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Original Article

Biomechanical Analysis of Weight Bearing Force and Muscle Activation Levels in the Lower Extremities during Gait with a Walker

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The biomechanics of using a walker for the partial weight bearing gait and as a method for gradually increasing the muscle activation level were examined with a force plate and surface electromyography. The results showed that the weight bearing force during gait with a walker is determined by the flexion angle of the hip joint. The value remains constant for each stride, indicating that a walker can be used for the partial weight bearing gait. Moreover, the muscle activation levels in the rectus femoris muscle and biceps femoris muscle per unit time during normal gait and gait with a walker with varying hip joint flexion angles were found to be correlated with the weight bearing force and to be constant for each stride. In addition, the muscle activation level was consistent with the level observed during the open kinetic chain resistance exercise with a specific loading level. These findings suggest that normal gait and gait with a walker may be applicable as a method for gradually increasing the muscle activation level.

Key words: gait with a walker, ground reaction force, integrated electromyogram, partial weight bearing gait, muscle activation level

In association with the aging of society and the resultant increase in the elderly population in Japan, the need for the rehabilitation of elderly patients after fractures or surgery is rapidly increasing. In the rehabilitation of elderly patients, there has been a tendency to encourage patients to leave their beds and start walking as early as possible for the prevention of the development of the bed-ridden state, dementia, and disuse syndrome. Under such circumstances, partial weight bearing (PWB) gait is an important approach to treatment.

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In the conventional PWB gait, the therapist makes a patient learn the optimal weight bearing force on the affected extremity by using a scale and then instructs the patient to walk on crutches while maintaining the weight bearing force at the optimal level. However, patients, especially elderly patients, may not be able to learn the optimal weight bearing force by this approach, and the weight bearing force has often been reported to deviate from the optimal level [1-3]. The possible reasons for this include inertia and fatigue [1], insufficient feedback from the weight-bearing site [2], and psychological stress [3]. In addition, variations in the base of support may also be involved. Moreover, this approach does not consider the acceleration resulting from the displacement of the body's center of gravity (COG) during walking, which may be another cause of the weight bearing force to deviate from the optimal. To solve such problems and to stabilize the weight bearing force, the author investigated the usefulness of a gait with a walker (walker gait), which is associated with smaller variations in the base and smaller displacement of the body's COG. In this study, the author investigated the weight bearing force and its variations during the walker gait biomechanically and explored the possibility of applying the walker gait to the PWB gait.

Moreover, as for the disuse syndrome, hypomotility should be avoided, and standing and gait activities should become the center of daily life to help maintain or improve muscle strength [4]. However, during gait, including the walker gait, how many muscles are active is not clear. Consequently, when the gait is included in a physical therapy program, it lacks an objective basis, and it is often entrusted to subjective judgment. Therefore, the author also investigated the relationship between the hip joint flexion angle and muscle activation level during the walker gait and compared the muscle activation level to that observed during an open kinetic chain resistance exercise. The possibility of applying the walker gait as a method to gradually increase the muscle activation level was explored.

Subjects

The same subjects were recruited for both of the following 2 experiments. The subjects consisted of 15 healthy males and 15 healthy females in their 20's, who did not have any past history of disease causing functional disorders of the upper and lower extremities or trunk. They were considered to have a standard body constitution according to the "Tables and Charts for Judging Overweight and Underweight" [5] proposed by the Japanese Ministry of Health and Welfare in 1986.

Their average age, body height, and body weight (mean $\pm$ SD) were 22.3 ± 3.3 years, 164.7 ± 6.9 cm, and 58.3 ± 9.1 kg, respectively. The objectives of the experiments were explained to the subjects, and their informed consents were obtained.

Methods

Experiment 1. In the present experiment, the subjects assumed a standing position with their forearms placed on the armrests of the walker. The lower extremities were placed in a vertical position with the knee joints extended, and the upper extremities were flexed to 90° at the elbow joints. The height of the walker was adjusted for each subject so that the flexion angles of the hip joint could be changed in steps of 10° in the range of 0° to 90°. Walker gait for 10 meters starting from this posture was repeated 10 times; a normal gait starting from the fundamental standing position for 10 meters was also repeated 10 times. Both were done at an arbitrary speed. The gait speed for the middle 5 meters was recorded, and the vertical ground reaction force at the fifth stride of the left leg was determined with a sampling frequency of 100 Hz. The waveforms of the vertical ground reaction force thus obtained were normalized for each subject and each hip joint flexion angle by regarding the body weight and the duration of the standing period as 100. The maximum values were obtained.

The maximum values thus obtained for each subject were examined by the chi-square test for goodness of fit with the level of significance at 95%. This was done in order to investigate whether there were any variations among the values obtained by the 10 determinations in each subject during the walker gait at each hip joint flexion angle. Moreover, the chi-square test for goodness of fit with a level of significance at 95% was conducted to examine the differences between the maximum values of the vertical ground reaction force during the walker gait obtained in the current experiment and the values calculated by Equation ① [6-9] (Fig. 1), as reported previously. This equation indicates the weight bearing force in the standstill state in a walker, which is determined by the hip joint flexion angle.

The waveforms of the vertical ground reaction force for each subject at each hip joint flexion angle were added and averaged to analyze their characteristics. The Delft Motion Analysis System (Primas, Holland) was used for the determination of the vertical ground reaction force (Fig. 2).

Experiment 2. The muscle strength of the left rectus femoris muscle and the left biceps femoris muscles during a maximal voluntary contraction (100% MVC) was measured 10 times (Fig. 3). A surface electromyogram was obtained while the muscle strength was kept at 50% of the mean of the above value (50% MVC) for 10 sec, 10 times, in the same method as that used for the determination of the 100% MVC, at a sampling frequency of 500 Hz. The data for the middle 5 sec were subjected to a high-pass filter of 20 Hz and integrated by

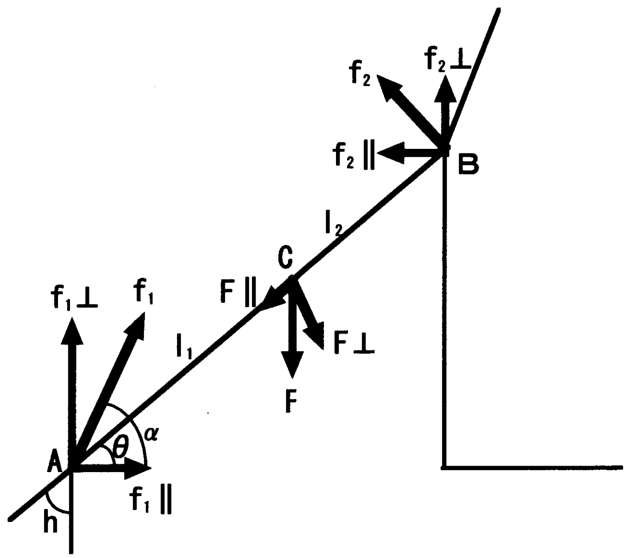

Fig. 1 Equation for calculating the weight bearing force in the standing position in a walker.
Ratio of weight bearing force to body weight; z
Hip joint flexion angle; h
$\theta = 90 - h$
Ratio of the weight of the head and trunk to the body weight; F = 53.1 [6]
Ratio of the weight of both lower extremities to the body weight; 34.4 [6]
Center of gravity when the head and trunk are regarded as one segment; C
Length from point C to the hip joint A; $l_1 = 55.3$ [7-9]
Length from point C to the shoulder joint B; $l_2 = 44.7$ [7-9]
Moment at the hip joint A; f_1
Moment at the shoulder joint B; f_2
$z = 53.1 - l_1 \cdot 53.1 \cdot \cos(90 - h) / (l_1 + l_2) \cdot \cos(90 - h) + 34.4$... ① [9]

eliminating movement artifacts. From this integrated electromyogram (iEMG), the average iEMG per second was obtained, and the standard muscle potential was determined for each subject by regarding the average iEMG as 50%.

Subsequently, the subjects were instructed to walk normally and to walk with a walker with the hip joint at various flexion angles, as in Experiment 1. The muscle activities of the left rectus femoris and biceps femoris muscles during the fifth stride were measured 10 times by surface electromyography with a sampling frequency of 500 Hz (Fig. 2). In addition, resistance was applied to the distal end of the lower leg in the same posture as that used for the determination of the 100% MVC, while gradually increasing the load by 0.5 kg from 0 to 5.0 kg.

Fig. 2 Determination of the ground reaction force and muscle activation level during the walker gait. The position of the electrodes for the rectus femoris muscle is the middle point of a line which connects the anterior superior iliac spine with the anterior superior margin of the patella. The position of the electrodes for the biceps femoris muscle is the middle point of a line which connects the ischial tuberosity with the head of fibula.

The electromyogram was obtained for 10 sec, 10 times, for each muscle at each loading weight. The subjects were allowed a good rest between each of the determinations. The average iEMG per second was obtained, and the % average iEMG was calculated on the basis of the standard muscle potential.

The correlation between the % average iEMG for each muscle thus obtained and the weight bearing force during the walker gait obtained in Experiment 1 was determined by using Spearman's correlation coefficient (level of significance at 95%). The variations in the % average iEMG for each muscle during normal gait and walker gait were examined by the chi-square test for goodness of fit (level of significance at 95%).

To identify the correlations, all combinations of the % average iEMG of the rectus femoris and biceps femoris muscles during normal gait and walker gait in each subject and the % average iEMG during resistance exercise were examined by the one-way analysis of variance and Fisher's PLSD method (level of significance at 95%). The combinations with significant differences were regarded as representative of those with the same muscle activation level.

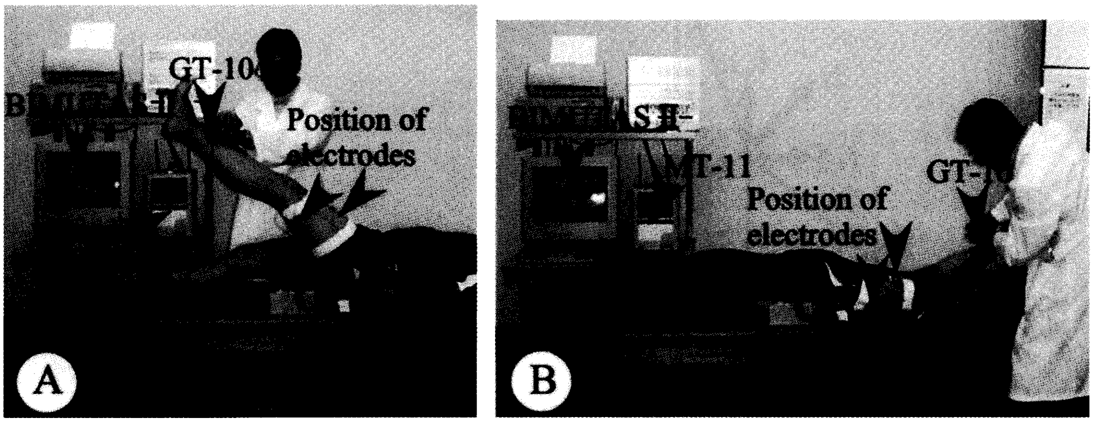

Fig. 3 Determination of the muscle strength of the rectus femoris (A) and the biceps femoris (B) during 100% MVC. The positions of the electrodes are described in Fig. 2. (A) In sitting, trick motions occur easily, so the subjects assumed a supine position. The hip joint was fixed at a flexion angle of 45° by a plaster cast. Then, the subjects were instructed to completely extend the knee joint, and the maximum load under which the subjects could maintain this extended knee position of the knee joint was determined 10 times with the Musculator applied at the distal end of the lower leg. To remove the influence of tight hamstrings, the hip joint was not fixed at 90° of flexion. (B) The 100% MVC of the left biceps femoris muscle was determined in the prone position with the hip joint in neutral and the knee joint at a flexion angle of 45°. Resistance was given with the maximum load under which the knee joint could be maintained at 45° of flexion.

Table 1 Ratio of the ground reaction force in each direction to the total ground reaction forcre (%)

	WG0°[a]	WG10°	WG20°	WG30°	WG40°	WG50°	WG60°	WG70°	WG80°	WG90°
Average of vertical ground reaction force	97.4	97.6	98.4	97.7	98.0	98.3	98.0	98.9	98.1	98.7
Standard deviation	1.5	1.5	0.7	1.8	1.2	1.6	1.7	1.0	1.4	0.9
Average of left-right ground reaction force	1.4	1.1	0.8	1.3	0.9	1.0	1.5	0.7	1.1	0.7
Standard deviation	0.9	0.6	0.4	1.1	0.9	0.9	1.4	0.9	1.1	0.4
Average of anterior-posterior ground reaction force	1.3	1.3	0.8	1.0	1.0	0.7	0.5	0.5	0.8	0.6
Standard deviation	0.8	1.0	0.5	0.8	0.6	0.9	0.9	0.4	0.6	0.6

[a], WG and the angles indicate the walker gait for each hip joint flexion angle.
The number of subjects was 7.

Muscle strength was evaluated with the Musculator GT-10 (OG Giken, Okayama, Japan). The electrodes were placed as suggested by Perotto [10]. The telemeter system MT-11 (NEC Medical Systems, Tokyo, Japan) was used for obtaining the surface electromyograms, and the multipurpose biophysical analysis program, BIMUTAS II (Kissei Comtec, Nagano, Japan) was used for the analysis of the surface electromyograms.

Results

Experiment 1. In Experiment 1, the maximum vertical ground reaction force was defined as the maximum weight bearing force. This definition is based on the results of a pilot study conducted on 7 subjects who satisfied the same conditions as the subjects in this study.

In the pilot study, the ratios of the ground reaction forces in the vertical, anterior-posterior, and left-right directions to the total ground reaction force were determined when the vertical ground reaction force reached the maximum value during the walker gait in the same posture as in Experiment 1. The mean ratio of the ground reaction force in each direction to the total ground reaction force in the 7 subjects was revealed to be as follows: $98.0 \pm 1.3\%$ in the vertical direction, $0.9 \pm 0.7\%$ in the anterior-posterior direction, and $1.1 \pm 0.8\%$ in the left-right direction (Table 1). Moreover, the total ground reaction force was regarded as 100, and the vertical ground reaction force during the walker gait and the standstill state was examined with the chi-square test for

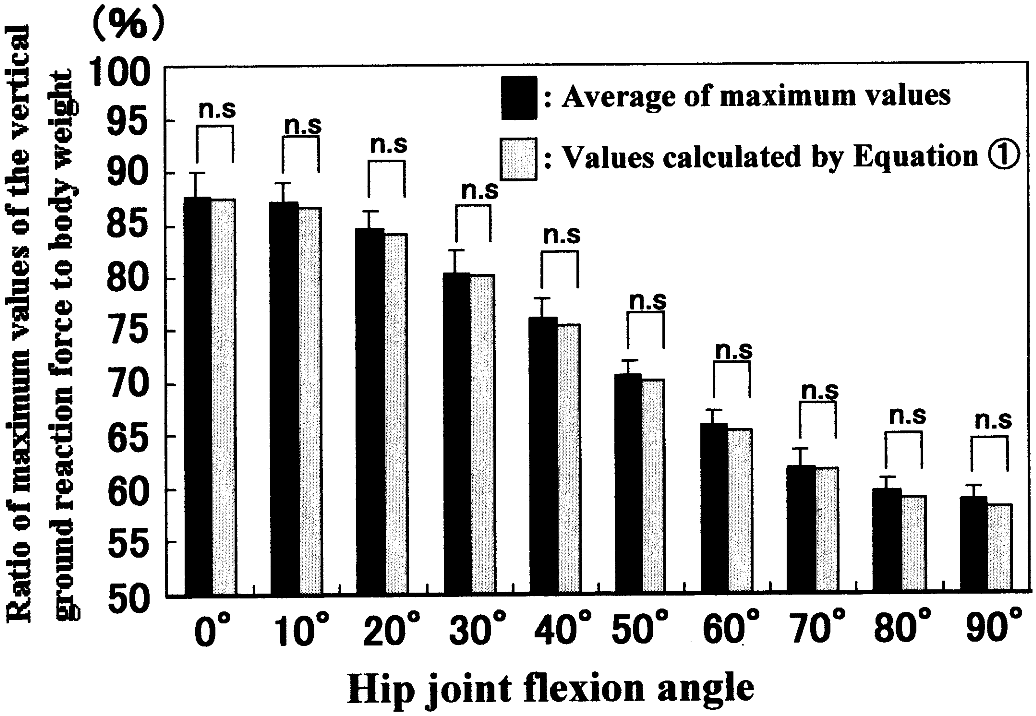

Fig. 4 The relationship between the maximum values of the vertical ground reaction force during a walker gait for each hip joint flexion angle and values calculated by Equation ①.
The calculated values for Equation ① were obtained when $l_1 = 55.3$ and $l_2 = 44.7$. Using these values as the expected frequency and comparing it with the measurement values in the chi-square test for goodness of fit, no significant differences were found (n.s). The average of the maximum values is the average of the 30 subjects, and the error bars represent the standard deviations.

goodness of fit. No significant differences were observed in any of the 7 subjects.

These findings showed that the ground reaction force is focused in the vertical direction during the walker gait. Thus, it is reasonable to ignore the ground reaction forces in the anterior-posterior and left-right directions, and to regard the vertical ground reaction force as the weight bearing force.

Moreover, in comparing the difference between males and females, no significant differences were observed in an unpaired t-test ($P > 0.99$), so all the subjects were included as one group in the statistical analysis.

Results of Experiment 1: the examination for variations in the maximum vertical ground reaction force during the walker gait in each subject yielded highly reproducible results. No significant variations were observed at any of the hip joint flexion angles. Therefore, it became clear that the maximum vertical ground reaction force during the walker gait, in other words, the weight bearing force, was stable. The maximum values thus obtained approximated the calculated values of the weight bearing force in the standstill state in a walker (Equation ① [9], Fig. 1). No significant differences were observed between the maximum values and the calculated values (Fig. 4).

The waveforms of the vertical ground reaction force during the walker gait did not resemble the double peak waves exceeding the body weight, which are usually observed during normal gait. The waveforms during the walker gait tended to remain flat around the maximum level in all subjects at all hip joint flexion angles (Fig. 5). The walker gait speed is, on an average, 69.8% of normal gait (Table 2).

Experiment 2. The Spearman's correlation coefficients obtained between the % average iEMG of each muscle during normal gait and walker gait and the weight bearing force during walker gait determined in Experiment 1 were as follows: 0.864 ($P < 0.001$) for the rectus femoris muscle and 0.827 ($P < 0.001$) for the biceps femoris muscle. Thus, a significant correlation

Table 2 Speed of walker gait (with nomal gait speed considered to be 100)

	WG0°[a]	WG10°	WG20°	WG30°	WG40°	WG50°	WG60°	WG70°	WG80°	WG90°	Mean
Average speed of walker gait[b]	74.2	72.6	70.5	72.1	69.8	69.6	69.1	67.0	68.4	64.9	69.8
Standard deviation	5.9	5.6	5.0	5.6	4.4	5.6	5.1	4.1	5.4	4.7	5.1

[a], WG and the angles are described in Table 1; [b], average speed of walker gait is expressed as an average of all the determinations of the 30 subjects.

Table 3 Matched muscle activation levels per unit time for gait and resistance exercise

Rectus femoris			Biceps femoris		
Gait	Load for resistance exercise	Number of subjects	Gait	Load for resistance exercise	Number of subjects
NG[a]	1.5 kg	28/30	NG	2.0 kg	28/30
WG0°[b]	1.0 kg	30/30	WG10°	1.0 kg	29/30
WG20°	0.5 kg	30/30	WG30°	0.5 kg	27/30
WG40°	0.0 kg	28/30	WG50°	0.0 kg	29/30

[a], NG indicates normal gait; [b], WG and the angles are described in Table 1.
This table shows the relationship between gait and resistance exercises; for example, the muscle activation level per unit time of the rectus femoris muscle during NG corresponded to that during the resistance exercise with a load of 1.5 kg. The number of subjects was 28 out of 30 subjects.

was observed. When the weight bearing force increases, the muscle activity accordingly increases as well. Therefore, as the hip flexion angle decreases in the walker gait, the level of muscle activity was found to increase.

No significant differences were observed among the % average iEMG values determined for each muscle during normal gait and walker gait at varying hip joint flexion angles in any of the subjects. Thus, it was found that if the hip flexion angle in the walker gait is the same, the amount of muscle activity per second is the same for each stride.

The % average iEMG of the rectus femoris muscle and biceps femoris muscle during normal gait and walker gait in each subject and those during resistance exercise showed no significant differences. In other words, the muscle activation level per unit time is the same during gait and resistance exercise (Table 3).

Discussion

Factors explaining the stable maximum vertical ground reaction force. In this study, the maximum values of the vertical ground reaction force during the walker gait were confirmed to be constant for each stride. This may be attributable to the structure of the walker.

During a PWB gait with crutches, the crutches are placed in different positions with each stride, and the base produced by both crutches and the soles changes. In contrast, in the walker gait, the 4, or possibly 3, supporting points remain constant, resulting in smaller changes in the base. This may be the reason why the weight bearing force remains stable.

Effects of acceleration during walker gait. The maximum vertical ground reaction force during the walker gait corresponded to the calculated weight bearing force in the standstill state (Equation ①). Equation ① was formulated on the assumption that the displacement of the body's COG and, therefore, the influences of acceleration, are absent. Its validity has already been demonstrated in the standstill state in a walker [9]. In other words, the maximum vertical ground reaction force during the walker gait seems unlikely to be influenced by the acceleration resulting from the displacement of the body's COG, in spite of the fact that gait is a dynamic activity. This is also evidenced by the absence of the double peak waves exceeding the body weight, which are usually observed in the waveforms of the vertical ground reaction force during normal gait. The double peak waves seen in normal gait are a result of the effect of accelera-

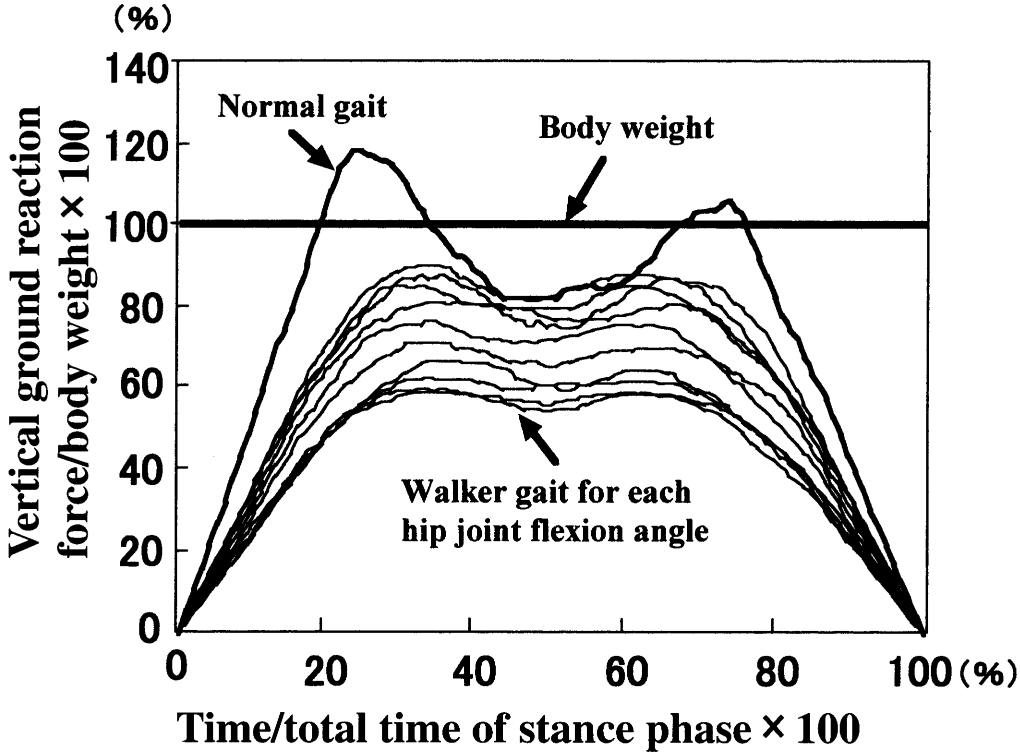

Fig. 5 An example of the waveform of the vertical ground reaction force during walker gait.
The wave forms for the walker gait are for the hip joint flexion angles 0, 10, 20, 30, 40, 50, 60, 70, 80, 90°, respectively, from the top down.

tion, which, as stated above, are not seen in the walker gait (Fig. 5).

One possible reason for the resistance to the influences of acceleration resulting from displacement of the body's COG may be associated with the gait speed and gait posture. With respect to the influences of gait speed on the vertical ground reaction force, Tsuchiya [11] described that when gait speed decreases to about 70% of the normal gait speed, the peak wave exceeding the body weight disappears, and a flat wave around the body weight appears. In our present study, the speed of the walker gait decreased to an average of 69.8% of the normal gait speed (Table 2), which is believed to be responsible for the resistance to the influences of acceleration.

Concerning the posture during the walker gait in this study, the forearms were fixed on the arm rests of the walker, and the elbow joint was maintained at a flexion angle of 90°. Restriction of the movement of the upper extremities appears to have resulted in a constant position of the shoulder joints. Therefore, the vertical movement of the trunk was reduced to only small movements with the shoulder joints as the fulcra, and any displacement of the body's COG was prevented.

The possible relationship between the muscle activation level and muscle strength during the walker gait. The reason that the gradual increase of the muscle activation level during the walker gait may be used as an index of an increase in muscle strengthening is based on a theory provided by Lawrence *et al.* [12]. According to Lawrence *et al.* [12], muscle strength and muscle activation level may not always exhibit a linear relationship, but they are strongly correlated. Moreover, according to Iwatsuki *et al.* [4], muscular atropy can be avoided and muscle strength can be maintained, if a certain amount of gait is continued, even if no special training is done. This is noted in clinical settings often, and, in the author's previous study [13], a similar result was found.

The necessary amount of muscle activation level is most likely the same unit time of the muscle activation level as seen for muscle strength when resistance exer-

Table 4 Examples of maximum and minimum % average iEMG for each muscle

			% average iEMG										
			Normal gait	Walker gait for each hip joint flexion angle									
			–	0°	10°	20°	30°	40°	50°	60°	70°	80°	90°
Rectus femoris	Maximum (subject no. 30)	Average	32.9	27.0	24.0	21.1	18.1	15.1	13.5	11.8	10.5	8.9	7.5
		Standard deviation	0.2	0.3	0.3	0.4	0.2	0.4	0.2	0.3	0.2	0.3	0.3
	Minimum (subject no. 22)	Average	13.2	11.0	10.0	8.6	7.3	6.0	5.4	4.9	4.2	3.6	3.1
		Standard deviation	0.2	0.3	0.3	0.1	0.2	0.1	0.2	0.3	0.2	0.3	0.1
Biceps femoris	Maximum (subject no. 30)	Average	24.0	19.5	17.9	15.5	15.0	13.5	12.0	10.5	9.0	7.5	6.1
		Standard deviation	0.3	0.2	0.3	0.5	0.2	0.2	0.3	0.2	0.3	0.3	0.2
	Minimum (subject no. 22)	Average	8.0	6.5	5.8	5.5	4.9	4.5	3.9	3.5	2.8	2.5	2.2
		Standard deviation	0.2	0.3	0.2	0.3	0.4	0.3	0.2	0.2	0.3	0.2	0.3

			Resistance exercise in a closed kinetic chain at each loading level										
			0.0 kg	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg	3.0 kg	3.5 kg	4.0 kg	4.5 kg	5.0 kg
Rectus femoris	Maximum (subject no. 30)	Average	15.0	21.0	27.0	32.9	40.5	47.8	55.4	67.5	74.9	88.6	94.6
		Standard deviation	0.4	0.3	0.2	0.2	0.3	0.3	0.3	0.3	0.4	1.7	3.7
	Minimum (subject no. 22)	Average	6.0	8.5	10.9	13.2	16.1	19.2	22.1	26.9	29.9	35.9	42.0
		Standard deviation	0.4	0.3	0.3	0.4	0.3	0.3	0.3	0.3	0.3	0.2	0.3
Biceps femoris	Maximum (subject no. 30)	Average	11.9	14.9	18.1	21.0	24.0	27.0	29.9	32.9	37.5	41.7	47.9
		Standard deviation	0.5	0.3	0.4	0.4	0.3	0.2	0.3	0.5	0.3	0.2	0.5
	Minimum (subject no. 22)	Average	3.9	4.9	5.9	7.0	7.9	9.0	9.9	11.0	12.5	13.9	15.9
		Standard deviation	0.2	0.2	0.3	0.3	0.2	0.3	0.5	0.4	0.2	0.5	0.3

Figures in the table indicate the average and standard deviation for 10 determinations of subject number 30, who showed the maximum value, and subject number 22, who showed the minimum values. They represent values obtained when the average iEMG per second during an isometric centraction at 50% MVC was regarded as 50%.

cises are done, as shown in the results of this study. Because these are 2 different activities, muscle strength is not necessarily equal, even if the amount of muscle activation level is equal. The relationship between the muscle activation level and muscle strength increase is entrusted to future research. However, the activity level of patients with disuse syndrome should be increased gradually. To gradual increase the muscle activation level, the walker gait is beneficial.

According to Hettinger *et al.* [14], a load of at least 70% of the maximum muscle strength needs to be applied for the enhancement of muscle strength, and a load of 30 % or less of the maximum muscle strength is not effective for increasing muscle strength. In the results of this study, the load was only 32.9% of the maximum muscle strength, even in the subject who showed the highest muscle activation level (Table 4). This value cannot be directly substituted for the level of muscle strength, but, since the muscle activation level at 50% MVC is normalized as 50%, the muscle activation level of 32.9% is believed to correspond to the average muscle strength of 50% or less. These findings could prompt concern that the loads created by normal gait and walker gait are too small to show a sufficient increase in the muscle activation level to validate the effects on muscle strengthening.

However, according to the results of this study, there were variations in the % average iEMG among the subjects, but those who showed a high % average iEMG in the resistance exercise also showed a high % average iEMG during the walker gait (Table 4). In other words, those with weaker muscle strength need greater muscle activation for movements, and the % average iEMG during resistance exercise becomes relatively high. This is also true of the walker gait. For a patient with a decrease in muscle strength, a high muscle activation level can be obtained, which may result in a maintained or an increase in muscle strength.

The relationship between muscle activity with a walker gait and the closed kinetic chain. Since the walker gait is a closed kinetic chain activity, muscle activity with a walker gait takes advantage of the merits of the closed kinetic chain.

Unlike the open kinetic chain, which is based on a concentric contraction, the closed kinetic chain consists of combinations of co-contraction, and eccentric and concentric contractions of antagonists and prime movers. Therefore, enhancement of muscle activity by a contraction process similar to muscle contractions utilized in daily activities is possible. With respect to the eccentric contraction generated by the closed kinetic chain, Seliger *et al.* [15] examined the muscle activation level and energy consumption of the quadriceps femoris muscle during various forms of contraction in squatting. They concluded that eccentric contraction was associated with the highest muscle activation level, in spite of the lowest energy consumption, and is suited for enhancement of muscle strength. In support of this, Komi *et al.* [16] studied eccentric and concentric exercises for 7 weeks and also reported that eccentric exercise is more effective.

Muscle activation is more likely to be yielded by a closed kinetic chain. The author's previous study [13] also revealed that activities of the closed kinetic chain resulted in a more marked enhancement of muscle strength within a shorter period than with resistance exercises in an open kinetic chain.

Clinical application of the walker gait to a partial weight bearing gait. The results obtained in this study show that, according to calculations, PWB with at least about 50% of the body weight can yield a stable weight bearing force. Moreover, no significant differences were observed in the values calculated by Equation ① (in which the standard position of the COG was used) and the measurement results obtained in this study. Therefore, the values calculated by Equation ① may be applied as an index for the use of the walker gait in PWB (Fig. 4).

In the critical pathway after a total knee arthroplasty and a total hip arthroplasty (Department of Orthopaedic Surgery, Okayama University Medical School), 50% PWB gait exercise is prescribed about 1 week postoperatively. However, the weight bearing force employed may often deviate from the optimal level in PWB gait with crutches. Learning of the optimal weight bearing force takes time, which may result in a delay in the progress of the program. Moreover, monitoring by the rehabilitation therapist or nursing staff is necessary to prevent falling. This tends to impose restrictions on the time and place for the practice of the PWB gait exercise. Thus, poor efficiency of this treatment approach has been noted.

In contrast, a PWB gait in a walker does not necessitate any learning or feedback, and a stable PWB gait is possible by solely making the patient assume the posture employed in Experiment 1. This may lead to a shortening of the critical pathway. Moreover, since the walker gait is more stable than a crutch gait, the risk of falling is reduced, and PWB gait exercise can be performed freely on a daily basis. Thus, the efficiency of this treatment approach provides good results.

Caution must be exercised if the elbow joint flexion angle changes during walking, which may cause the hip joint flexion angle or a component of the force of the head weight toward the sole to change. Consequently, a change in the weight bearing force would occur [17, 18]. It is important to maintain the elbow joint flexion angle at 90° to obtain stable PWB during the walker gait in clinical settings. Moreover, as compared to normal gait, the walker gait is done with the hip joint in flexion and trunk in forward flexion, which might cause low back pain. To clarify this point, the lumbar erector spinae muscle activity during the gait was analyzed. It was found that there is less muscle activity than during normal gait, and the chance of fatigue occurring to be minimal [19]. The possibility of low back pain occurring is considered to be scarce.

Clinical application of the walker gait to progressive muscle activation level increase as an exercise. Use of the walker gait as therapeutic exercise to achieve a gradual increase in the muscle activation level per unit time is possible, if it is practiced with the hip joint flexion angle yielding the same muscle activation level as that achieved by carrying out a resistance exercise. Then, when an increase in the muscle activation level is needed, the hip flexion angle should be adjusted to correspond to the muscle activation level reached with stronger resistance exercise (Table 3). At this time, muscle strength during the walker gait may not be the same muscle strength as during the resistance exercise; however, the muscle activation level should be the same. Because the walker gait can produce continued muscle activity for a long time, a muscle strengthening effect can be expected. Using some of the examples provided in this study, a more objective physical therapy program may be developed (Table 3).

In clinical settings, disuse syndrome often develops as a result of age-related or postoperative hypomotility, and muscle strengthening is difficult to achieve [4]. In such patients, the effects of resistance exercise in the training room may be offset by the decreased activity at other times [13]. It is important to overcome hypomotility [20-22].

With the application of the walker gait, as proposed in this study, it is possible to perform progressive muscle activation exercise every day in activities of daily living. The intensity can be gradually increased by adjusting the hip joint flexion angle. This may help to overcome hypomotility, enable efficient muscle activity, and allow early functional recovery and early return to community life.

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