Table 1: Descriptive Characteristics of the Poststroke Subjects

Subjects	Age (y)	Time Since Onset (y)	Type of Stroke	Side of the Paresis	BMS (LE)	Motricity Index (UE/LE)	MMT Score (DF/PF)	Spasticity Score (ankle)
1	60	10	Ischemic	Left	4	51/53	2/3	1
2	54	4	Hemorrhagic	Left	3	40/38	1/1	1
3	58	13	Ischemic	Right	4	40/59	4/4	1
4	74	10	Hemorrhagic	Left	6	77/76	4/5	0
5	66	5	Ischemic	Left	5	65/70	4/4	0
6	51	2	Hemorrhagic	Right	5	66/76	4/5	0
7	64	6	Hemorrhagic	Right	5	73/76	5/5	0
8	75	4	Ischemic	Left	5	71/59	3/4	0
9	66	11	Ischemic	Left	3	48/48	1/1	1
10	62	13	Ischemic	Right	3	38/30	1/1	1
11	72	6	Hemorrhagic	Left	4	40/53	2/3	0
12	61	3	Hemorrhagic	Left	5	77/76	4/4	0

Abbreviations: BMS, Brunnstrom motor score; DF, ankle dorsiflexors; LE, lower extremity; MMT, manual muscle testing; PF, ankle plantarflexors; UE, upper extremity.

$$COP(t) - COM(t) = -I/mgh \times \text{COM acceleration}(t) \quad (2)$$

where COP(t) and COM(t) is the COP and the COM position with respect to the ankle joint in the horizontal directions (eg, anteroposterior [AP], ML direction), respectively (in meters), I is the moment of inertia of the body about the ankle joint in the horizontal direction (in newton meters), m is the mass of the body minus the mass of feet (in kilograms), and g is the gravitational acceleration coefficient (in meters per second squared), h is the COM height above the ankle joint (in meters), and COM acceleration(t) indicates the horizontal acceleration of COM (in meters per second squared). As such, COM acceleration is theoretically proportional to the COP-COM position when the postural control during quiet standing is characterized as an inverted pendulum model.[25] It has been experimentally confirmed that there is a high correlation between the time series of COP-COM and COM acceleration in young subjects[25,28,29] and between the standard deviations (SDs) of COP-COM and COM acceleration fluctuations in young and healthy elderly subjects.[10,11] Thus, COM acceleration can be an alternative parameter when the postural control of the healthy young and elderly is evaluated. However, it is necessary to test if COM acceleration can be applied to poststroke patients during evaluation. This is because the previously mentioned theoretic relationship between COP-COM and COM acceleration is true only under the assumption that the quiet stance can be approximated as an inverted pendulum, and changes in the posture control induced by stroke may contradict the assumption.

Although it has been confirmed that equation 2 is true in the young and elderly for postural sway in the AP direction,[10,11] it remains to be tested whether this relationship can be applied in the ML direction (eg, the body behaves as an inverted pendulum in the ML direction during quiet standing). It is also unknown if both COM acceleration and COP-COM can distinguish the postural control feature between the young and elderly in the ML direction. In healthy subjects, it has been reported that the body in the ML direction tends to behave as a 2-link inverted pendulum[30] that has an additional joint of rotation at hip and that the correlation between COM acceleration and COP-COM in the ML direction was lower than that in the AP direction.[25,28,29] Furthermore, these relationships should be tested especially in poststroke patients because they have shown a significant increase in postural sway in the ML direction and in weight-bearing asymmetry,[14-19] and such a feature can make one think that there is a possibility that the body sway behaves like a 2-link inverted pendulum in those patients. Consequently, the validity of equation 2 and the effectiveness of COM acceleration and COP-COM in distinguishing postural control feature should be tested among the young, elderly, and poststroke patients.

The purpose of this study was to confirm the following 2 hypotheses: (1) with COM acceleration, the difference in postural control characteristics among the healthy elderly, healthy young, and poststroke patients can be evaluated, and (2) COM acceleration is proportional to COP-COM in quiet standing in the ML direction in the young, elderly, and stroke patients. Thus, we could determine how useful COM acceleration in the evaluation of postural control during quiet standing is compared with COP-COM.

METHODS

Participants

Subjects with stroke. Twelve subjects with hemiparesis caused by stroke (11 men, 1 woman; mean age ± SD, 64.7±8.2y; height, 160.3±7.6cm; weight, 55.4±6.6kg) living in the community took part in this study. At the time of recruitment, all patients had already completed rehabilitative training and were therefore considered stable in their neurologic recovery. Strokes of the cerebellum were excluded. Subjects were also excluded if they had other neurologic problems such as Alzheimer's disease, Parkinson's disease, peripheral vascular disease, vestibular problems, diabetic polyneuropathy, cognitive deficits, or medical conditions that would interfere with the testing protocol. Subjects had to be able to give informed consent, to remain standing without help or support for at least 2 minutes, and to understand simple instructions. For a stroke group of 12 subjects, the period since onset of the stroke was 7.3±4.1 years (range, 2–13y); 8 subjects had left hemiplegia and the others right, and 6 subjects had ischemic stroke and the others hemorrhage. The physical examination consisted of (1) a lower-limb motor function recovery of the paretic side score according to the 6-stage defined by Brunnstrom,[31] (2) the hemiplegic limb strength measured by the Motricity Index,[32,33] (3) the strength of ankle dorsiflexors and plantarflexors by manual muscle testing,[34] and (4) the spasticity of the ankle score by Modified Ashworth Scale.[35,36] The characteristics and results of clinical examination are presented in table 1.

Healthy subjects. The control groups consisted of 22 healthy elderly subjects (10 men, 12 women; mean age ± SD,

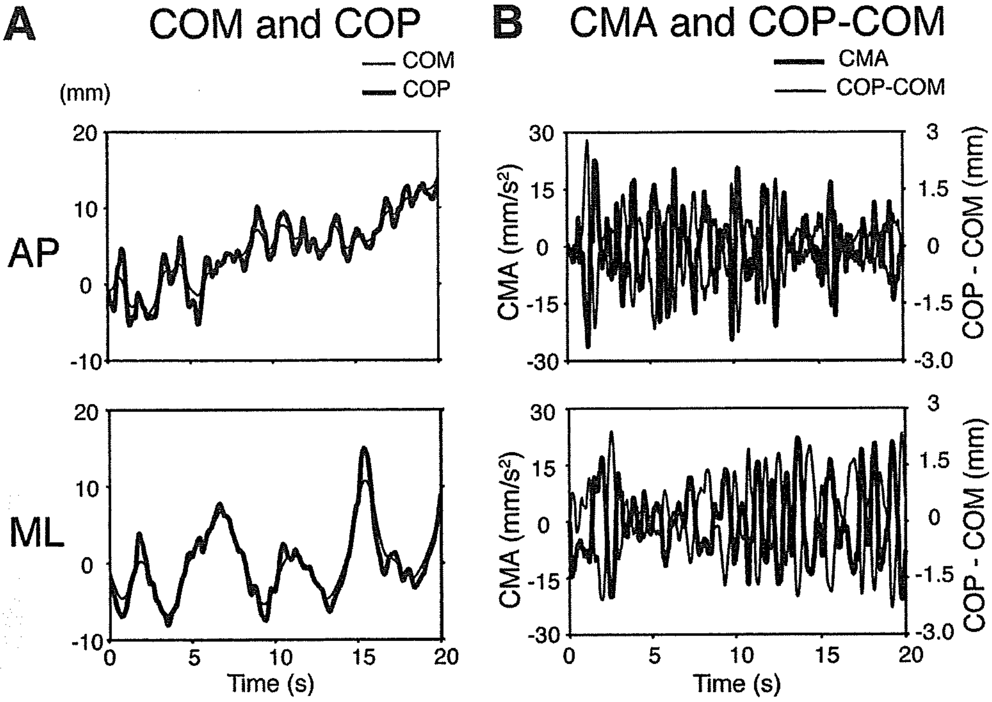

Fig 1. (A) Typical 20-second recording from a subject standing quietly in the young group for COP and COM in AP (top) and ML (bottom) directions. (B) Typical 20-second recording from a subject standing quietly in the young group for COM acceleration and COP-COM in AP (top) and ML (bottom) direction, respectively. Note that the positive value indicates forward in the AP and right leaning in the ML and the negative value indicates backward in the AP and left leaning in the ML direction of motion, respectively. Abbreviation: CMA, COM acceleration.

66.5±4.9y; height, 156.6±6.9cm; weight, 57.9±8.3kg) and 25 healthy young subjects (11 men, 14 women; mean age ± SD, 27.2±4.5y; height, 167.1±8.2cm; weight, 58.8±7.5kg) who showed interest in this study. The eligibility criteria were to be living independently in the community without musculoskeletal impairments or neurologic disorders according to self-reports. We defined the age range as greater than 20 years old and less than 40 years old for the healthy young and greater than 60 years old for the healthy elderly group. Control subjects were excluded if they reported vestibular, visual, or somatosensory impairments or if they were taking medications known to adversely affect balance.

All subjects were required to sign an informed consent document to participate in the study, and the study protocol was reviewed and accepted by the local ethics committee.

Procedures

Each subject was asked to maintain a quiet standing posture on a force platform[a] standing barefoot with eyes open, feet parallel, and 6-cm distance between their heels. All subjects performed 5 successive trials with a sufficient resting interval between trials. The duration of each trial was approximately 100 seconds, and data from the latter 90 seconds were used in subsequent analyses. During the data collection in quiet standing, the subject was also instructed to keep his/her arms hanging at the sides and to place his/her head in a normative forward-looking position. One of the investigators stood nearby to ensure safety but did not touch the subject or provide additional instructions. We collected the ground reaction forces data at a sampling frequency of 1000Hz and then converted it to a digital signal by using a 16-bit analog-to-digital converter.[b] Then, the COP displacement was calculated. The signals were thereafter low-pass filtered by using a Butterworth filter with a cutoff frequency of 10Hz. The filtering and the subsequent data analyses were performed with Matlab software.[c]

Data Analyses

The characteristics of the subject were first described by the mean and SD for continuous variables and by frequency and percentage for categoric variables. The COM acceleration variable was estimated with equation 1 as described previously. To obtain the COM displacement, we adopted the zero-to-zero-point double-integration technique method. This method was initially proposed by King and Zatsiorsky[26] and described later in more details.[27] Then, the time series of the COP-COM was calculated according to the principle of the method described by Lafond et al.[37] Note that the latter compared these methods as well as the filtering method and concluded that the 0-to-0-point double-integration technique method and the stereophotogrammetric method[25] yield similarly better estimation of the COM. We assessed the fluctuation amount of COM acceleration and the number of COP-COM variables in terms of SD for each trial in both ML and AP directions. The mean of the 5 trials of all variables was used for subsequent statistical analyses. First, all variables were screened for normality by using a Kolmogorov-Smirnov test. Then, statistical analyses were used according to the characteristics of the data. Because the data were not normally distributed, nonparametric (Kruskal-Wallis test for trigroups, Mann-Whitney *U* test for bigroup) tests were performed to compare the groups (stroke, elderly, young). The Spearman correlation coefficient was used to evaluate the correlation of the COP-COM variable with the COM variable. For the analyses, the SPSS statistical package[d] was used.

There were no significant differences among the groups for height and weight (analysis of variance, *P* range, .10–.20). Because there was no significant difference in the age between elderly group and stroke group (P=.50), the comparison between these 2 groups revealed the stroke-related postural control characteristics. No differences existed between right and left (*P* range, .10–1) hemiparetic subjects or between ischemic and hemorrhage (P range, .31–.94) hemiparetic subjects for COM acceleration or for COP-COM. Thus, the analyses were carried out without considering the effect in the type of stroke or hemiplegic side.

RESULTS

Figure 1 shows a typical time series plot of COP, COM, COM acceleration, and COP-COM, respectively, for a subject in the young group. The positive value indicates forward movement in AP and right leaning in ML directions and the negative value indicates backward movement in AP and left leaning in ML directions. Note that 20 seconds of data out of the analyzed 90 seconds are presented in the figure to isolate the signal features. Figure 1A shows COP and COM in AP (top) and ML (bottom) directions. The COP is tracking the COM in phase and is oscillating either side of the COM. Figure 1B shows COM acceleration and COP-COM in AP (top) and ML (bottom) directions, respectively. The change in the time series of COP-COM resembles that of the inverse form of COM acceleration (eg, a large or small amplitude of COM acceleration corresponds to a large or small amplitude of COP-COM with the opposite sign).

Figure 2 shows the results from the nonparametric statistical analysis comparing all variables. A Mann-Whitney U test revealed that, in the AP direction, COM acceleration and COP-COM were greater in the stroke and elderly groups compared with the young group ($P<.01$, $P<.05$, respectively), whereas differences were not significant between the elderly and stroke groups (P range, 0.56–1.00). In the ML direction, COM acceleration and COP-COM were greater in the stroke group compared with the young and elderly groups ($P<.01$, $P<.05$, respectively), whereas no significant differences were shown between the elderly and young groups (P range, 0.39–1.00).

Figure 3 shows the correlations between COM acceleration and COP-COM in each group in both AP and ML directions. The coefficients were shown to be significantly and similarly high among all groups (r range, .906–.979; $P<.001$).

DISCUSSION

In the present study, we evaluated the postural control in the healthy young, healthy elderly, and the age-matched poststroke elderly patients by using COM acceleration and COP-COM. Both variables were greater in the poststroke group than in the healthy groups in the ML direction, whereas both variables in the AP direction were greater in the poststroke group and the healthy elderly group than in the healthy young group. The results suggest that impaired postural control related to stroke can increase COM acceleration and COP-COM in the ML direction when maintaining quiet standing balance, whereas the impaired postural control related to age can increase COM acceleration and COP-COM in the AP direction when maintaining balance. Therefore, both COM acceleration and COP-COM variables can successfully extract the changes of postural control. We also found good correlations between COM acceleration and COP-COM in the ML and AP directions in all groups. These results strongly suggest that COM acceleration is applicable for the evaluation of postural control and can be as effective as COP-COM.

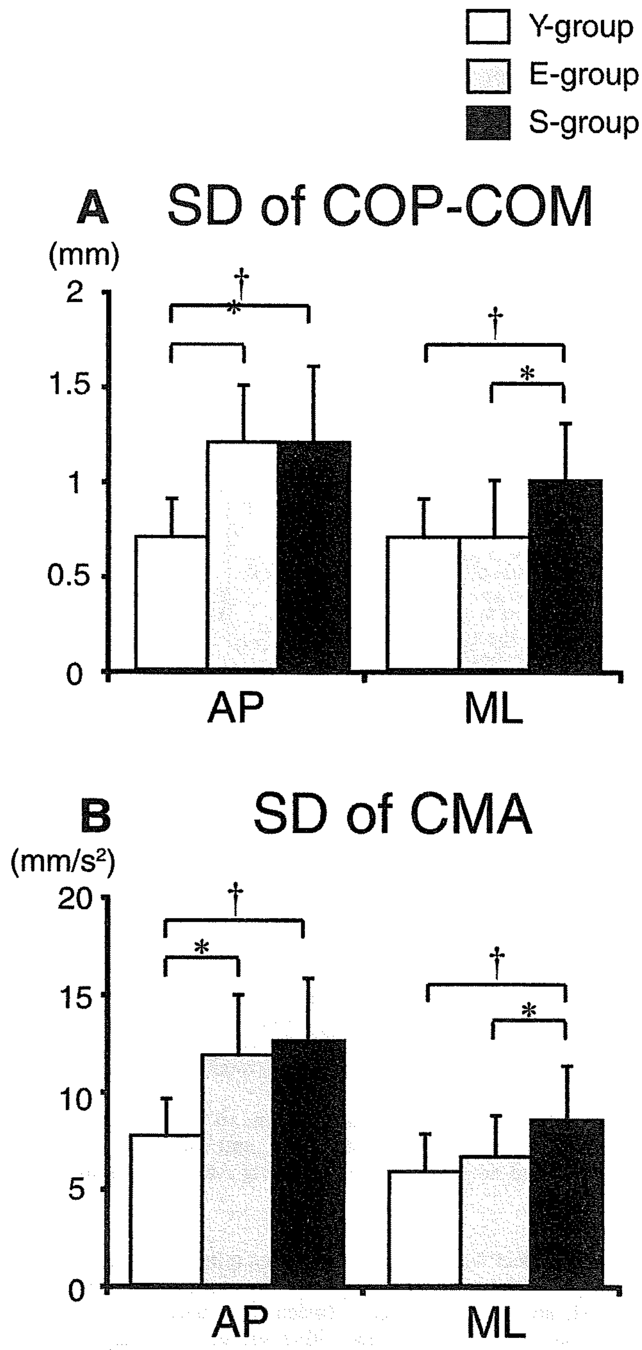

Fig 2. The results from nonparametric statistical analysis in comparison for all variables for SD of (A) COP-COM and (B) COM acceleration in the AP and ML directions. Abbreviations: E-group, elderly group; S-group, stroke group; Y-group, young group. *$P<.05$ (between groups); †$P<.01$ (between groups).

COP is proportional to the ankle torque that restores equilibrium forces of the postural control system (ie, the control variable of the system). COM is an imaginary point at which the total body mass can be assumed to be concentrated. The position of the COM has been hypothesized to be subject to body postural control, which is the controlled variable of the system. Thus, postural control can be defined by the relation between the COP and the COM. Because the relationship between the controlling and controlled variables is sensitive to changes in the control system, it is assumed that COP-COM is also sensitive to the changes in the control system. As such, the alteration of the control strategy caused by neurologic impairments and/or aging can affect COP-COM. Therefore, it has been suggested that COP-COM is a measure that captures

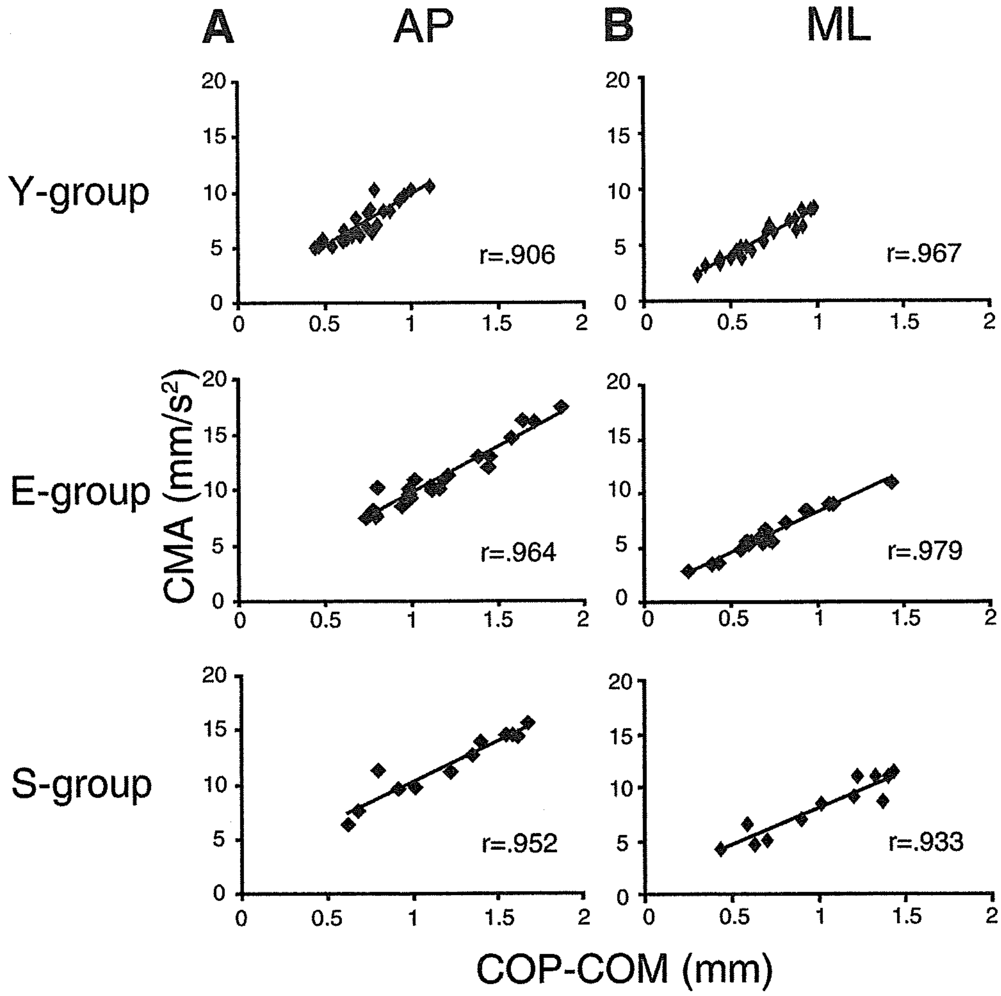

Fig 3. The relationships between COM acceleration and COP-COM in each group and in each direction. The left plots (A) show the relationships in AP direction, and the right plots (B) show those in ML direction. The top plots show the results for young group, the middle plots for elderly group, and the bottom plots for stroke group. The lines in the plots indicate the regression lines. The *r* values in the plots indicate the Spearman correlation coefficients.

characteristics of the postural control related to postural instability and the risk of falls.[22]

Previous related studies performed by using COP-based parameters (eg, COP displacement, COP velocity) also showed the changes of postural control as in this study. Marigold and Eng[19] found that people with stroke had greater COP displacement and velocity in the ML direction than those found in the healthy elderly and showed that COP velocity was related to asymmetry. Other studies[38-40] indicated that ML sway can be thought to be much more specific to the postural problems associated with stroke disease. Meanwhile, age-related studies showed an increase in postural sway[13] especially in the AP direction.[12,40] However, although these COP-based parameters have some advantages to differentiate disease-related and age-related changes in postural sway compared with COM acceleration or COP-COM, several measures only provide a parametric description of the postural sway[41] and do not reflect specific physiologic characteristics of the control system. Increasing such COP parameters as path length, area, displacement, or velocity do not necessarily link to postural instability. Subjects with high COP velocity values could reflect a stable control system in which the COP makes frequent postural correction to stabilize the COM as long as the COP does not approach the limits of the base of support.[13] In contrast, COP-COM seems to be sensitive and related to postural imbalance, and COP-COM amplitude provides an estimate of the efficacy of postural control.[42-44] Thus, we believe COM acceleration and COP-COM have a potential to assess the unique feature of the control system.

According to the inverted pendulum theory, the horizontal acceleration of the COM is proportional to COP-COM. If the COP is ahead of the COM, then the COM is being accelerated backward and vice versa. Similarly, if the COP is to the right of the COM, the COM is being accelerated to the left and to the right if the COP is to the left of the COM.[25] In other words, COM acceleration must have a close relationship with COP-COM in theory. Our results experimentally showed that COM acceleration was highly correlated to COP-COM in both AP and ML directions in all groups. Masani et al[10,11] showed the same correlation in the AP direction in the healthy young and elderly by using the same method as that used in the present study. Winter[25,29] and Gage[28] and colleagues showed that the correlation coefficient between the fluctuation amount of COM acceleration and the COP-COM time series in the young (mean age, 26y) were −.91 and −.954 in the AP direction and −.758 and −.84 in the ML direction, respectively, by using an anthropometric model in the calculation of COP-COM. Our study additionally showed that there are high correlations between COM acceleration and COP-COM in subjects of all groups in both AP and ML directions. These findings suggest that the inverted pendulum model is applicable even for poststroke patients with hemiplegia. Because COP-COM is suggested to be a valid parameter for evaluating postural sway,[21,22] it is implied that instead of the COP-COM variable postural control can be also assessed by the COM acceleration variable accurately. Compared with COP-COM, COM acceleration is easier to measure and calculate; therefore, it is an effective and practical parameter in evaluating postural control in stroke rehabilitation, especially in clinical assessment.

We tried to find a correlation in the poststroke subjects between the COM acceleration and the physical clinic scores assessed in the study (eg, the Brunnstrom motor score, Motricity Index score, manual muscle testing score, spasticity score). Except for COM acceleration in the AP direction, which had a moderate correlation with the muscle strength of the ankle dorsiflexors (Spearman ρ=–.721, P<.01), no significant correlations could be found. Although further study with a larger sample size should determine the correlations more in detail, this result might indicate that COM acceleration and physical examinations reflect the different aspects of physical impairment respectively. In our study, the poststroke subjects were in relatively stable condition, and the small sample size limited the external validity of the research. Further studies exploring stroke and other disease-related impaired postural control by using COM acceleration with larger samples are required.

CONCLUSIONS

We showed that both COM acceleration and COP-COM were able to extract the postural control features affected by aging and stroke disease. Because COP-COM reflects the relationship between the controlling and controlled variables of the balance control mechanism under the assumption of an inverted pendulum model, COP-COM can provide better insight into the assessment of postural control than COP and COM taken separately. However, it is cumbersome to calculate COP-COM, making it impractical to use in clinical assessments. We revealed that COM acceleration and COP-COM are highly proportional following the inverted pendulum theory and are valid in the ML direction and in stroke patients. Therefore, COM acceleration can be an alternative and convenient measure in the evaluation of postural control.

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Suppliers

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Effects of Passive Leg Movement on the Oxygenation Level of Lower Limb Muscle in Chronic Stroke Patients

Enkhsaihan Jigjid, MD, PhD, Noritaka Kawashima, PhD, Hisayoshi Ogata, PhD, Kimitaka Nakazawa, PhD, Masami Akai, MD, PhD, Fumio Eto, MD, PhD, and Nobuhiko Haga, MD, PhD

Objective. To evaluate the effects of passive leg movements on the muscle oxygenation level and electromyographic (EMG) activity in the lower limbs in chronic stroke patients. *Methods.* With a gait training apparatus, passive movements were imposed on the lower limbs of 15 chronic stroke patients at a frequency of 0.8 Hz for 10 minutes. During the passive leg movements, muscle oxygenation level and muscular EMG activity of the paretic and nonparetic calf muscles were assessed. *Results.* The passive leg movements caused increases in the EMG activity and muscle oxygenation level in both paretic and nonparetic lower limbs. Although a significant difference was found in the concentration changes of the oxygenated hemoglobin (Oxy-Hb), both paretic and nonparetic sides of the muscle showed enhancement of the tissue oxygenation level (TOI). The degree of the changes of the Oxy-Hb depended on the level of motor recovery after stroke; subjects with good motor recovery showed less difference in the Oxy-Hb level between the paretic and nonparetic sides of the muscle. *Conclusion.* Passive leg movements have the capacity to induce muscular activity and enhance oxygen metabolism, even in the paretic lower limb muscle of chronic stroke patients. This type of exercise might be a useful and efficient method for the prevention of metabolic deterioration in the lower limb paretic muscles of chronic stroke patients.

Key Words: *Stroke—Hemiplegia—Paretic muscle—Near infrared spectroscopy (NIRS)—Muscle oxygenation.*

From the Department of Rehabilitation Medicine, Graduate School of Medicine, University of Tokyo (EJ, FE, NH) and Department of Rehabilitation for the Movement Functions, Research Institute of the National Rehabilitation Center for Persons with Disabilities (EJ, NK, HO, KN, MA, FE), Tokyo, Japan; Institute of Biomaterials and Biomedical Engineering, University of Toronto, Toronto, Ontario, Canada (NK); and Japanese Society for Promotion of Science (NK).

Address correspondence to Noritaka Kawashima, PhD, Lyndhurst Centre, Toronto Rehabilitation Institute, 520 Sutherland Dr, Toronto, Ontario M4G 3V9, Canada. E-mail: nori.kawashima@utoronto.ca.

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Stroke is the third leading cause of mortality worldwide. Stroke survivors usually experience significant mobility impairments. Only 25% of stroke patients return to an activity level of everyday life and physical functioning compared with community matched persons who have not had a stroke.[1] In addition to the neurological changes and their numerous manifestations, metabolic changes that contribute to the poststroke complications and residual disability are evident.

Physiological deteriorations commonly occur after stroke because of the acute medical illness and associated bed rest and immobility. The decrease in physical activity leads to alteration of normal physiological processes resulting in paretic muscle atrophy and fat deposition,[2-4] reduced exercise capacity,[5] and osteoporosis.[6-8] Previous investigations have revealed that there is unilateral blood flow impairment and reduced hyperemic blood flow in the paretic limbs.[9,10] Moreover, Warlow et al[11] reported that impaired blood flow increases the risk of deep venous thrombosis in the paretic leg by a factor of 10. Because there is a strong causal correlation between blood flow and muscle metabolism,[12] hypocirculation in the inactive muscles will cause reduced metabolism in the paretic limbs.

We recently reported that passive leg movements can induce not only muscular activity, but also alteration in the muscle oxygenation level in the paralyzed lower limbs of complete paraplegics.[13,14] Enhanced muscle oxygenation is a key parameter of muscle metabolism under various conditions and is a potential factor to facilitate blood circulation and metabolism in the paralyzed area. This viewpoint can be applied to stroke patients as well. There are no studies concerning the effects of the passive movement on the paretic muscle metabolism in stroke patients. The aim of the present study was therefore to investigate the effects of passive movements on the muscle oxygenation and the electromyographic activity in the lower limb muscles of chronic stroke patients. Comparison between the paretic side and the nonparetic side enables us to identify their

Table 1. Summary of the Characteristics of the Subjects

Subject	Age	Weight (kg)	Height (cm)	Time Since Onset (y)	Stroke Type	Paretic Side	BMS	MI UE/LE	MMT (Gas + Sol)	Cadence	Spasticity (Ash. Score)	Aids	Group
S1	74	52	150	10	Hemor	Left	5–5–6	77/76	5	100	0	Cane	Good recovery
S2	66	49	157	5	Isch	Left	4–4–5	65/70	4	86	0	No	Good recovery
S3	51	58	157	2	Hemor	Right	5–5–5	66/76	5	83	0	No	Good recovery
S4	64	66	168	6	Hemor	Right	5–5–5	73/76	5	103	0	No	Good recovery
S5	75	49	148	4	Isch	Left	5–5–5	71/59	4	83	0	Cane/AFO	Good recovery
S6	61	63	162	3	Hemor	Left	5–5–5	77/76	4	72	0	Cane	Good recovery
S7	60	53	164	10	Isch	Left	3–3–4	51/53	3	54	1	Cane/AFO	Poor recovery
S8	54	56	161	4	Hemor	Left	3–2–3	40/38	1	73	1	Cane/AFO	Poor recovery
S9	58	66	168	13	Isch	Right	3–2–4	40/59	4	86	1	No	Poor recovery
S10	66	61	168	11	Isch	Left	3–3–3	48/48	1	70	1+	Cane/AFO	Poor recovery
S11	62	64	167	13	Isch	Right	2–2–3	38/30	1	NA	1+	WC/AFO	Poor recovery
S12	72	50	154	6	Hemor	Left	3–2–4	40/53	3	86	0	Cane/AFO	Poor recovery
S13	62	69	164	2	Hemor	Bilat	5–5–6	92/91	5	110	0	Cane	NA
S14	56	48	150	4	Hemor	Bilat	4–2–4	61/58	3	86	0	Cane/AFO	NA
S15	54	58	158	4	Hemor	Bilat	6–5–6	77/100	5	83	0	Cane	NA

Ish = Ischemic; Hemor = hemorrhagic; Ash. Score = Ashworth score for spasticity; MMT = muscle manual test; BMS = Brunnstrom motor score; AFO = ankle foot orthosis. For patients with bilateral motor impairment, data for the most impaired side are shown.

physiological characteristics and differences. We hypothesized that the muscle oxygenation level should change in accordance with electromyographic (EMG) activities in the paretic lower limb muscle, as was observed in persons with spinal cord injury.[14] We also hypothesized that the character of oxygenation changes during the passive leg movement is affected by the level of motor recovery after stroke. If the muscle oxygenation can be facilitated by the passive leg movement in stroke patients, it may have significant implications for the rehabilitation and especially prevention of secondary impairments following stroke.

METHODS

Participants

For this experiment we recruited 15 participants who incurred a stroke. The inclusion criteria were as follows: (1) At least 6 months poststroke; (2) absence of cardiorespiratory disease; and (3) absence of stroke complications (deep venous thrombosis, contracture or high-grade spasticity interfering with the gait). Average age of subjects was 63.1 ± 7.7 years and time since onset of the stroke was 6.9 ± 4.2 years. Twelve patients had hemispheric stroke (6 ischemic and 6 hemorrhagic) and 3 had bilateral involvement. Eight of 12 patients with hemispheric stroke suffered from residual left hemiparesis and 4 had paresis on their right side.

Before the experiment, clinical assessments were performed to identify motor and neurological recovery status of the subjects. A Brunnstrom score for motor recovery[15] and the Motricity Index[16] for the lower limbs were assigned to each patient. The degree of spasticity in the limbs was evaluated by Modified Ashworth Scale,[17] and muscle strength of soleus and gastrocnemius muscles was measured with the Daniels Manual Muscle Test.[18] For further clinical reference, each patient's anthropometric data such as height and weight were also recorded. Also, cadences during walking at comfortable speed and preferred assistive devices were

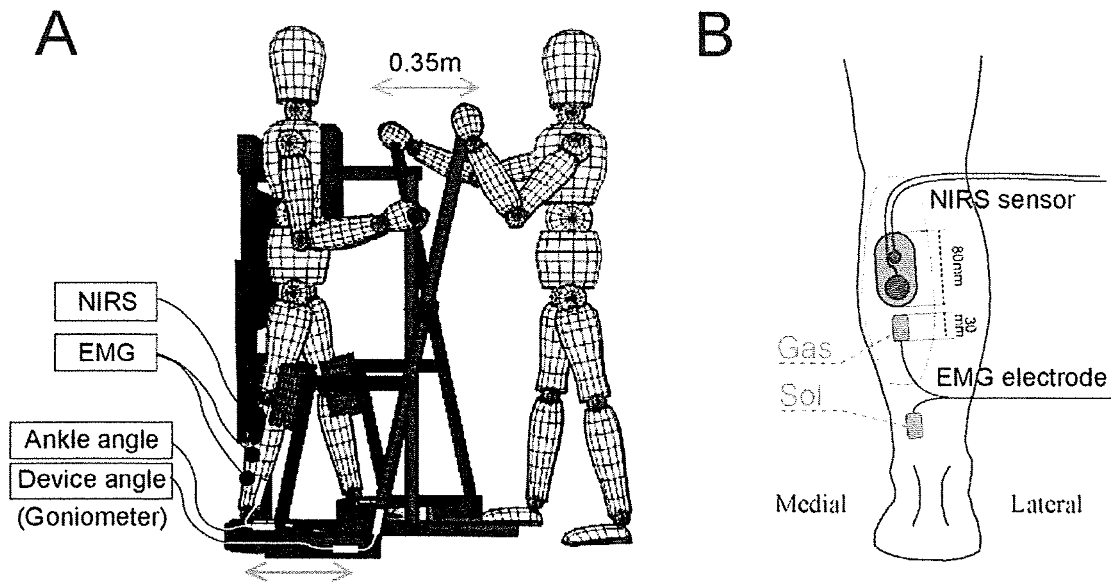


Figure 1. A, Experimental setup. This apparatus enables stroke patients to stand securely with their trunk, pelvis, and knee supported by front and back pads. By moving the handle connected to the footplate, passive leg movements can be imposed on the lower limbs of the patient. B, Placement of the EMG electrodes and NIRS sensor. Because it is not possible to place both the NIRS sensor and the EMG electrode at the same place, they were placed proximally and distally on the medial side of the muscle.

recorded. The summary of the general data and clinical evaluation is shown in Table 1.

All subjects gave their written informed consent for the experimental procedure, which was conducted in accordance with the Helsinki Declaration of 1975 and approved by the ethics committee of the research institute of the National Rehabilitation Center for Persons with Disabilities, Tokorozawa, Japan.

Experimental Procedure

A gait training apparatus (Easy Stand Glider 6000; Altimate Medical Inc, Morton, Minn) was used to impose passive leg movements, as shown in Figure 1A. This device was developed for the physical exercise of persons with movement disorders. The apparatus enables the patients to stand securely with their trunk, pelvis and knee supported by anterior and posterior support pads. By moving the handles connected to the footplates, passive movements can be imposed on the hips and ankles of patient.

The experimenter induced passive leg movements by moving the handles following the sound of a metronome set to a frequency of 0.8 Hz. Since the Easy Stand Glider imitates walking pattern (except the knee movement) in the lower limbs, we decided to choose the frequency adjusted to the lower cadence in stroke patients. The exercise was maintained for 10 minutes, after which patients remained in the standing position for 10 minutes for recording postexercise data. During the exercise and recovery period, subjects were asked to relax their body and refrain from voluntary movement and speaking. The experimenter moved the handle and simultaneously checked the angle data from the goniometer, trying to keep the leg movement within the same range of motion throughout the exercise.

Electromyography

Muscular activity of the medial head of gastrocnemius and soleus was recorded with the use of surface EMG. Sampling rate of data was set at 1000 Hz. Bipolar surface electrodes were used for recording. Because it is not possible to place both the NIRS probe and EMG electrodes on the same place, they were attached proximally and distally on the medial side of gastrocnemius muscle, as shown in Figure 1B. The electrode (DE-2.3; DelSys, Inc, Boston, Mass) was placed at least 2 cm proximal to the point of insertion of the medial head of gastrocnemius. This electrode consists of parallel bars (1 cm long and 1 mm wide) spaced 1 cm apart and is designed with a built-in bandpass filter from 20 to 450

Hz. Skin preparation (abrasion, cleaning with alcohol) was carried out carefully before recording. The obtained EMG signal was amplified by Bagnoli-8 EMG system (DelSys, Inc). Maximal voluntary contractions were also recorded for further reference.

Near Infrared Spectroscopy

NIRS is a noninvasive method for measuring tissue oxygenation and hemodynamics. First introduced by Chance et al[19] in 1985 to examine oxygenation changes in exercising muscles in humans, it has been applied to measure oxygenation in a variety of tissues including skeletal muscle. The validity and data reproducibility of NIRS for skeletal muscle has been proved in previous studies (see studies by Van Beekvelt et al[20] in 2001 and others[21,22]).

In the present study, the muscle oxygenation level was recorded using a NIRS device (NIRO-300; Hamamatsu Phototonics, Inc, Shizuoka, Japan) with dual channel laser diodes. The device can calculate changes in oxygenated and deoxygenated hemoglobin by measuring light attenuation at 775-, 813-, 850-, and 913-nm wavelengths and analyze with an algorithm incorporating the modified Beer-Lambert law. The modified Beer-Lambert law can be defined as follows:

$$\Delta A = L * \Delta \mu a$$

where A is light attenuation, L is differential path length, and μ is the absorption-scattering coefficient.[23] Changes in the Hb values were calculated relative to the preexercise data and represented in units of micromolar per liter (μM/L). Tissue oxygen index (TOI) is an absolute one-time value calculated rapidly by the device and shows the ratio of Oxy-Hb to Total-Hb. All data were digitized at 6 Hz, which is the maximal sampling rate of the device.

The NIRS probe was placed on the upper portion of the medial head of gastrocnemius muscle, as shown in Figure 1B. A calibration procedure was performed to provide an optimal range of measurement. Before the exercise, subjects were kept in a standing posture for several minutes until the level of Total-Hb reached a steady state, after which the level of oxygenation parameters were set to zero, and preexercise data were recorded for 1 minute. Then NIRS parameters during actual passive movement and following rest were recorded. To know the general tendency of the NIRS changes, we conducted a preliminary experiment for 10 of 15 stroke patients before the experiment. The NIRS changes during passive leg motion were almost the same, comparing the preliminary and final experiment. This result supports the reproducibility of the NIRS data.

Heart Rate Monitoring

To confirm whether imposed passive leg movement alters central circulation, we measured heart rate during and after exercise using an integrated telemetric monitor (S810i; Polar, Vantage, Finland).

Electrogoniometry

Goniometry (goniometer system; Biometrics Ltd, Ladysmith, Va) was conducted to ensure the same range of motion of ankle joints in both legs during the passive leg movements.

Data Analysis

All data were recorded by Power Lab software (Chart ver. 5.0.1; AD Instruments Inc, Milford, Mass) and were digitized at 1 kHz. Mean amplitudes of the EMG data were considered for analyzing muscle activity during and after the passive leg movement. To determine the degree of activation, we calculated the root mean square of the EMG signal as the most reliable parameter in the time domain.

For the NIRS data, 4 main parameters were recorded: oxygenated hemoglobin (Oxy-Hb), deoxygenated hemoglobin (Deoxy-Hb), total hemoglobin (Total-Hb), and TOI. For analysis of the NIRS data during and after the passive leg movements, the baseline value for each individual was corrected to zero by subtracting the initial resting value. Thus, in our calculation we could eliminate basal muscle metabolism and directly compare metabolic changes during the passive leg movements. Average data of the last minute of exercise and first minute of the recovery stage were considered for the comparison between the paretic and nonparetic sides.

Statistical Analysis

All values are given as mean ± SD. Statistical software SPSS 14.0 was used to perform all related analyses. A paired *t* test was used to determine significant differences between the paretic and nonparetic calf muscle oxygenation and muscular activity, and nonpaired *t* test was used to assess the difference between the clinical subgroups. Significance was accepted at $P < .05$.

RESULTS

All participants tolerated the exercise well and the experiment was performed undisrupted throughout the

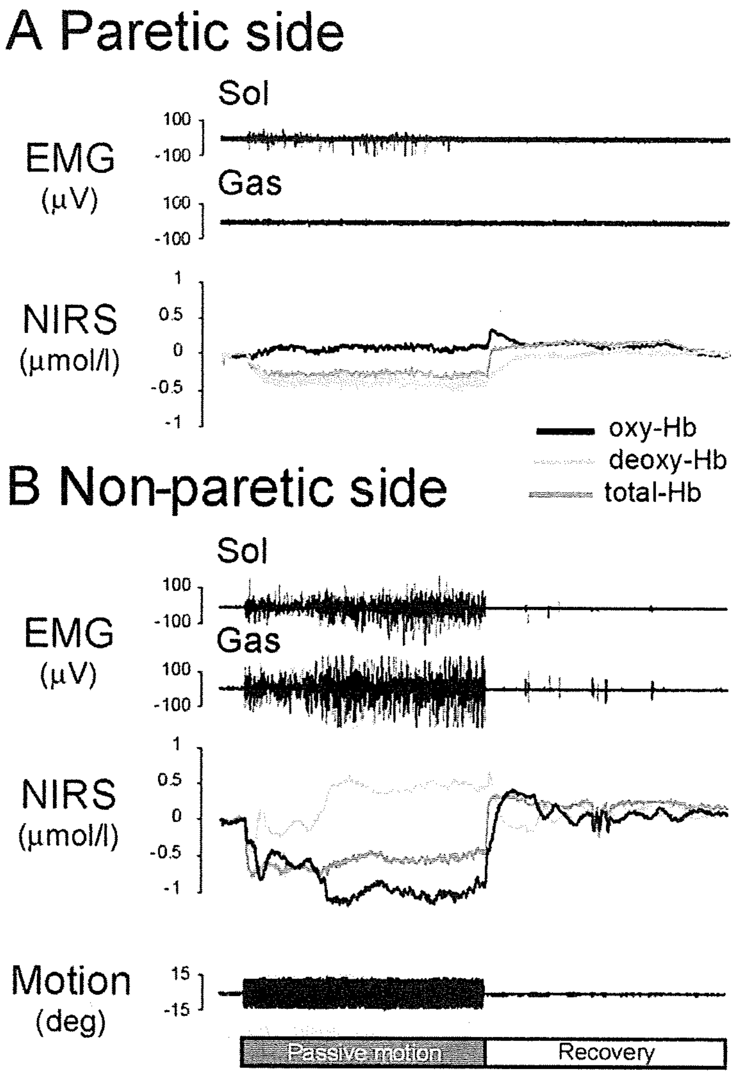
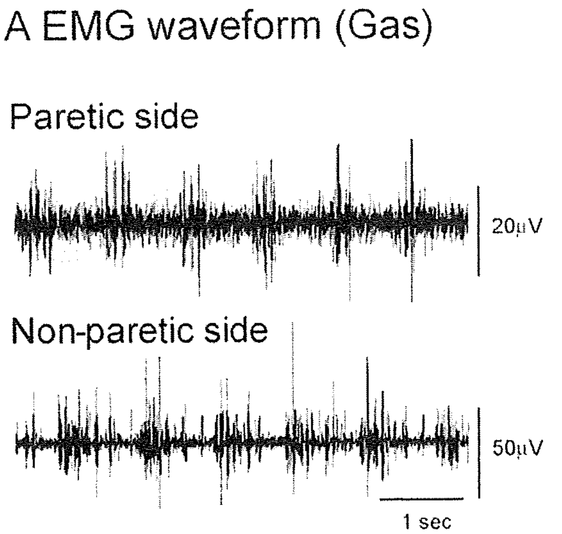


Figure 2. Typical waveform representing oxygenation changes and muscular activity in the muscles of the paretic and nonparetic legs in patients with unilateral paresis. Evident differences in the overall oxygenation of the paretic and nonparetic muscles of the lower limbs of chronic stroke patients.

exercise. During the data analysis we found that patients with bilateral stroke impairment have oxygenation changes similar on both sides, whereas hemiparetic patients show obvious differences between the paretic and nonparetic legs. For further analysis and comparison between the paretic and the nonparetic muscle changes during the passive movement, we compared the data obtained from 12 subjects with those of the hemiparetic patients.

Figure 2 shows a typical example of the NIRS and EMG data of the stroke patients. As is clearly shown, oxygenation changes during passive leg movements were quite different between the paretic and nonparetic sides. Concentration of Oxy-Hb in the paretic muscles started to increase just after the onset of the passive leg movements, but it rapidly decreased in the nonparetic muscle. In both leg muscles, the TOI increased during the passive leg movements and gradually decreased after the cessation of the exercise. In the present study, heart rate data were measured during the experiment to know the change of central circulation in response to the passive leg motions. The results showed no remarkable change of the heart rate during exercise and recovery period.



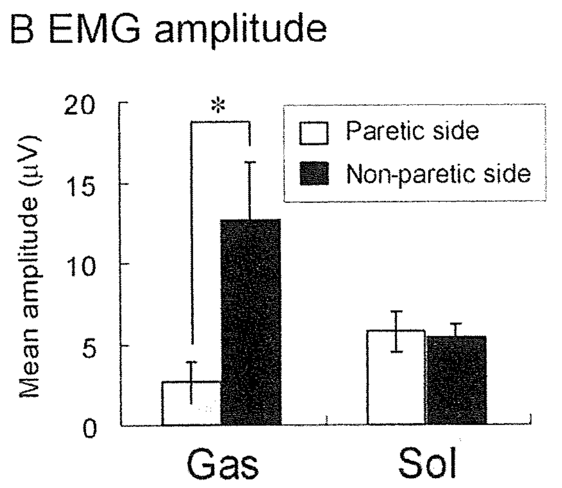


Figure 3. EMG activity of the gastrocnemius muscle in the paretic and nonparetic lower limbs during the passive leg movement. The average amplitude of the nonparetic muscle was significantly larger in comparison with the paretic (*$P < .05$).

EMG Activity

During the passive leg movements, almost all patients showed EMG activity in both the paretic and nonparetic legs. As shown in Figure 3A, the amplitude of the EMG activity was variable during passive leg motion. Although the amplitude of the EMG activity in the soleus muscle was similar (paretic vs nonparetic: 5.76 ± 4.3 vs 5.43 ± 4.3 μV; NS), that in the gastrocnemius muscle was remarkably different (Figure 3B) between paretic and nonparetic legs (2.66 ± 2.7 vs 12.66 ± 12.3 μV; $P < .05$).

We compared average amplitude of EMG activity between subgroups with good and poor motor recovery of lower limbs. There was no significant difference in the amplitude of soleus muscle (7.61 ± 5.27 μV in the poor recovery subgroup vs 5.81 ± 6.23 μV in the good recovery subgroup; NS) or in the amplitude of gastrocnemius muscle (3.89 ± 4.27 μV in the poor recovery subgroup vs 5.73 ± 4.83 μV in the good recovery subgroup; NS).

Maximal voluntary contraction (MVC) level was significantly larger in the gastrocnemius (132 ± 122.2 μV vs 42.6 ± 75.4 μV; $P < .01$) and soleus (155.4 ± 390 μV vs 52.3 ± 28.9 μV; $P < .01$) muscles of the nonparetic leg in comparison with the paretic. In the good recovery subgroup the MVC level in the paretic soleus muscle was larger in comparison with the poor recovery subgroup (73.3 ± 25.7 μV vs 31.4 ± 10.8 μV; $P < .01$). MVC level of gastrocnemius muscle was also higher in the good recovery subgroup, but there was no statistical significance.

NIRS Parameters

As indicated in Figure 4, a clear difference between the level of Oxy-Hb in the paretic and nonparetic legs was observed. The concentration level of Oxy-Hb in the paretic leg was higher than that of the nonparetic leg (0.13 ± 0.09 μM/L vs –0.18 ± 0.10 μM/L; $P < .05$). The increase of Oxy-Hb was not followed by increments of Deoxy-Hb. The concentration of Deoxy-Hb in both legs was almost equal (–0.89 ± 0.12 μM/L in the paretic and –0.79 ± 0.20 μM/L in the nonparetic; NS). During the recovery stage, the concentration of Oxy-Hb remained higher in the paretic leg (0.26 + 0.09 μM/L vs 0.12 + 0.05 μM/L; $P < .05$). There was no difference between the concentration levels of Tot-Hb (–0.75 ± 0.09 μM/L vs –0.96 ± 0.14 μM/L; NS) and TOI (0.11 ± 0.02 vs 0.08 ± 0.03; NS) in the paretic and nonparetic muscles.

To investigate effects of passive leg movement with regard to the functional level of stroke patients, we divided the hemiparetic group into 2 subgroups in accordance with patients' motor recovery level. Patients with Brunnstrom scores of 5 and 6 for the lower limb were classified as the subgroup with good motor recovery ($n = 6$), and patients with scores of 4 and below were classified as the subgroup with poor motor recovery ($n = 6$). In comparison with the poor recovery subgroup, the good recovery subgroup had higher Motricity Index (72.1 ± 2.8 vs 46.8 ± 4.4; $P < .01$) and muscle strength scores (4–5 vs 1–4). None of the good recovery subgroup had spasticity in the lower limbs. Average cadence at the individually chosen comfortable speed was not significantly different between these 2 subgroups (87.8 ± 4.7 in the good recovery subgroup vs 61.5 ± 13.2 in the poor recovery subgroup; NS).

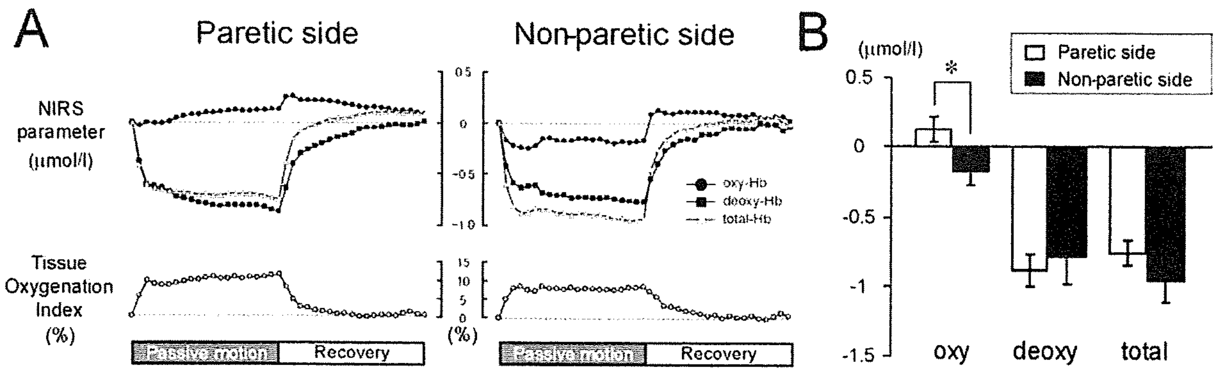


Figure 4. A, Oxygenation changes during the passive leg movements in the paretic lower limb muscle of stroke patients. Concentration level of Oxy-Hb during the exercise in the paretic leg was higher in comparison with that of the nonparetic leg. The concentration remained higher after the cessation of exercise (*$P < .05$). B, Bar graph overview of the oxygenation changes during the passive leg movement.

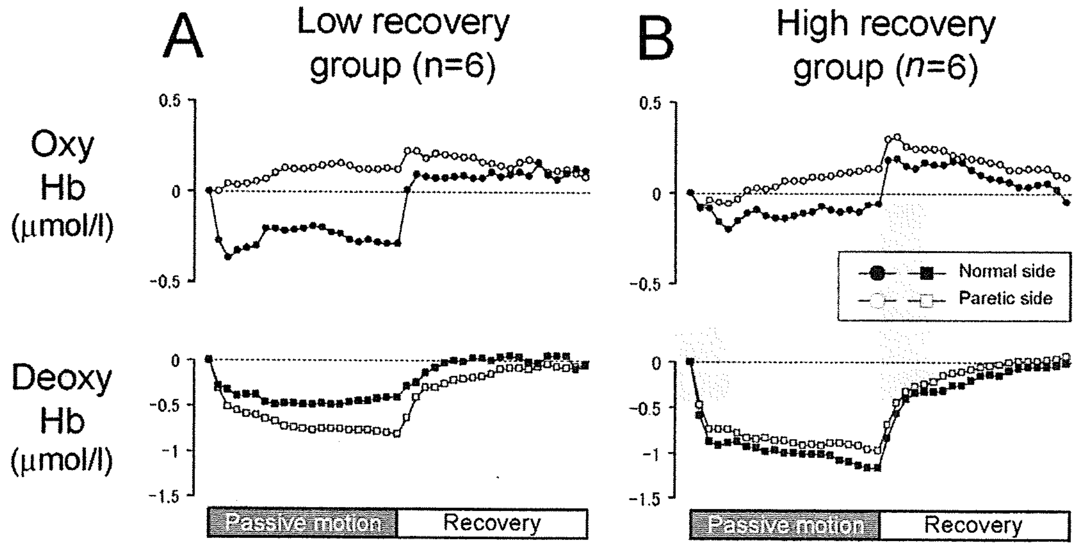


Figure 5. Muscle oxygenation changes in the lower limbs of the patients with good and poor motor recovery after stroke. Strong discrepancy in the level of Oxy-Hb between paretic and nonparetic legs was observed in the group with the low level of motor function.

Figure 5 shows the averaged data of the NIRS for the 2 subgroups. Although the difference of oxygenation changes between these two subgroups was not significant, there was a difference between the paretic and nonparetic legs within the clinically differentiated subgroups. For instance, in the poor recovery group, the concentration level of Oxy-Hb in the paretic leg was significantly higher during the passive leg movements (0.12 ± 0.10 μM/L vs –0.29 ± 0.18 μM/L; $P < .05$) and recovery stage (0.22 ± 0.25 μM/L vs 0.05 ± 0.06 μM/L; $P < .05$). TOI at the end of exercise was also higher in the paretic leg in comparison with the nonparetic leg, but there was no statistical significance (0.11 ± 0.01 vs 0.04 ± 0.04; NS).

In the group with higher motor function, oxygenation changes of both legs were more symmetrical in comparison with those of the poor recovery group.

DISCUSSION

We examined the effects of passive movement on the oxygenation level of the calf muscles in chronic stroke patients. The primary observation in this study was that the muscle oxygenation changes during imposed passive leg movements were significantly different between paretic and nonparetic calf muscles of chronic stroke patients. On the other hand, although the degree of change of muscle oxygenation was significantly different, the muscular activation pattern and the enhancement of the muscle oxygenation were similar in both paretic and nonparetic legs during passive leg movements. In the following section, possible mechanisms for the differences between paretic and nonparetic legs are discussed.

Muscular Activity During Passive Leg Movement

In this experiment, all patients showed a significant level of muscular activity as evidenced by EMG in both paretic and nonparetic muscles. During passive leg movements, afferent inputs from proprioceptors and load receptors were periodically provided. The rhythmic EMG activity might be elicited in response to these neural inputs. This finding is in accord with our previous research, which studied appearance of the EMG activity in the paretic muscles of persons with spinal cord injury.[13,14]

We obtained the MVC level in both soleus and gastrocnemius muscles and observed that it was more than 3 times smaller in the paretic side. On the other hand, both muscles demonstrated a different tendency of the EMG changes between paretic and nonparetic legs. Namely, although the muscle activation level in the soleus muscle was similar, that of the gastrocnemius muscle was significantly larger in the nonparetic leg. Although these two muscles contribute equally to plantar flexion motion, previous studies suggested that each muscle functions differently during walking or voluntary contraction, which presumably depends on different neural and mechanical properties.[24,25] These points can serve as an explanation for the present result.

Interestingly, although the patients had a remarkable deficit of the voluntary activation of the paretic muscles, passive movement–induced EMG activity was observed in both paretic and nonparetic muscles. Also, in the patients with good motor recovery, the MVC level of the paretic calf muscle was higher than in the poor recovery subgroup, but there was no difference in the passive movement–induced EMG activity. These results therefore suggest that the passive movement could induce the EMG activity in the paretic muscle regardless of its voluntary contraction capacity.

Muscle Oxygenation Changes

We recently examined the effects of passive leg movement on the muscle oxygenation in persons with spinal cord injured persons and confirmed alteration of muscle oxygenation in the paralyzed muscle.[14] Similarly, the present results clearly indicate that the muscle oxygenation level of both paretic and nonparetic calf muscles could be enhanced by imposing passive leg motion. However, the results demonstrated some different behavior in the NIRS data between the paretic and nonparetic side. The level of Oxy-Hb gradually increased during the passive leg movements in the paretic lower limb, whereas it decreased in the nonparetic lower limb. Despite the large difference in the concentration changes of Oxy-Hb, those of Deoxy-Hb were almost equal in both legs. It is well recognized that the concentration changes of Oxy-Hb and Deoxy-Hb are determined by the relative condition of the tissue oxygen demand and supply.[26] Therefore, it might be possible that the larger Oxy-Hb in the paretic leg reflects that the oxygen delivery far exceeded the oxygen extraction compared to the nonparetic leg. Taken together with the present results that the EMG activity during passive leg movement of the gastrocnemius muscle was significantly smaller in the paretic side, it is very likely that the increased Oxy-Hb reflects a smaller extent of oxygen consumption. This explanation might be reasonable because the relative changes between Oxy-Hb and Deoxy-Hb are strongly dependent on the muscle contraction level.[27,28] The presence of muscle atrophy in the paretic muscle of stroke patients could be one possible explanation for this imbalance. Because the lean muscle mass[4] and the number of fibers are reduced[2] in the paretic muscle of stroke patients, it is possible that the oxygen extraction is reduced. However, in addition to the oxygen consumption, other two potential factors, oxygen supply and total blood volume, may influence the present change of Oxy-Hb. The level of Oxy-Hb is influenced directly by oxygen supply, which means blood flow. Therefore the increased level of Oxy-Hb may be a result of the increased blood flow to the paretic muscle.

In the poor recovery subgroup, the Oxy-Hb level also increased significantly during the passive leg movements in the paretic leg and decreased in the nonparetic leg (Figure 4). Although the level of Oxy-Hb increased considerably, we did not observe increments of Deoxy-Hb in this group. In contrast, patients with good motor recovery had a higher concentration of Deoxy-Hb in the paretic muscle. Because the recovery level was higher, an increment of Deoxy-Hb was observed, suggesting that oxygen extraction of the acting muscle improves with the motor recovery. The extent of recovery, of Oxy-Hb just after the passive leg movements was larger in the nonparetic leg. The same condition was observed in the good recovery group. In general, the oxygenation level of the calf muscles during the passive movement tended to be higher in the paretic leg, particularly in those with poor recovery.

Despite the imbalance between the oxygenation parameters in the paretic muscle, TOI increased in both legs during the passive movement. TOI reflects the changes of tissue oxygen saturation relative to the rest. Unlike Oxy-Hb, the TOI is not affected by the total blood volume, but it does depend on the dynamic balance between the oxygen supply and oxygen consumption. As evidenced by EMG, the paretic muscle was contracting at a constant amplitude during the induced leg movement. The increased level of TOI in the presence of muscle contraction at a constant rate would mean that there was constant oxygen consumption during the exercise.

We did not measure the skin temperature before and during the experiment, so it is difficult to ignore the possibility of increased skin temperature during exercise and the effect of that on the NIRS signals. This is one of the limitations of the present study. Some previous studies reported conflicting findings on the influence of skin temperature. Mancini et al[29] showed minimal contribution of skin temperature to NIRS-derived tissue oxygenation, and another study confirmed that even a small increase in skin blood flow may affect NIRS light absorption and potentially confound the interpretation of the underlying muscle oxygenation measurement.[30]

Implications for Rehabilitation

Passive leg movements are broadly used to reduce the risk of developing contractures and pressure sores in chronic neurological conditions. Passive movements proved to be effective in reducing spastic hypertonia in chronic stroke patients[31] and also in eliciting useful activation patterns in the affected hemisphere.[32] Moreover, continuous passive motion has been shown to benefit the healing of soft tissue such as articular cartilage,[33] ligament, and tendon,[34] and may be beneficial in nerve repair and regeneration in animal models.[35] On the other hand, little is known about the metabolic and physiological changes underlying success of the previously mentioned measures. Especially, there is no literature available on the effect of passive leg movements on the paretic muscle metabolism of stroke patients. Evidence of enhanced muscle oxygenation, equality of activation pattern, and amplitude of calf muscle activity in both paretic and nonparetic legs suggest that the passive leg movements we used have significant benefit for enhancing blood oxygenation and longitudinal adaptations of the involved muscles.

A previous study that examined muscle metabolism during a cycling exercise in stroke patients[10] reported that during 2-leg cycling the oxygen uptake for the paretic leg was significantly lower in comparison with the nonparetic leg, and stroke patients were unable to distribute the work equally between the 2 legs. In that type of exercise, the nonparetic leg performed almost two-thirds of the total work. In the case of our experiment, it appears that the Easy Stand Glider allows equal application of exercise load on both paretic and nonparetic legs. It is encouraging that the passive leg movements could alter oxygenation levels in the paretic muscles of chronic stroke patients, who are usually thought to have achieved their maximal functional recovery at 3 months after the stroke onset.[36,37] However, further studies are needed to prove the actual benefit from the passive leg movements and to design a complete training program.

CONCLUSION

Passive leg movements have the capacity to induce muscular activity and enhance the oxygen metabolism in the paretic leg of chronic stroke patients. The effects of passive movements varied in accordance with the motor recovery level. The oxygen extraction was decreased in the paretic muscle and this condition was improved with the motor recovery level. This type of exercise may be a useful and efficient method for prevention of muscle metabolic deterioration in the rehabilitation of stroke patients.

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30 開眼片脚起立時間

北 潔，新村秀幸，糟谷明彦（北整形外科）
浅井 剛（神戸学院大学）

key words 開眼片脚起立時間，one−leg balance，運動器不安定症

はじめに

最近の介護認定の報告をみると軽度要介護者の増加が際立っている．これに伴い転倒予防の概念が普及しバランス機能の評価と運動が広く行われるようになった．本題の開眼片脚起立時間（One−leg balance）は静的バランスの代表的指標である．正常の運動発達で片足立ちができるようになるのは3歳で，処女歩行から遅れること2年である．

1 開眼片脚起立時間

❶ 開発者

片脚起立を臨床的に観察した報告はかなり古くからあり，開発者を特定することは今回できなかった．

❷ 開発時期・初出文献

整形外科の分野で広く知られているのはTrendelenburg徴候（図1）[1]でUber den Gang bei angeborener Huftgelenksluxationと題し*Deutsch Med Wschr* 1：21−24, 1895に発表されている[2]．外科医であるTrendelenburgは先天性股関節脱臼の患者を観察し，従来の説を覆し動揺性歩行の原因は中小殿筋の機能不全とした．一方，神経学の分野では理学所見として30秒できるかが評価されていた[3]．測定方法を図示した記載は3つある（図2～4）[4-6]．

わが国における測定法は図2のごとくであるが，詳細を図5に記す[4,7]．図4は再現性のテストに用いられたものである．

開眼片脚起立時間（One−leg balance）に関して，はじめて地域在住高齢者を対象に大規模調査を行ったものはBohannonらが記した以下のものである（図2）．

▶ Bohannon RW et al：Decrease in timed bal-

図1 Trendelenburg徴候[1]

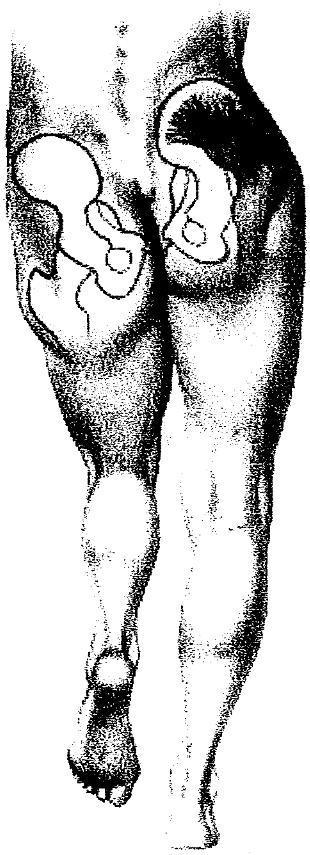

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図2 新体力テスト[4]

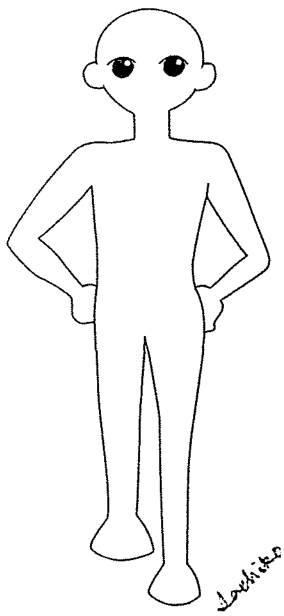


図3 Bohannon ら[5]

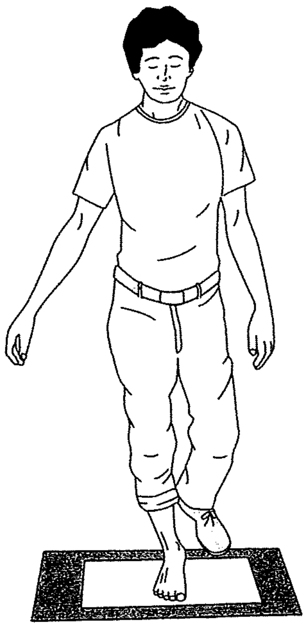


図4 Giorgettei ら[6]

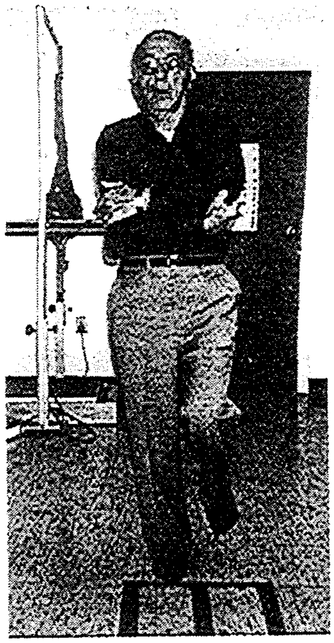
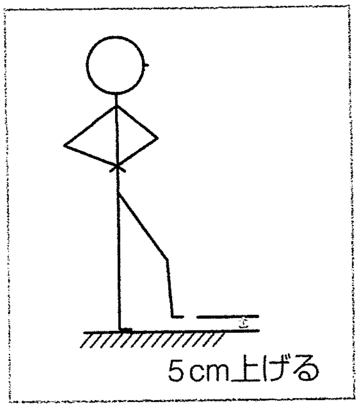


図5 開眼片脚起立時間の測定法[4,7]

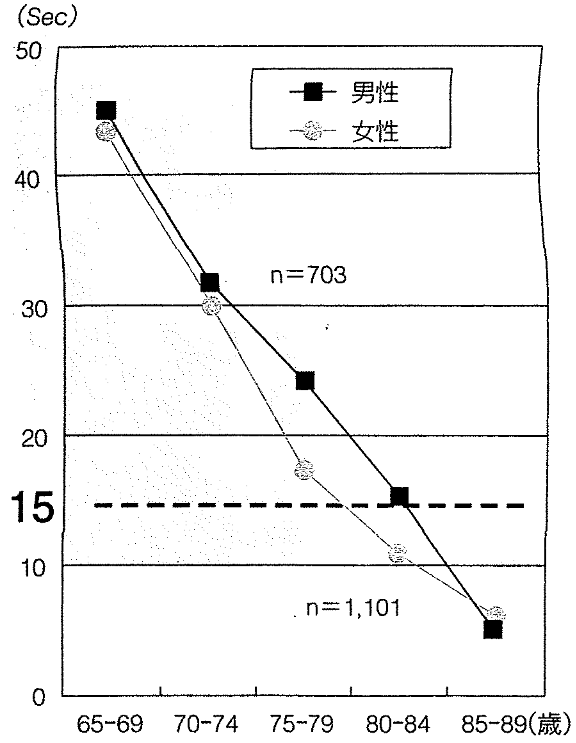
1, 2 度, 手技を実践してもらい, その後, 測定する.



1. 準備　　ストップウォッチ
2. 方法
   1) 可能なら素足, 診療所では安全を考え, 靴を履いても可.
   2) 両手を腰に当てて, 片足立ちの姿勢を確かめる.
3. 記録
   1) 片足立ちの時間を計測する. ただし, 最長は 120 秒とする.
   2) 左右とも 2 回実施し, 少数第 1 位まで記録し, それぞれを記録する.
   3) 1 回目の記録が 120 秒を超えた場合, 2 回目は行わない.
4. 実施上の注意
   1) 滑らない床の上で実施する.
   2) 被測定者の周りには物を置かない, 段差や傾斜がある場所も避ける.
   3) 実施前に, 被測定者に以下の事項を伝える.
      ①片足立ちでできるだけ長く立つテストである.
      ②片足立ちの姿勢は, 支持脚を伸ばし, もう一方の足を前方に少し挙げ, 挙げた足は支持脚に触れない姿勢であること.
      ③テスト終了の条件は, a) 挙げた足が支持脚や床に触れた場合, b) 支持脚の位置がずれた場合, c) 腰に当てた両手, もしくは片手が腰から離れた場合であること.
5. 「はじめ」という合図をすると, 合図だけでバランスを崩す人がいるので, 片足を挙げての合図をし, 片足立ちになった時から計測するほうがよい.
6. 終了の条件を徹底しておき, また, 被測定者に練習させておくとよい.

(文部科学省の新体力テストを一部改変)

図6 地域在住高齢者の開眼片脚起立時間の変化(埼玉県)[9)]



ance test scores with aging. *PhysTher* **64**: 1067-1070, 1984[5)].

彼らの報告は開眼片脚起立時間の上限値を60秒としているが，加齢に伴う著明な低下を明らかにしている．

❸ 特徴

開眼片脚起立時間は40歳を過ぎると著明に低下する[8,9)]．図6に埼玉県で行われた地域在住高齢者の開眼片脚起立時間の変化を示す．80歳を過ぎると15秒できる人の割合は50%を切り，生活機能低下がさらに進行し施設利用者レベルとなると3秒未満となってくる．Tinettiは，5秒未満の高齢者をhigh riskとし特別な戦略を立て，転倒予防を行っている[10)]．本検査法の臨床上の最大の特徴は，場所をとらず診察室でも在宅でも簡単に評価できることである．しかし，高齢者が片脚起立をこわがって躊躇したときは，随時測定不能と判断し危険を回避することが重要である．測定中の事故が実際に報告されている．

表5 運動器不安定症の定義[4)]

運動器不安定症の定義
高齢化により，バランス能力および移動歩行能力の低下が生じ，閉じこもり，転倒リスクが高まった状態

診断方法
下記の疾患の既往があるかまたは罹患している者で，日常生活自立度あるいは運動機能が以下に示す評価基準1または2に該当する者

運動機能低下をきたす疾患
- 脊椎圧迫骨折および各種脊柱変形(亀背，高度腰椎後弯・側弯など)
- 下肢骨折(大腿骨頸部骨折など)
- 骨粗鬆症
- 変形性関節症(股関節，膝関節など)
- 腰部脊柱管狭窄症
- 脊髄障害(頸部脊髄症，脊髄損傷など)
- 神経・筋疾患
- 関節リウマチおよび各種関節炎
- 下肢切断
- 長期臥床後の運動器廃用
- 高頻度転倒者

評価基準
1 日常生活自立度判定基準ランクJまたはA(要支援+要介護1，2)
2 運動機能評価1)または2)
　1)バランス能力：開眼片脚起立時間　15秒未満
　2)移動歩行能力：3m Timed up and go test　11秒以上

❹ 信頼性・妥当性

(1) 内部一貫性

今回，内部一貫性に関する文献を見出すことはできなかった．

(2) 検者間信頼性

Giorgettiらは地域社会で生活する非障害者と，地域社会で生活する障害者を対象に，検者間信頼性を調査している．検者間級内相関係数(ICC)は非障害者サンプルでは0.75で，障害者サンプルでは0.85と測定の信頼性が確認された[6)]．

(3) 再テスト信頼性

Linらは地域在住者1,200名に対し，4種類のバランステストの信頼性テストを行った．そのうち60名に検者内信頼性と検者間信頼性を調査し，いずれも級内相関係数(r)[22]は0.93～0.99と極めて高い値を示している[11)]．

(4) 内容妥当性

前述したLinらの調査では，弁別妥当性は高く，高齢者群，転倒歴群，歩行補助具使用群および

ADLの低下が大きい群において，開眼片脚起立時間は短い値を示した．一方，収束妥当性は低く，各検査に対して中程度未満であった[11]．

（5）予測妥当性

同様にLinらの調査では，片脚起立時間が短くなりADLが悪化するとした［オッズ比0.98（95% CI 0.97－0.99）AUC値0.610］．しかし，転倒の発生やADLの改善は予測できなかった[11]．

❺ 普及度

リハビリテーション医学，整形外科学，神経学，老年医学，耳鼻咽喉科学など幅広い分野で用いられており，Vellasらは開眼片脚起立時間が最も臨床的に有用であるとしている[12]．わが国では新体力テストのなかに含まれ，高齢者の体力指標の1つとされ[4]，新しく生まれた疾患概念の運動器不安定症の機能評価基準に用いられている（表）．

❻ その他トピックス

バランスは，静的バランスと動的バランスに分けて評価されるが，開眼片脚起立時間は静的バランスに分類される．評価の再現性，妥当性や予測性では動的バランス（主としてTUG）が優る．立場を変え，姿勢制御観点からみると動的バランスより静的バランスの獲得のほうがより困難とされる[13]．移乗，階段昇降，振り返りや障害物の乗り越えには片脚起立が基軸となるため，加齢変化が顕著である静的バランス障害を，克服することは臨床上大きな課題であるといえる．

近年の報告ではYoudasらは，Trendelenburg徴候が股関節外転筋の機能評価に有効なことを示し，同徴候が現在なお身近な理学所見であることがわかる[14]．Baeznerらは大脳白質の萎縮との関係を調べ，大脳白質の萎縮度の進行とともに，15秒未満に低下した高齢者の比率が増加することを示している．Mildな萎縮では42.3%，中程度では51.8%で，かなりの萎縮では63.6%であった[15]．Estradaらはestrogen 2年間投与例（骨粗鬆症）のDTXの分析を行い，体重当たりの全身の筋量あるいは下肢筋量が，開眼片脚起立時間に相関するとしている[16]．

最後に新しい疾患として認められた，運動器不安定症（表）の筆者らの調査結果について述べる．筆者らは開眼片脚起立時間15秒以下の通院患者を調査対象としたが，対象症例の転倒率（過去1年に転倒を経験した人の割合）は，36%と地域在住高齢者と施設利用者にみられる転倒率の中間の値を示していた．すなわち，通院患者の生活機能低下のレベルは，健常者から要介護者へ移行期にあることがわかる．そして，性差をみると男女比1：3.6で大腿骨近位部骨折の性差とほぼ同等で[17]，Euro QOL値も0.636と低く，運動器不安定症の克服が介護予防の最重要課題であることがわかる．

おわりに

片脚起立は身近な理学所見であるが，片足で立つこと自体が高齢者にはむずかしく，測定にばらつきが生じやすく統計学的分析に向かない．反面，片足立ちで立てるということは優れた姿勢制御が維持されていることを意味し，静的バランスを高める運動指導は高齢者の自立に欠かせない視点である．

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