by letting the lower leg drop freely after it has been lifted up (Watenberg, 1951). Many researchers have attempted to quantify the spinal cord stretch reflex from this pendulum motion in order to diagnose spasticity (Fowler et al., 2000; Kaeser et al., 1998; Lin & Rymer, 1991; Nordmark & Andersson, 2002; Stillman & McMeeken, 1995; Vodovnik et al., 1984). However, even today, much remains unknown about the relationship between this pendulum motion and the mechanism that produces the stretch reflex. For this reason, quantification studies on the stretch reflex have progressed slowly.

One method to advance the quantification of the stretch reflex may be to implement the following items in order:

1. Analyze the unknown behaviors in the pendulum test from various view points by trial and error, using existing physiological, clinical, and control engineering knowledge and theory as appropriate.
2. Modify the existing pendulum test model (Jikuya et al., 1991) based on the results of 1.
3. Elucidate the detailed mechanism of the stretch reflex using the model in 2, and investigate quantification methods.

We have already elucidated various phenomena following this procedure, but in this process, it has often been necessary to know angle, angular velocity, and angular acceleration values at arbitrary times during knee joint motion as initial and boundary conditions to solve nonlinear differential equations. Obtaining this kind of waveform with existing simple methods is difficult, as described below.

In principle, various existing sensors can be used to detect knee joint motion. However, several such sensors are not practical because of the knee joint's unique structural complexity. In addition, all existing sensors can measure only one of angle, angular velocity, or angular acceleration. Because of this, the only method that we can produce more than one type of waveform using such sensors is to differentiate and integrate the measured waveforms. As a result, it is difficult to ensure sufficient amplitude accuracy for waveforms obtained in this way and precise synchronization with measured waveforms.

For these reasons, we have recently begun investigating sensors that are suitable for the pendulum test. We have developed a new sensor that can precisely measure knee joint motion using two linear accelerometers. This article provides a comprehensive description of this sensor and related matters.

Section 2 briefly explains basic matters related to the pendulum test, such as the skeletal structure of the knee joint and the kinesiology of the stretch reflex. section 3 explains the measurement principle, assessment of accuracy in the laboratory, and the precision estimates when measuring subjects with the knee joint motion measurement system that is the main topic of this article. section 4 examines the results with the knee joint motion measurement system using these sensors; that is, the angle waveform and angular acceleration waveform of the knee joint in the pendulum test. We then touch briefly on a pendulum test simulator and an inverse simulation of measured waveform to more effectively utilize the results of the measurements, including the future outlook. section 5 provides a brief summary.

2. Biomechanics of the knee joint

2.1 Structure of the knee joint

The general motion of the knee joint is flexion and extension in the sagittal plane, caused consciously (actively) or unconsciously (passively). The leg structure that contributes to this motion is shown in Fig. 1.

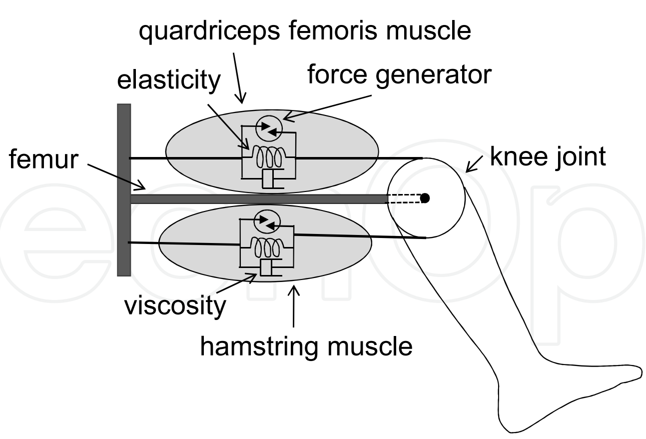


Fig. 1. Mechanism of extension and flexion.

The disc located at the end of the femur represents the knee joint, and the center of the disc is the rotation axis of the knee joint. The lower leg is fixed to the disc. The upper and lower ellipses are the quadriceps femoris muscle and hamstring muscle, respectively. One end of each muscle is fixed on the circumference of the disc. The spring and dashpot drawn in each of these ellipses are the respective elasticity and viscosity of the muscle. The force generator is the source that generates muscle contraction. The efferent fiber that controls it is not drawn. The knee joint oscillatory system consists of elasticity, viscosity, muscle contraction, and lower leg mass.

The above-mentioned quadriceps femoris muscle and hamstring muscle are the agonist and antagonist, respectively. When contractile force occurs in the agonist, the agonist shortens regardless of whether it is triggered consciously or unconsciously (when it is conscious, the antagonist also extends simultaneously), and consequently the knee extends. Similarly, the knee flexes when contractile force occurs in the antagonist. When conscious contractile force disappears or external forces that flex or extend the knee passively are eliminated, the lower leg will subsequently have damped oscillation with repeated flexion and extension unless it is resting in a stable position. In the following, we call such a dumped oscillation free one.

Next, let us look at the movement of the knee joint rotation axis. In general the knee joint is classified as a uniaxial joint that performs flexion and extension movement, but strictly speaking its rotation axis, as described below, moves according to a complex mechanism in which the lower end of the femur slides while rolling along the top of the tibia (Kapandji, 1970). That is, though the position of the knee joint rotation axis seems as if it is fixed to the center of the disk, it slightly moves together with flexion or extension. The rotation axis that moves based on this kind of phenomenon is called the axis of motion.

The skeletal structure of the knee joint is shown in Fig. 2(a). The axis of motion during flexion and extension corresponds to the imaginary point where the collateral ligament and cruciate ligament intersect (shown with a black dot (●)). Fig. 2(b) shows the migration of the intersection. The uppermost and lowermost black dots are the positions of the axis of motion in full extension and full flexion, respectively. When the knee joint rotates from full extension toward flexion, the condyle of the femur moves by rolling only up to a certain angle, beyond which an element of incremental sliding begins to apply. At the vicinity of the maximum flexion, there is only sliding movement. The relationship between the amount of movement and the angle of the knee joint is therefore mechanically complex, and analyzing it quantitatively is not an easy task. Similarly, neither the position nor the trajectory is easy to estimate by any simple means.

For the above reasons, unlike the elbow and other joints, it is not easy to measure exactly the knee joint motion in the pendulum test.

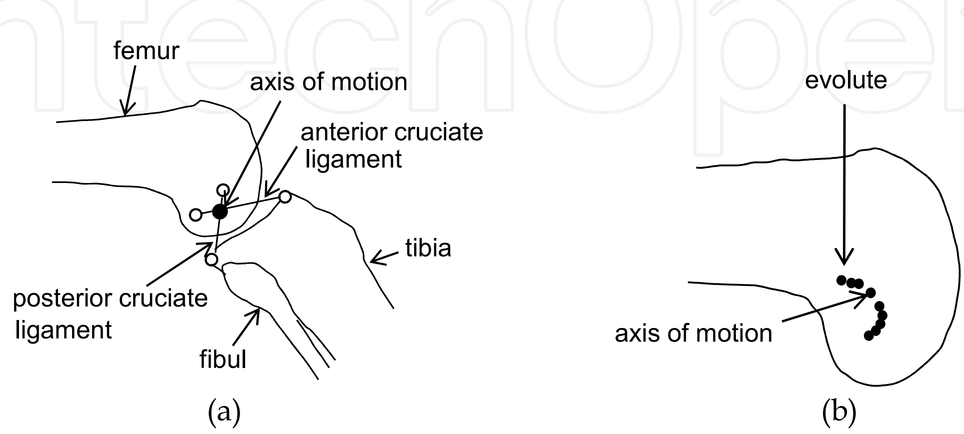


Fig. 2. Skeletal structure and locus of axis of motion of knee joint. (a) Skeletal structure of knee joint; (b) Locus of axis of motion.

2.2 Physiology of the stretch reflex

2.2.1 Principle of the stretch reflex

When muscle is stretched, it reflexively contracts in response. This kind of muscle response is called the stretch reflex. The stretch reflex is the target of the pendulum test. Fig. 3 shows the conceptual pathway of the stretch reflex. When muscle is stretched by some factor, receptor (called muscle spindle) detects it as a stimulus and transmit it as an afferent signal up to the spinal cord i.e., the α-motoneuron. The spinal cord receives the signal and sends a command (efferent signal) to effector (muscle) to restore this stretched state to the original state. These processes are executed unconsciously. Afferent fiber and efferent fiber function respectively as the transmission pathways for the afferent signal and efferent signal, which are both transmitted as impulses.

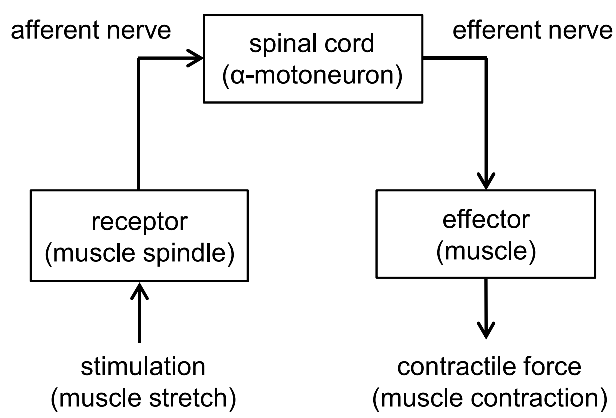


Fig. 3. Path of stretch reflex.

The reflex pathway described above (receptor → spinal cord → effector) is called a reflex arc. The stretch reflex is generated not only by passive muscle stretching, but also in response to conscious stretching of a muscle. The pendulum test is a test to estimate the sensitivity with which the reflex arc responds to the stimulus of knee flexion (extension of the quadriceps femoris muscle). The knee joint motion in this case is induced unconsciously by adding external force with the subject in a resting state for ease of estimation.

2.2.2 Structure of the spindle and its functions

Muscle is made up of many extrafusal muscle fibers arranged in parallel. Both ends of each muscle spindle are attached to one of these extrafusal muscle fibers. The muscle spindle is covered with a capsule, as shown in Fig. 4. In the capsule, there exist two types of intrafusal muscle fibers, called nuclear bag intrafusal muscle fiber and nuclear chain intrafusal muscle fiber. Stretching of the extrafusal muscle fiber affects the nuclear bag intrafusal muscle fiber and nuclear chain intrafusal muscle fiber, and stretch velocity and displacement, respectively, are detected. The detection sensitivities of the stretch velocity and displacement are regulated by efferent commands that are sent from phasic γ-motoneuron and tonic γ-motoneuron present in the spinal cord, respectively. The two kinds of detected information are consolidated into the afferent signal within the muscle spindle and transmitted to the spinal cord through Group Ia afferent fiber. Group II afferent fiber that send only nuclear chain intrafusal muscle fiber information to the spinal cord is also present, but they have a little influence on the stretch reflex in the pendulum test, and so it is not shown in the figure.

The afferent signal of Group Ia fiber is given as follows as the impulse frequency f_s (primary approximation) (Harvey & Matthews, 1961).

$$f_{\mathrm{s}} = k_{1\mathrm{s}}x + k_{2\mathrm{s}}f_{\gamma\mathrm{s}} + k_{1\mathrm{d}}\dot{x} + k_{2\mathrm{d}}f_{\gamma\mathrm{d}} \tag{1}$$

Here, x is extrafusal muscle fiber (muscle) displacement, $f_{\gamma d}$ and $f_{\gamma s}$ are the respective impulse frequencies from the brain to phasic and tonic γ-motoneurons, and k_{1s}, k_{1d}, k_{2s}, and k_{2d} are constants. As shown in the above equation, there are two types of components in stimuli detected by the muscle spindle in the stretch reflex: a stretch velocity component expressed by the first and second terms, and a muscle displacement component expressed by the third and fourth terms.

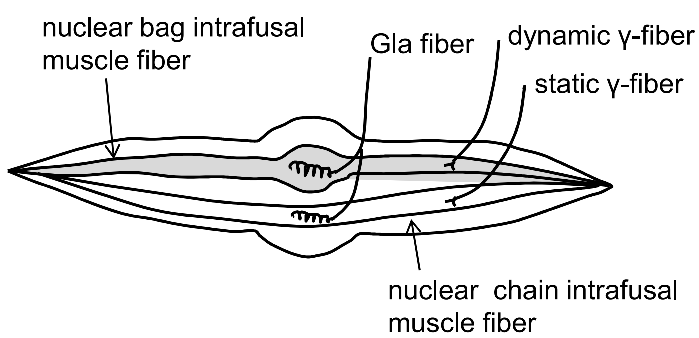


Fig. 4. Structure of muscle spindle.

2.2.3 Phasic stretch reflex and tonic stretch reflex

Commands to control the lower extremities are transmitted from the brain to muscle via the spinal cord. They are broadly divided into commands for flexion and extension, commands for maintaining of posture and commands for adjusting of the muscle spindle sensitivity. The first commands are generated only consciously, the second and third ones are generated consciously and/or unconsciously. Measurements of knee joint motion in the pendulum test are however done under the unconscious state of the subjects, and so the commands in this case are only unconscious ones to maintain posture and adjust the muscle spindle sensitivity. Consequently, the presence or absence of the efferent command toward the muscle and its strength during the pendulum test are determined only by these unconscious commands.

Fig. 5 shows the reflex arcs in the pendulum test schematically with a focus on the quadriceps femoris muscle. It includes phasic γ-motoneuron, tonic γ-motoneuron and α-motoneuron that play principal roles in the stretch reflex. The upper part enclosed by the solid line is the spinal cord. Signals f_e and f_i are commands to determine the posture, and represent frequencies of the impulses from the brain to the α-motoneuron and presynaptic inhibition part, respectively. The presynaptic inhibition part usually suppresses afferent signal from the muscle spindle so that it does not reach the α-motoneuron. Signals $f_{\gamma d}$ and $f_{\gamma s}$ are commands to adjust the muscle spindle sensitivity, and represent frequencies of the impulses from the brain to the phasic γ-motoneuron and tonic γ-motoneuron, respectively.

In normal subjects, f_e, $f_{\gamma d}$, $f_{\gamma s}$ have rather small values and f_i has rather large value, so that the α-motoneuron does not fire and no reflex occurs. Consequently, the knee joint motion at pendulum test becomes a free oscillation. On the contrary, in subjects having injuries of central nervous system, more than one of f_e, $f_{\gamma d}$, $f_{\gamma s}$ have rather large values and/or f_i has rather small value, so that the α-motoneuron fires and the stretch reflex occurs in the knee joint. Consequently, the knee joint motion is forced to disturb from free oscillation by the contractile force. In the following, we call such an oscillation forced oscillation.

The forced oscillation is classified into two types (William, 1998). One is a forced oscillation that is caused by stretch velocity component included in the afferent signal from the muscle spindle to the α-motoneuron. The value of the contractile force induced by such a component becomes maximum at the time when stretch velocity of the quadriceps femoris muscle reaches about maximum value. We call the reflex caused by such a component phasic reflex. The other is a forced oscillation that is caused by displacement component in the afferent signal from the muscle spindle. The contractile force in this case has maximum value at the time when the displacement of the muscle is about maximum. We call the reflex caused by such a component tonic reflex.

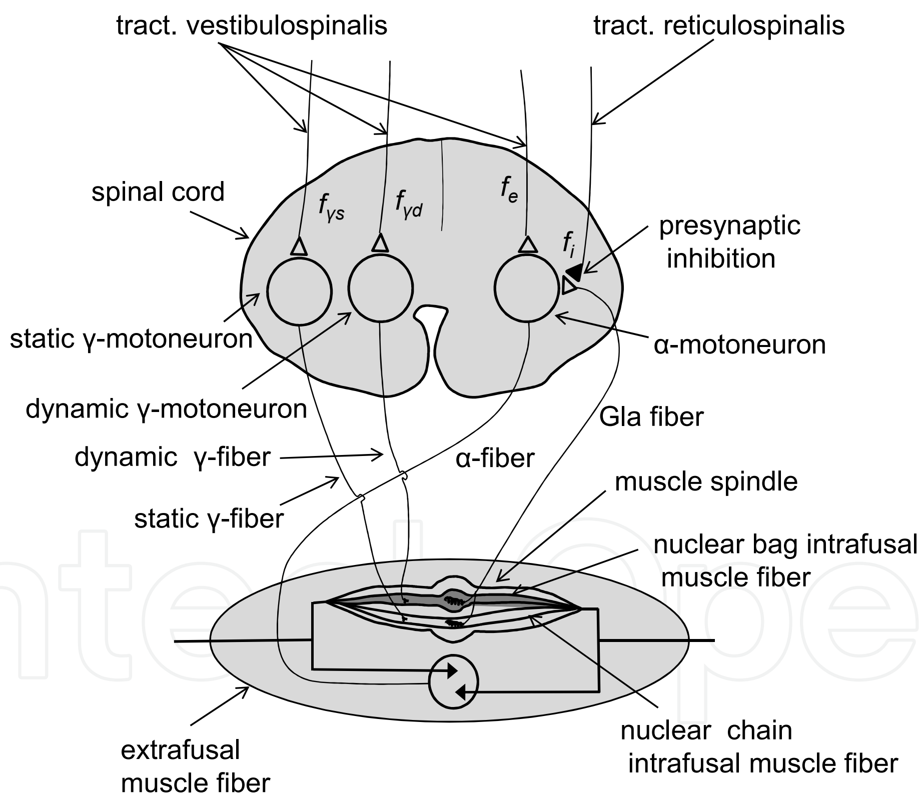


Fig. 5. Reflex arc.

As mentioned above, two types of reflexes can occur in the reflex arc composed of spindle, GIa fiber, presynaptic inhibition, α-motoneuron, α fiber and muscle. Evaluation of these reflexes therefore requires consideration of not only the size of the reflex but also the timing of their generation. Naturally, therefore, measurements of knee joint motion used in analyzing these reflexes demand high accuracy.

3. Detection of knee joint motion using acceleration sensors

3.1 Accelerometers as biosensors of knee joint motion

Measurements of physical movement have long been done focused on gait analysis. Recently, various types of advanced measurement technology are used in the field of sports science. Among them, sensors and measurement systems thought to be applicable to measurements of lower extremity motion, including sensors for pendulums described in 3.2-3.3, may be listed as follows.

a. Electrogoniometer (commonly called potentiometer)
This is a fixed rotation axis sensor that uses a rotating variable resistor. The rotation angle is detected as an electrical potential proportional to it. It has high reliability. On the other hand, it is unsuitable for measurement of high-speed movement, because large torque is required to drive contact points and they are abrasive.
b. Magnetic-type goniometer
This is a fixed rotation sensor with multiple magnetic pole and magnetic elements arranged along its circumference. It detects an electrical potential proportional to the rotation angle. It has high reliability and high accuracy. It requires a little torque since it is a non-contact-type device, and has no wearable parts.
c. Distribution constant-type electrogoniometer ("flexible goniometer")
This sensor was developed for angle measurements of complex joints (Nicol, 1989). It is not affected by movement of the joint axes, with the basic axis and movement axis set on either end of a bar-shaped resistor that changes electrical resistance with changes in shape. The angle between the two axes is measured as change in the resistance value. It has both rather large non-linearity and hysteresis.
d. Marking point measurement (or motion capture system)
Many marking points are attached to the surface of the subject's body, and images are made while the subject is moving (Fong et al., 2011). The subject is completely unrestricted. The angles at multiple points can be measured simultaneously. Its application is limited to experimental use for reasons of large filming space requirement, low time resolution, large scale of the system, etc.
e. Accelerometer
This sensor detects the movement of an object along a single axis as an acceleration signal, using a built-in strain gauge or similar element set. It is applicable to detection of accelerations in a wide range of fields, and various types have been developed from perspectives such as model type, accuracy, and stability. It does not restrict the movement of subjects, because a sensor only needs to be attached to one side of a joint even for joint movement measurements. It can also measure angle and angular acceleration simultaneously.
f. Gyroscope
Ultra-small devices have been developed using the Coriolis force and piezoelectricity based on micro-electro-mechanical systems (MEMS) technology (Tong & Granat, 1999). Currently, however, stability and reliability remain problematic.

To summarize, the requirements for knee joint motion measurement systems suitable for the pendulum test in clinical practice include: (1) sufficient accuracy; (2) low susceptibility to effects from the motion of the knee joint axis; (3) no restriction of the knee joint when worn; (4) ability to be attached simply and stably; and (5) ability to obtain waveforms of angle, angular velocity, and angular acceleration simply and with high accuracy. In the light of the above, the following conclusions may be reached with regard to the suitability of these sensors or measurement systems.

First, potentiometers are the most basic kind of angle sensor, and they have been used by Vodovnik et al. (1984), Lin & Rymer (1991), and others in studies of the pendulum test. However, when measuring knee joint angle using one potentiometer, accurate measurements cannot be made because of the axis of motion of the knee joint. Furthermore, it is not easy to attach and maintain the axis of the potentiometer in alignment with the rotation axis of the knee joint, and knee joint movement is restricted. Moreover, when seeking angular velocity and angular acceleration, one must depend on the differential, which is problematic in terms of accuracy. Magnetic goniometers perform well as angle meters, but they have the same problems as potentiometers with respect to the motion of the knee joint axis. Flexible goniometers have good properties with respect to the motion of the knee joint axis and ease of use, but the sensor itself has inadequate accuracy. Moreover, for the optical motion capture system that measures score, it is expected that the angle of knee joint motion (in some cases, angular velocity) will be detected faithfully with no contact mode, but the construction of the apparatus is too large for measurements of knee joint angle only with the body at rest, making it difficult to apply clinically. In recent years, many types of small and lightweight gyroscopes have been developed, and they have many features, such as ease of attachment, that make them suitable for measuring knee joint motion. However, stability and reliability are lacking in ultra-small types. In addition, the values detected are basically limited to angular velocity or one of the angles.

From the above, one can conclude that accelerometers fulfill nearly all of the preceding requirements, and, overall, they are the best option.

3.2 Principle of the knee joint motion measurement system using two accelerometers

We developed a method that can detect knee joint angle and angular acceleration simultaneously using two linear accelerometers in accordance with the conclusions stated in 3.1 (Kusuhara et al., 2011).

The fundamental configuration for the detection of knee joint pendulum motion is shown in Fig. 6(a). Accelerometers 1 and 2 are fixed on an accelerometer mounting bar separated by a certain distance (L_1, L_2) from point A on the rotation axis. The sensing direction of the accelerometer is the direction orthogonal to the bar on the paper. At this time, the direction of sensor attachment must be accurately fixed. However, attachment of the bar when measuring knee joint motion only needs to be fixed freely in a position within the plane of rotation of the knee joint and along the fibula as shown in Fig. 6(b). The lower leg is lifted until the bar reaches a certain angle θ (left on paper), and the pendulum motion is generated by letting the leg drop freely.

The outputs of accelerometers 1 and 2 with respect to this pendulum motion are taken as a_1 and a_2, respectively. a_1 and a_2 are given as follows.

$$\alpha_1 = L_1\ddot{\theta} + g\sin\theta \tag{2}$$

$$\alpha_2 = L_2\ddot{\theta} + g\sin\theta \tag{3}$$

Here, g is the gravity acceleration.

For both equations (2) and (3), the first term on the right side is angular acceleration from the pendulum motion, and the second term is the sensing direction component of the accelerometer, influenced by the gravity acceleration.

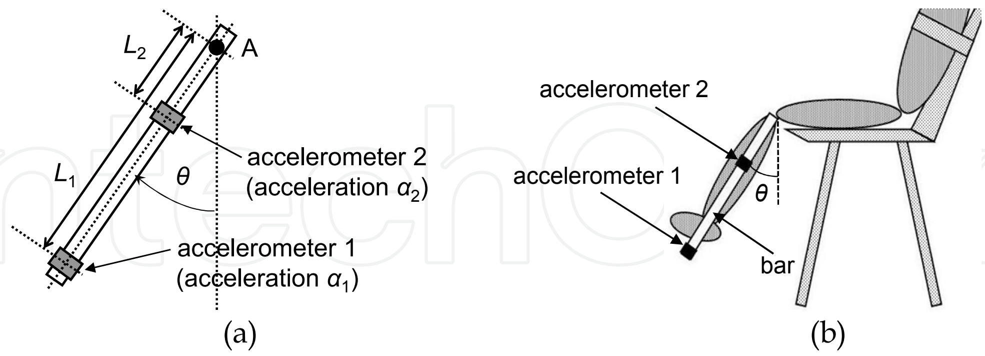


Fig. 6. Rotary motion detection by two linear accelerometers. (a) Accelerometer-bar with two linear accelerometers; (b) Attachment of accelerometer-bar on the lower leg.

When the first term on the right side disappears from both equations, the above-mentioned sensing direction component $g\sin\theta$ of the acceleration due to gravity and the angle θ of the knee joint are obtained in equations (4) and (5), respectively.

$$g\sin\theta = \frac{L_2\alpha_1 - L_1\alpha_2}{L_2 - L_1} \tag{4}$$

$$\theta = \sin^{-1}\frac{L_2\alpha_1 - L_1\alpha_2}{g(L_2 - L_1)} \tag{5}$$

When the second term on the right side disappears from equations (2) and (3), the angular acceleration $\ddot{\theta}$ of the pendulum motion unaffected by the acceleration due to gravity is given as follows.

$$\ddot{\theta} = \frac{\alpha_1 - \alpha_2}{L_1 - L_2} \tag{6}$$

In addition, for the angular velocity $\dot{\theta}$, the temporal differential of values on the right side of equation (5) and temporal integration of values on the right side of equation (6) can be obtained from the following equation.

$$\dot{\theta} = \frac{d\theta}{dt} = \int \ddot{\theta}\, dt \tag{7}$$

From the above, according to the proposed method, waveforms for angle, angular velocity, and angular acceleration that are unaffected by the acceleration due to gravity and synchronized are obtained with the addition of a single differentiation or integration.

3.3 Evaluation of the measurement system in the laboratory

3.3.1 Generation of simple pendulum motion

When evaluating the performance of the knee joint motion measurement system constructed in accordance with the principles described in 3.2, error can arise from the movement of the knee joint axis, imperfect attachment of the bar when evaluation is done, etc. This makes it difficult to accurately grasp the performance of the instrumentation body unit. In the following, therefore, we evaluate the instrumentation body unit by generating pendulum motion in a simulation.

The prototype performance evaluation system made for this purpose is shown in Fig. 7. The reference angle gauge is a high-accuracy, non-contact type, rotation angle gauge (CP-45H, Midori Precisions, Japan) used for comparison and evaluation of detector performance in the proposed method. An aluminum bar corresponds to the bar in Fig. 6(a), to which a weight is attached midway to make the period of the pendulum about the same as the lower leg. The fulcrum point A is set on the rotation axis of the reference angle gauge as in the figure for accurate comparison of the detection results. Accelerometers 1 and 2 (AS-2GA, Kyowa Electronic Instruments, Japan) are located in positions separated by only L_1 (60 cm) and L_2 (15 cm), respectively, from the rotation axis on the aluminum bar. Acceleration a_1 and a_2 detected by the accelerometers are input to a computer via matching amplifier and an A/D converter (PCD-300B, Kyowa Electronic Instruments). The output of the other rotation angle gauge is input to a computer via an A/D converter.

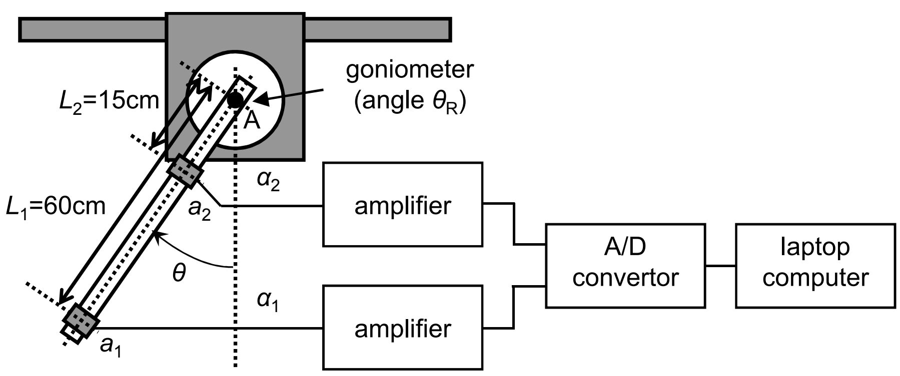


Fig. 7. Construction of performance evaluation system.

Here, the aluminum bar of the apparatus in the figure was moved as a rigid pendulum, and performance was evaluated from the results of simultaneous measurements of the pendulum motion with the detector of the proposed method and the reference angle gauge.

3.3.2 Results of evaluation

Pendulum motion was induced by freely dropping the aluminum bar after tilting it to about 40 deg. This pendulum motion had damped oscillation of a sinusoidal waveform with a period of 1.14 s, nearly the same as knee joint motion.

Output waveform examples of accelerometers a_1 and a_2 when the amplitude of damped oscillation is about 30 deg are shown in Fig. 8. a_1 and a_2 are in opposite phases because, with a_1, acceleration from pendulum motion is greatly affected by acceleration due to gravity, whereas the opposite is true with a_2.

The $g\sin\theta$ waveform obtained from equation (4) and the angle θ waveform obtained from equation (5) using these waveforms are shown in Figs. 9 and 10, respectively. In Fig. 10, the reference angle gauge output θ_R (broken line) and the $error_\theta$ between θ and θ_R (thin solid line) are added. From Figs. 8 and 9 it is seen that the values for the $g\sin\theta$ component included in a_1 and a_2 are large enough that they cannot be ignored. In addition, as understood from the example in Fig. 10, the θ and θ_R waveforms are in excellent agreement. Moreover, the component of acceleration due to gravity remaining in angle waveform θ is small enough to be indistinguishable from noise (see $error_\theta$). The correlation coefficient of θ and θ_R for 10 periods, including the 2 periods shown in this figure, was 0.999, and RMSE was 0.992 deg.

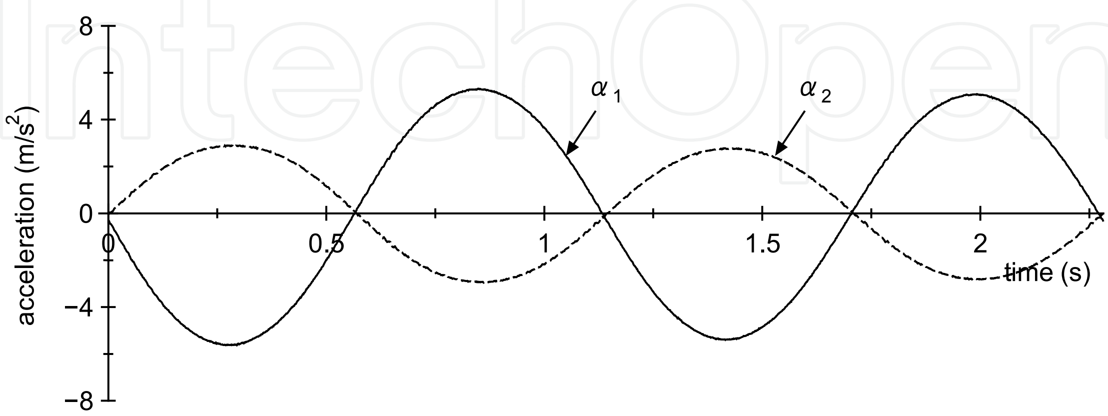


Fig. 8. Output waveforms of linear accelerometers (a_1 and a_2).

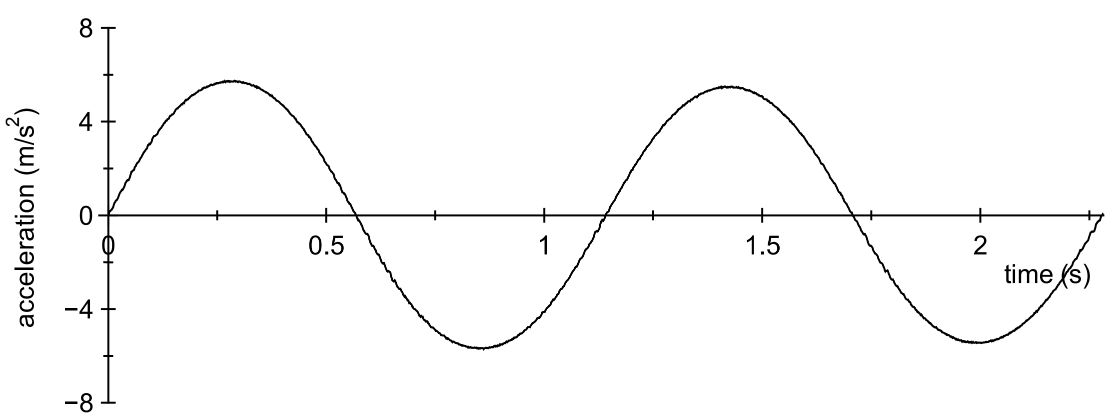


Fig. 9. Gravity acceleration components ($g\sin\theta$) in a_1 and a_2.

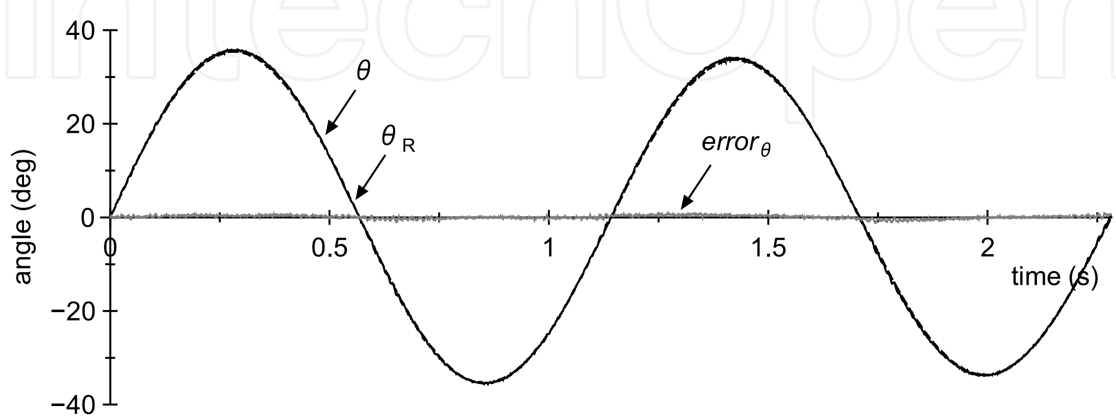


Fig. 10. Angle waveforms (θ and θ_R).

The detection results for angular acceleration $\ddot{\theta}$ were as follows. The angular acceleration $\ddot{\theta}$ obtained by substituting acceleration waveforms a_1 and a_2 from Fig. 8 into equation (6), and the angular acceleration $\ddot{\theta}_R$ obtained by twice differentiating θ_R in Fig. 10, are shown in Fig. 11. The solid and broken lines are $\ddot{\theta}$ and $\ddot{\theta}_R$, respectively, and the thin solid line is the *error* $\ddot{\theta}$ of the two. Noise is superimposed in $\ddot{\theta}_R$ obtained with a differential, but there is good agreement between the two. The correlation coefficient of $\ddot{\theta}$ and $\ddot{\theta}_R$ and RMSE for 10 periods, including the 2 periods shown in Fig. 11, was 0.998 and 0.749 rad/s^2, respectively.

Next, let us look at the angular velocity waveform. The angular velocity waveform $\dot{\theta}_1$ obtained by differentiating angle waveform (θ) in Fig. 10 (solid line) and the waveform $\dot{\theta}_2$ obtained by integrating angular acceleration waveform $\ddot{\theta}$ in Fig. 11 (broken line) are shown in Fig. 12. There is good agreement between the two, although this is due partly to the fact that these values were obtained in a simulation trial done in a laboratory with little noise.

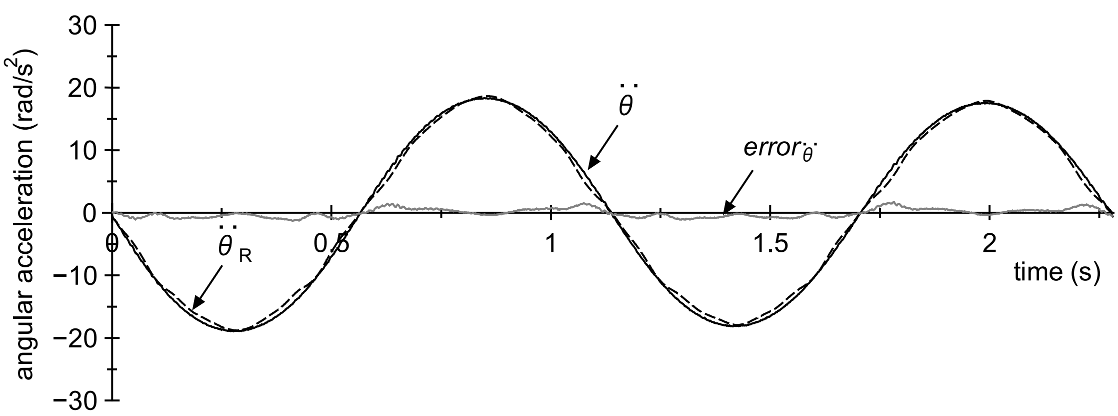


Fig. 11. Angular acceleration waveforms ($\ddot{\theta}$ and $\ddot{\theta}_R$).

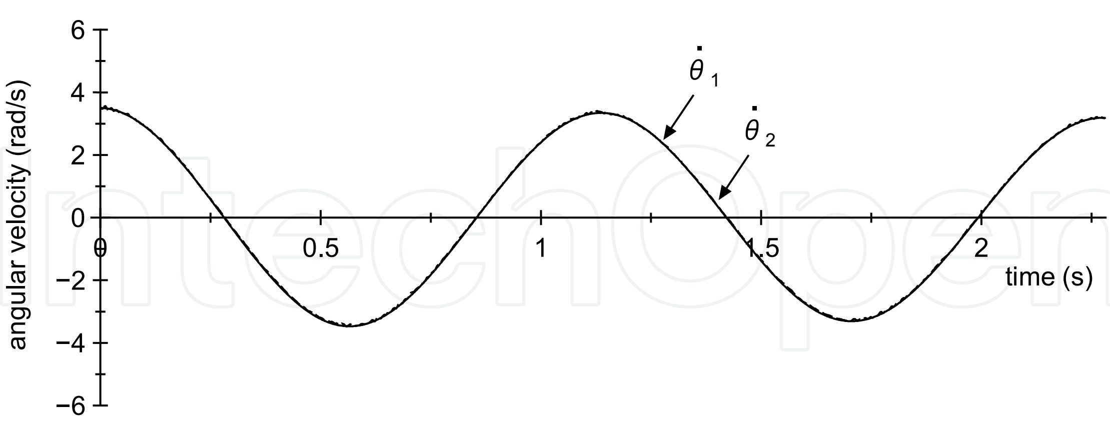


Fig. 12. Angular velocity waveform ($\dot{\theta}_1$ and $\dot{\theta}_2$).

The accuracy of the above-mentioned knee joint motion measurement system itself was obtained using two accelerometers of the same type purchased with no special conditions. This accuracy, when compared with the accuracy of detecting uniaxial arm motion with a gyroscope, goniometer, and potentiometer (correlation coefficients (0.9997-0.9999) and RMSE (1.37-1.47 deg (Furuse et al., 2005)) for similarity between these measured values), had a similar correlation coefficient and RMSE of about 30% smaller. Attaching the aluminum bar to the subjects is easier than these sensors. The next subsection discusses the effect on knee joint motion.

3.4 Estimation for the accuracy of the proposed system

For the principles given in 3.2, when using this knee joint motion measurement system created for the pendulum test, there is the problem of axis of motion mentioned in 2.1, in addition to the unique aspects of biological measurements, such as the state of attachment of the aluminum bar to the knee joint and slight changes of posture by the subject during the test. Therefore, one would predict that the decrease in accuracy due to these factors cannot be ignored. However, we have found no method that can directly and precisely evaluate the decrease in accuracy resulting from these factors. In this study, therefore, the following indirect method was used to evaluate the decrease in accuracy when measuring subjects.

First, the subject sitting in a chair for measurement and the evaluation system used in 3.3 were arranged as shown in Fig. 13. The chair and subject seen on the plane of the paper are located just in front of the evaluation system. The height of the evaluation system from the floor and the left-right positions seen on the plane of the paper were adjusted so that the rotation axis of the evaluation system and the rotation axis of the knee were on about the same line. The aluminum bar was fixed to the lower leg with rubber bands as shown in the figure so that the motion of the knee joint would be restricted as little as possible.

Next, the pendulum test was done by freely dropping the lower leg after it had been lifted about 50 deg.

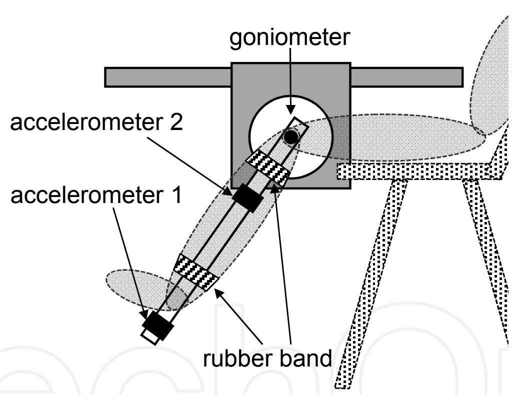
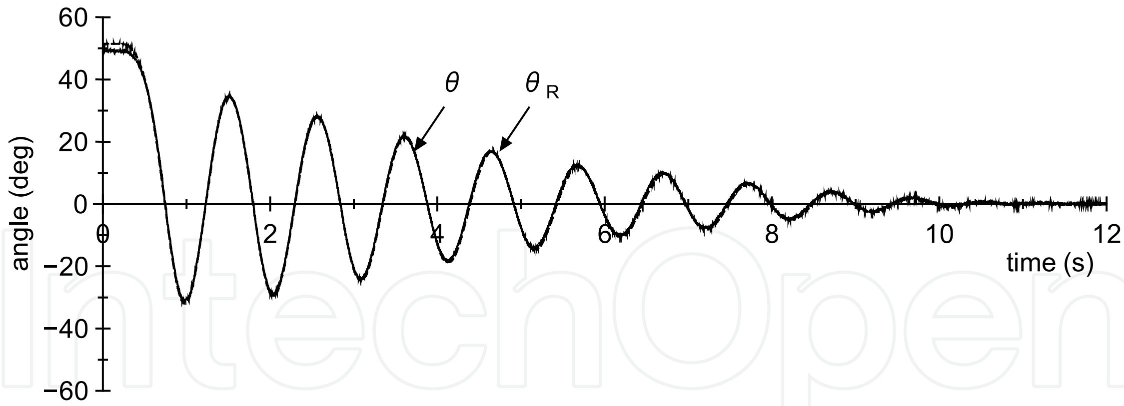
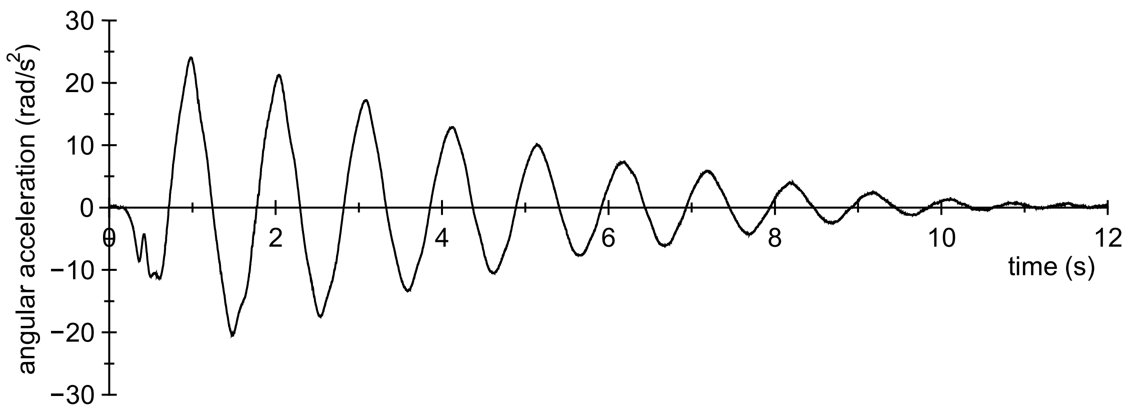


Fig. 13. Evaluation system for knee joint motion detector.

Angle waveforms obtained in this way are shown in Fig. 14 (a). θ and θ_R are the angle measured with the present method and the angle measured with the reference angle gauge, respectively. It is clear from this figure that the agreement is so close that it is difficult to distinguish the two angles. To examine the angle detection accuracy with this method in greater detail, Fig. 15 shows a window display of one section of the waveform in Fig. 14 (a). The correlation coefficient in this part was 1.000, and RMSE was 0.672 deg. The RMSE value, compared with the value for simple pendulum motion (Fig. 10), was 1.84-fold, equivalent to 1.94% with respect to the maximum amplitude (34.7 deg) of θ_R. Fig. 14 (b) shows the angular acceleration waveforms measured with this method. Good agreement between this waveform and the waveform when the reference angle gauge output was differentiated twice was also confirmed.



(a)



(b)

Fig. 14. Angle and angular acceleration waveforms measured from a normal subject by pendulum test. (a) Angle; (b) Angular acceleration.

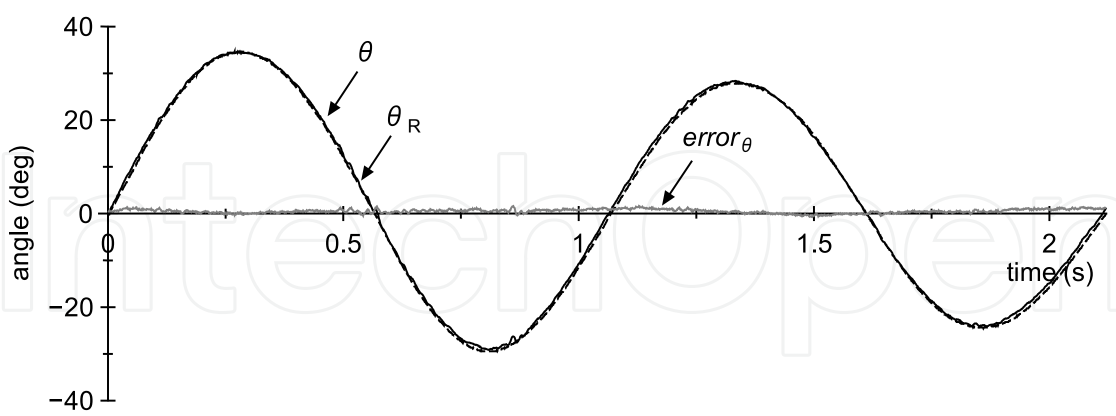


Fig. 15. A window display of the angle waveforms in the Fig.14(a).

From the above, the accuracy of the knee joint motion measurement system based on the present method was assured to be comparable to that of a reference angle gauge when applied to the pendulum test.

This knee joint motion measurement system is also thought to have sufficiently high accuracy to be applicable to the pendulum test for the following reasons.

The evaluation system used for application to the pendulum test is the same as the system used in the waveform measurements in Fig. 10. Therefore, the difference in RMSE for the waveforms in Fig. 10 and the waveforms in Fig. 15 is thought to have been produced by the difference in damped oscillation that is generated artificially and damped oscillation that occurs in the biological body, by whether or not there was positional displacement or distortion of the aluminum bar with shaking of the lower leg, or by the rotation axis movement described in 2.1. This result means that, when the fulcrum point of the aluminum bar is nearly matched to the knee joint rotation axis, both of RMSEs for the angles with this method and with the angle gauge worsen by only about 0.3 deg compared with the instrumentation body unit.

Finally, let us briefly consider the decrease in accuracy with the addition of the aluminum bar. When the aluminum bar (85 g) and two accelerometers (15 g × 2) are added, the lower leg mass of an average normal subject (about 3 kg) increases by roughly 4%. However, in the accuracy evaluation results mentioned up to this point, the descriptions have shown that there is almost no effect. Even so, the effect on measured knee joint motion in subjects cannot be ignored. When the center of the gravity of the lower leg changes with the addition of the aluminum bar and accelerometers, the moment of inertia changes in proportion to the square of the distance to the rotation axis, affecting the period of the oscillation and the damping coefficient in the pendulum motion. In both θ and $\ddot{\theta}$, the effect of the aluminum bar and two accelerometers on the knee joint motion in an average normal subject is an increase of about 3% in the time for one period and a decrease of about 6% in the damping coefficient in the results of rough trial calculations. When more precise measurements are required, the increase in the moment of inertia can be suppressed, and the influence on the period and the damping coefficient can be decreased, by shifting the attachment position of the aluminum bar upward.

4. Analysis of knee joint motion using the simulator with waveforms measured

In this section, we will deal with spastic patients as the subject of the analysis. Such patients have high phasic reflex.

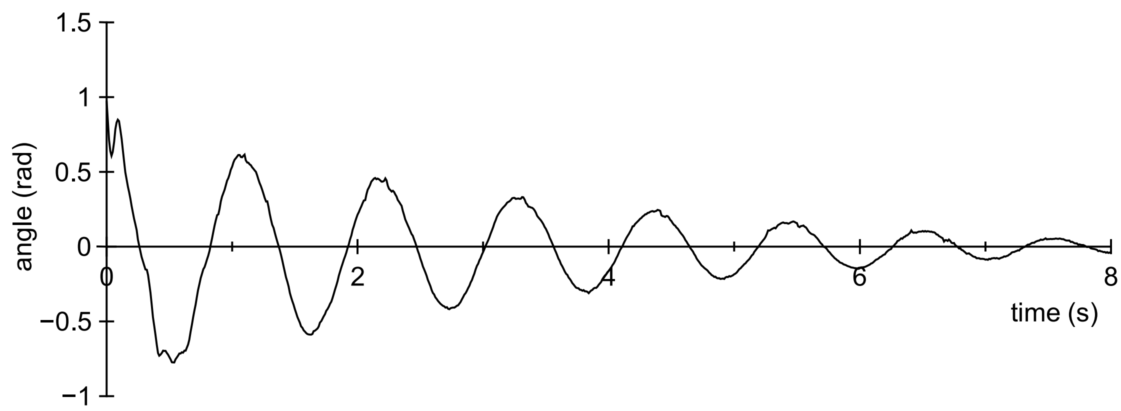
4.1 Examples of waveforms measured

The knee joint motion of a normal subject was measured in the pendulum test using the measurement system shown in Fig. 6 in section 3. Fig.16 shows examples of the waveforms measured. In the figure, (a) and (b) show the angle waveform and angular acceleration waveform, respectively. There is absolutely no restriction on motion of the lower leg, since only two accelerometers were attached to it. It is difficult to estimate the error in this measurement result quantitatively, but the measurement was probably made with about the same accuracy as obtained in the investigation in the preceding section. The collapse of the waveform that appears in the early stage of oscillation is noise produced by the state of contact between the hand of the investigator and the lower leg in the instant when the lifted leg was released. If this portion is eliminated, the angle waveform and angular acceleration waveform have typical damped oscillation with nearly the same periods, although the phases differ. These waveforms are the free oscillations mentioned in subsection 2.1. Both waveforms are described theoretically by the following differential equation derived by Vodovnik et al. (1984).

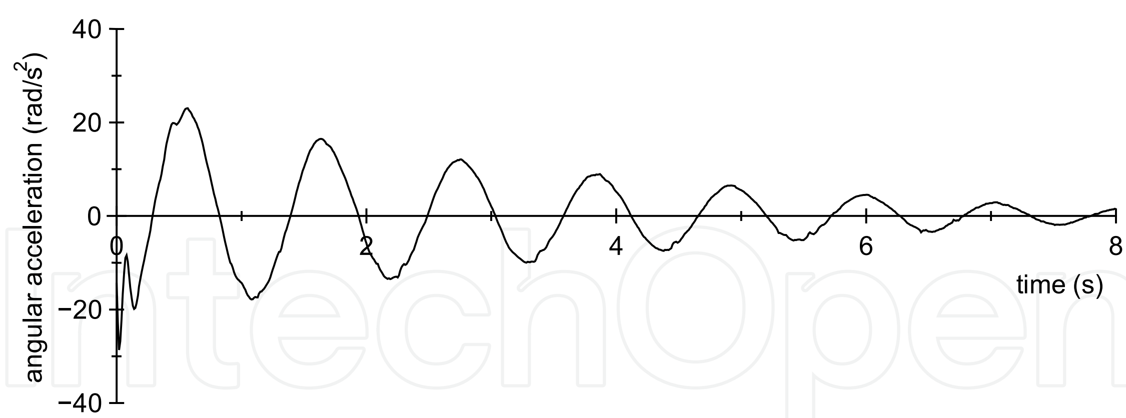
$$J\ddot{\theta} + B\dot{\theta} + K\theta + \frac{mgl}{2}\sin\theta = 0 \tag{8}$$

Where J, B, K, m, ℓ, and g are the moment of inertia, viscosity, elasticity, mass, length of the lower leg and gravity acceleration.

Even if each coefficient value is taken as a constant, this is not simple to solve analytically. However, according to the result of numerical analysis, the waveforms have similar damped oscillations as in Fig. 16.



(a)

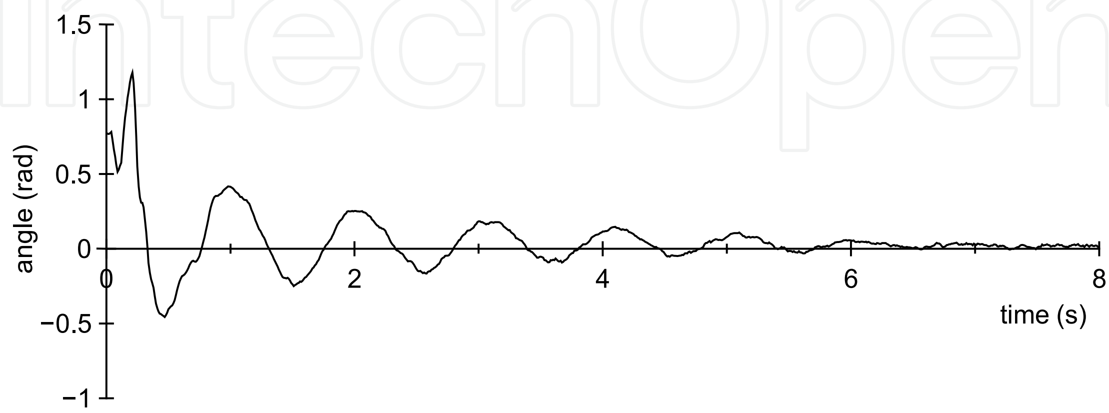


(b)

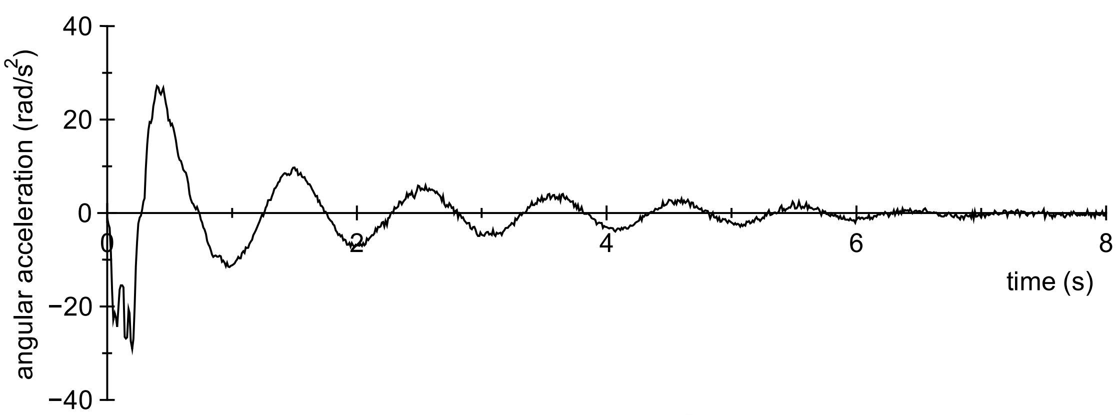
Fig. 16. Waveforms of knee-joint motion measured from a normal subject by pendulum test. (a) Angle; (b) Angular acceleration.

Next, we will look at the waveforms of spastic patients. Sample waveforms are shown in Fig. 17. The subject is a spastic patient with a moderately increased phasic reflex. Comparing them with the waveforms in Fig. 16, the first peak of the angular acceleration waveform is larger. This is because the muscle stretch velocity reaches a maximum in the vicinity where the acceleration first intersects the time axis and a phasic reflex is produced, and the resulting contractile force in the quadriceps femoris muscle acts to extend the lower leg (see 2.2.3). The knee joint motion in this case is not free oscillation, but restricted one by the contractile force. Thus, as shown in the following equation, such a knee joint motion is given with an equation that is obtained by adding this contractile force Q_h to the right side of equation (8).

$$J\ddot{\theta} + B\dot{\theta} + K\theta + \frac{mgl}{2}\sin\theta = Q_{\mathrm{h}} \tag{9}$$



(a)



(b)

Fig. 17. Waveforms of knee-joint motion measured from a moderate spastic patient by pendulum test. (a) Angle; (b) Angular acceleration.

The angle waveform and the angular acceleration waveform are accurately synchronized, because they are calculated directly from the outputs of the same linear accelerometers. Consequently, the results of measurement with the above-mentioned measurement system are suitable for the purpose of investigating behavior while rigorously referencing both waveforms.

As seen from the above examples, according to the measurement system shown in Fig. 6 of section 3, the angle and angular acceleration of the knee joint motion in the pendulum test can be measured simultaneously with high accuracy. However, investigating in detail the phasic reflex generation mechanism of each subject or estimating the degree of increase in the reflex often requires values of physical quantities and/or waveforms for various sections. Constructing a simulator for the pendulum test using the model described in the next subsection, the values of physical quantities and waveforms appropriate for an arbitrary purpose will be able to be calculated freely.

4.2 Pendulum test simulator

Since equation (9) described in 4.1 is a basic model expressing knee joint motion, a simulator can be constructed using it. However, to obtain a simulator with good accuracy, the nonlinearity of B and K needs to be considered together with a detailed mathematical formulation of Q_h. The following briefly describes these implementation methods.

Q_h is muscle contraction that occurs in the quadriceps femoris muscle during the stretch reflex, expressed by equations (10) and (11) (Jikuya et al., 2001).

$$Q_{\mathrm{h}} = \exp(-T_{\mathrm{d}}\mathrm{s})\frac{F_\alpha}{(1+T_{\mathrm{m}}\mathrm{s})^2} \tag{10}$$

$$F_\alpha = \frac{F_{\gamma\mathrm{d}} - k_{\mathrm{f}}\theta\mathrm{s}}{F_{\mathrm{i}}} + F_{\mathrm{e}} \tag{11}$$

$\exp(-T_d s)$ and $1/(1+T_m s)^2$ are transfer functions that express the sum of the transmission delay times in Group Ia fiber and α-fiber (T_d: time required for an impulse to pass through Group Ia fiber and α-fiber) and the characteristic of excitation contraction coupling (T_m: twitch contraction time of muscle), respectively. s is a Laplace operator, and k_f is a coefficient to convert knee joint angle to muscle length. F_α is the output of α-motoneuron. $F_{\gamma d}$, F_i and F_e are normalized command frequencies of $f_{\gamma d}$, f_i and $f_e - V_{th\alpha}$ ($V_{th\alpha}$: α-motoneuron threshold), respectively. Equation (11) expresses that the phasic afferent information $F_{\gamma d} - k_f\theta s$ sent from muscle spindles, after being inhibited by presynaptic inhibition F_i, is added to command F_e from the brain and becomes the output of α-motoneuron.

Next is a description of the modeling of B and K with large nonlinearity. Each extrafusal muscle fiber that make up muscle contain many actin and myosin molecules. It is known that, in a resting state, the vast majority of these molecules are in a gel state, but when the muscle starts to flex or extend movement, these molecules solate in accordance with the velocity of the flexion or extension (Lakie et al., 1984). Using the properties of these actin and myosin molecules, the temporal changes in B and K values are described by the following differential equations (Jikuya et al., 1995).

$$\dot{B} = a(B_{\mathrm{M}} - B) - b(B - B_{\mathrm{m}}) \mid \dot{\theta} \mid \tag{12}$$

$$\dot{K} = c(K_{\mathrm{M}} - K) - d(K - K_{\mathrm{m}}) \mid \dot{\theta} \mid \tag{13}$$

Here, $B_M(K_M)$ and $B_m(K_m)$ are the maximum and minimum values, respectively, of $B(K)$, and a, b, c, and d are all constants. These equations express equality of the differential of viscosity and elasticity coefficients to the value when the proportion that is solated ($b(B-B_m)\mid \dot{\theta} \mid$) is subtracted from the portion of actin and myosin molecules that is gelated ($a(B_M-B)$).

The pendulum test simulator is a program specifically for analyzing knee joint motion, prepared according to the mathematical models in equations (9)–(13) above. The inputs to the program are values of the constants that appear in these equations, and the output is angle waveform (θ), angular velocity waveform ($\dot{\theta}$), and angular acceleration waveform ($\ddot{\theta}$).We created this program in C language for execution on a personal computer.

4.3 Applications of pendulum test simulator

Using the simulator described in the previous section, knee joint motion can be freely analyzed for arbitrary combination of input values of the simulator that cannot be directly measured from subjects. In addition, by slightly modifying the program as needed, the waveforms for arbitrary section of the model can be easily analyzed. Moreover, obtaining a model for each subject by inverse simulation using the simulator, it becomes possible to analyze the knee joint motion of that subject with the model only. The following 4.3.1, shows a simple analysis of knee joint motion using the simulator, and 4.3.2 shows a case of high-level analysis of the stretch reflex using inverse simulation.

4.3.1 Examples of waveforms obtained from simulation

Figures 18, 19, 20, and 21 are waveform examples of knee joint motion in a normal subject and patients with mild, moderate, and severe spasticity, respectively. In all of the figures, (a), (b), (c) and (d) show angle waveform, angular velocity waveform, angular acceleration waveform, and muscle contraction waveform, respectively. It is clearly understood from the figures that there is a relationship between knee joint motion and muscle contraction that changes as spasticity increases. Although not shown in the figure, output waveforms such as for muscle spindles and α-motoneurons can also be analyzed simply.

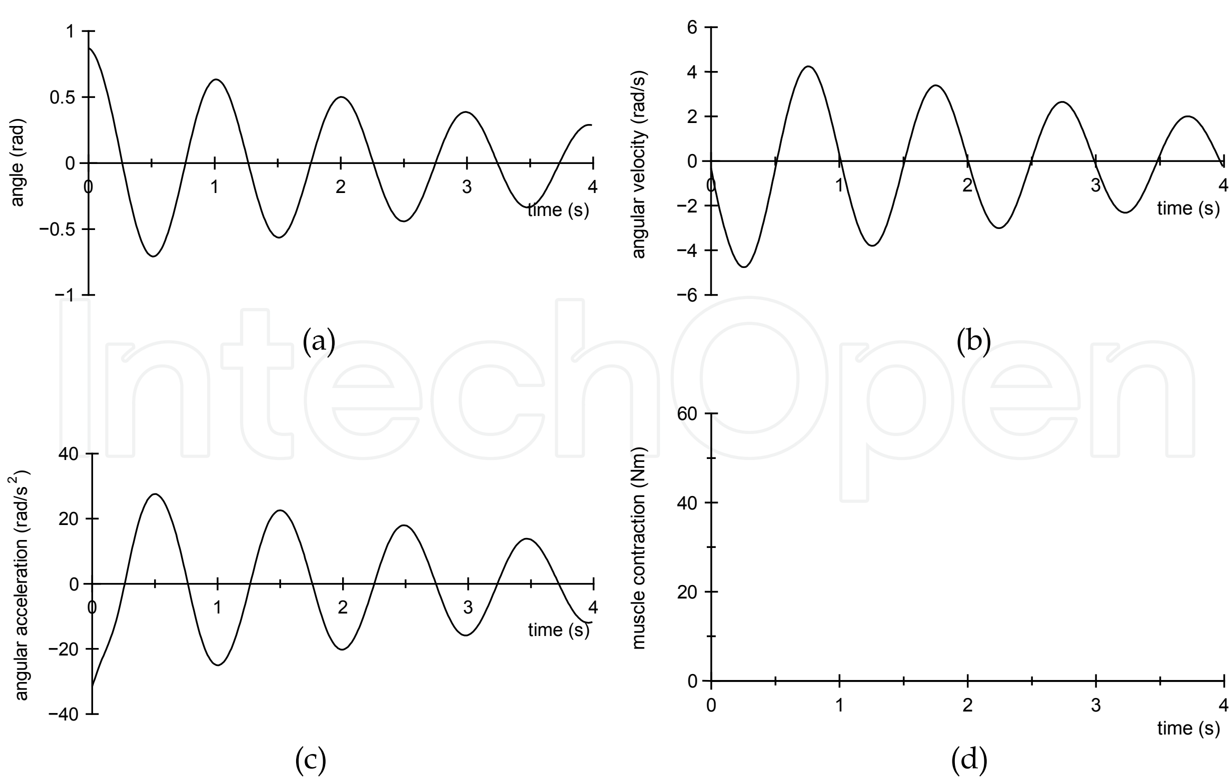


Fig. 18. Simulation result of a normal subject. (a) Angle; (b) Angular velocity; (c) Angular acceleration; (d) Muscle contraction.

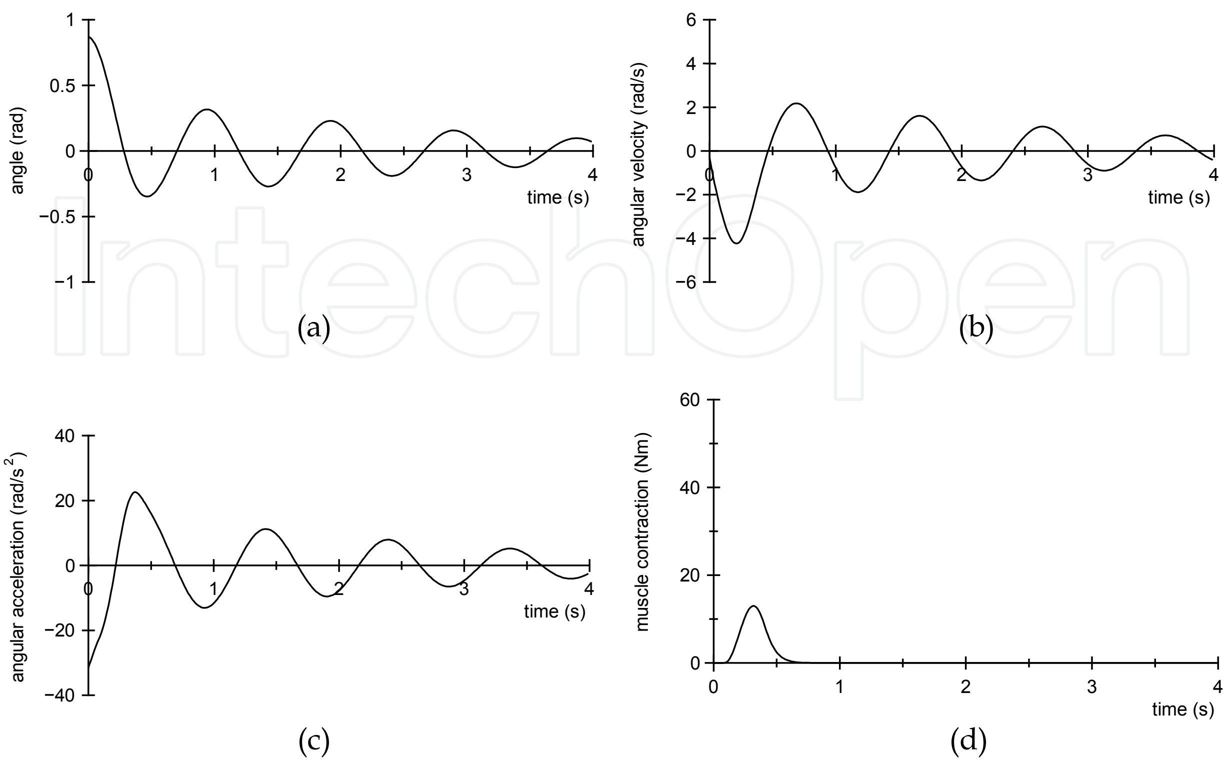


Fig. 19. Simulation result of a mild spastic patient. (a) Angle; (b) Angular velocity; (c) Angular acceleration; (d) Muscle contraction.

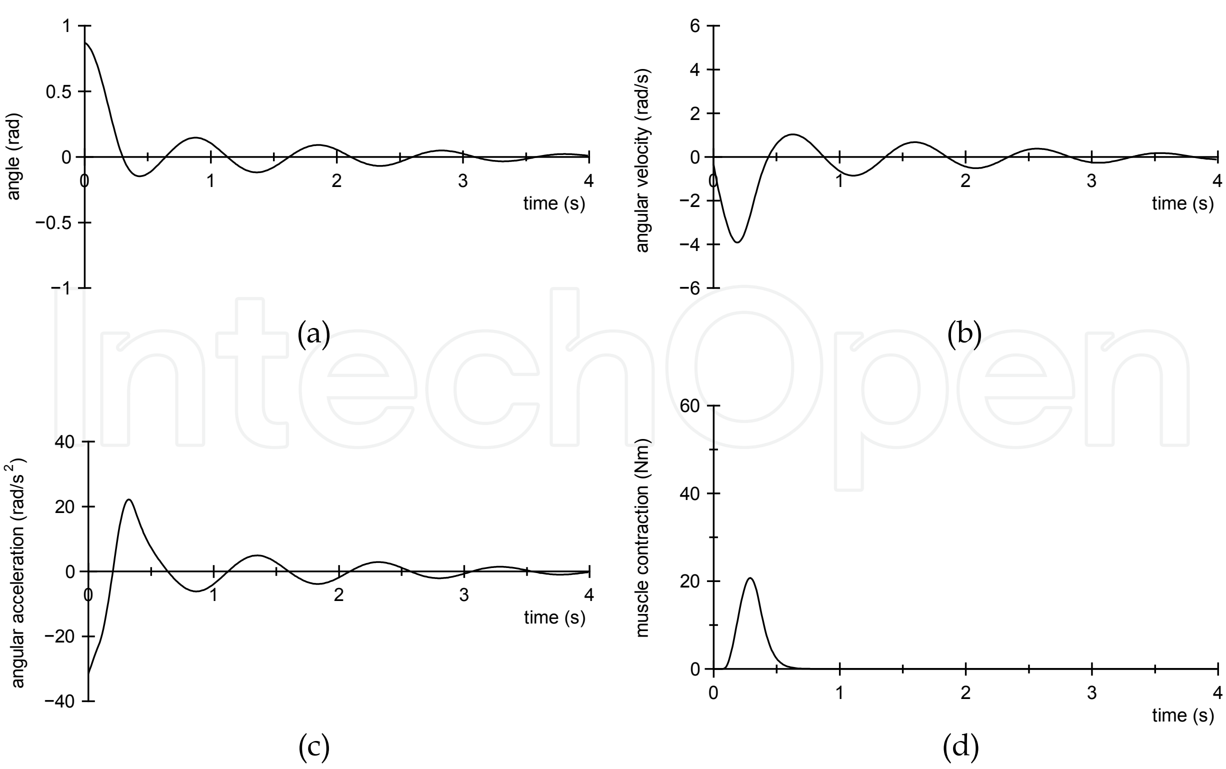
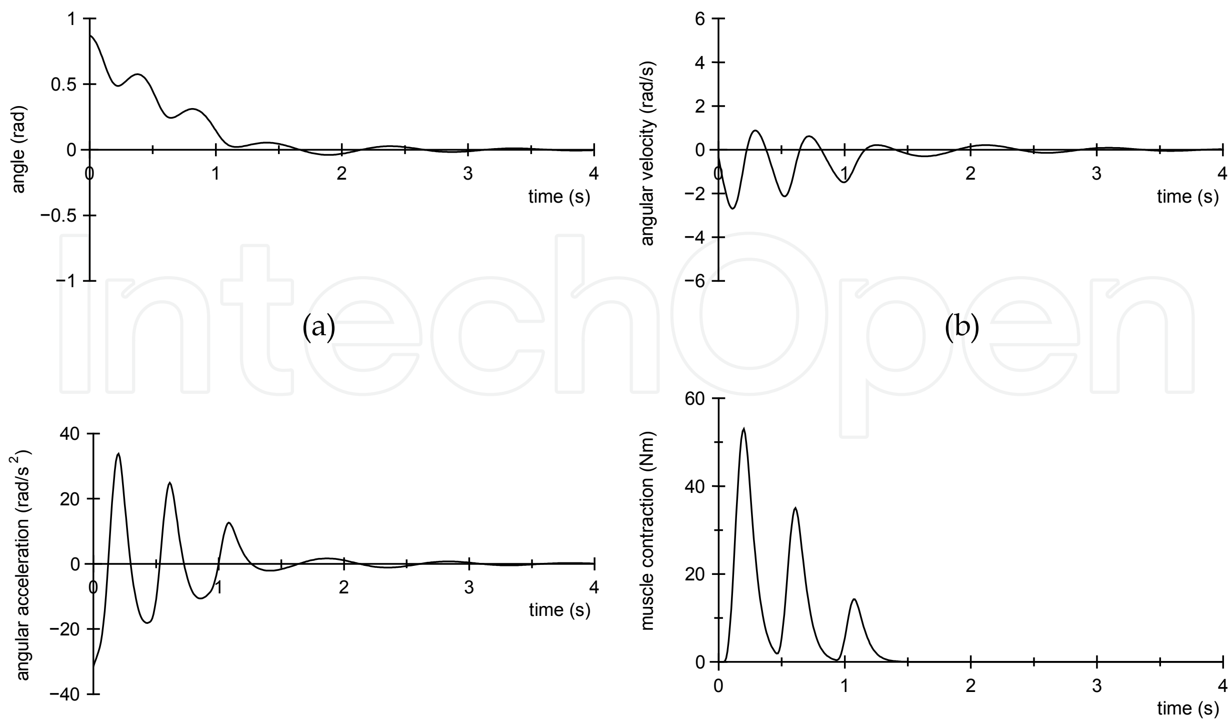


Fig. 20. Simulation result of a moderate spastic patient. (a) Angle; (b) Angular velocity; (c) Angular acceleration; (d) Muscle contraction.



(a) (b) (c) (d)

Fig. 21. Simulation result of a severe spastic patient. (a) Angle; (b) Angular velocity; (c) Angular acceleration; (d) Muscle contraction.

4.3.2 High-level analysis of the stretch reflex by inverse simulation

Determination of the input values for the simulator, so that the waveforms obtained with the pendulum test simulator agree as closely as possible with the waveforms for actual knee joint motion measurements, is called inverse simulation. If inverse simulation is conducted for knee joint motion measured with the pendulum test, the waveforms generated in the simulator, as already mentioned, are nearly the same as the measured waveforms for that subject. The constant and command frequency values at this time are values that characterize the individual subject. Therefore, if simulation is conducted based on these constant and command frequency values and waveforms and information for each part of the reflex arc are analyzed, a detailed understanding of the enhancement of the reflex for that subject can be gained.

The results of inverse simulation for one patient with spasticity are shown in Fig. 22. The solid line shows the result of actual measurement, and the broken line shows the result of simulation. There is extremely close agreement between the two results. Considering the sufficient accuracy of the knee joint motion measurement system and this kind of good agreement between the two with this method, the simulator is also assumed to have sufficient accuracy. At the current stage, however, some problems remain in aspects such as the accuracy of constant and command frequency values obtained from the inverse simulation and the time required to implement inverse simulation.

If these problems are solved, it is expected that the following issues can be resolved with application of inverse simulation.

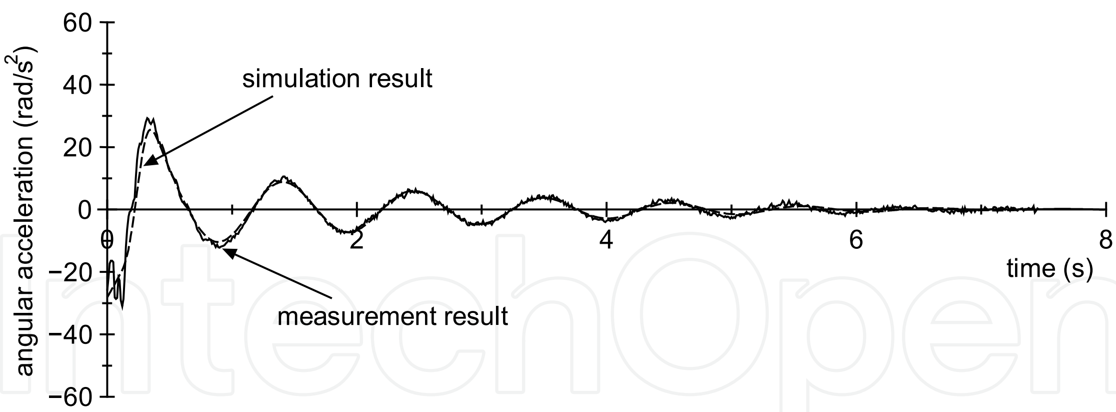


Fig. 22. Result of inverse simulation of a spastic patient.

1. Regular estimation of the status of the increase in spasticity in specific patients with $F_{\gamma d}$ and F_i values.
2. Reduction of individual differences and quantitative assessments of spasticity based on a uniform scale by standardizing inputs values of the pendulum test model for all subjects, under the premise that subject's body type and geometric structure of internal tissue are considered to be similar.
3. Verification of the effect of antispasticity drugs using this system.

5. Conclusions

This article thoroughly discusses a new knee joint motion measurement system constructed using two linear accelerometers, from the basic stretch reflex to the analysis of measurement results, focused on achievements obtained up to this time.

In section 2, we first explained the mechanical structure of knee joint flexion and extension as background knowledge to understand the discussion in section 3 and subsequent subsection, briefly touching also on the movement of the knee joint axis. Next, we looked at the source of the generation of the stretch reflex, which is the subject of measurement of the knee joint motion measurement system. We also showed the phasic and tonic reflex loops centered on muscle spindle function.

In section 3, we summarized the principles and results of performance evaluation of the knee joint motion measurement system. First, we looked comprehensively at accelerometers, which are the best option among sensors that can be used to measure knee joint motion, and then discussed the principles of the knee joint motion measurement system using two linear accelerometers. Next, we showed that the detection error of this device for simple pendulum motion and the pendulum motion of subjects is about the same as with high-accuracy, rotation angle gauges.

In section 4, we showed that the angle and angular acceleration of the knee joint could be simultaneously synchronized and measured in patients showing spasticity with enhanced phasic reflexes, and that the timing at which reflexes are produced could be easily estimated. Next, we showed the principles of a simulator to analyze measured waveforms and examples of analysis using this simulator, together with an additional statement on the outlook for high-level analysis of reflexes in patients with spasticity.

As a general conclusion, we demonstrated that the developed knee joint motion measurement system does not restrict subjects' movement as other systems do, and that it has many features that other systems do not, such as simple system configuration and the ability to acquire large amounts of information with simple data processing.

Future issues are the accumulation of clinical data using the features of this knee joint motion measurement system and quantification estimates of abnormal stretch reflexes such as spasticity, rigidity, and rigidospasticity based on those data.

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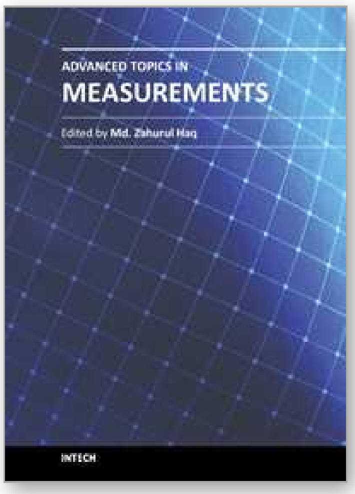

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