Constraints of Walking Movement in the Use of a Short Leg Brace

YUKARI OHASHI1), KATSUHIKO NARUSHIMA2), HIROYUKI NAKAMURA2), HİSANAGA YAGYU2), TAKESHI MATSUOKA2)

1)Dept. Physical Therapy, Ibaraki Prefectural University of Health Sciences: 4669–2 Ami-Machi, Inashiki-County, Ibaraki 300-0394, Japan. TEL +81 0298-88-4000
2)Tokyo Medical University

Abstract. The purpose of the present study was to determine (1) whether the pattern of gait with a short leg brace definitely deviated from that of normal gait, and (2) whether a hybrid mass-spring pendulum model proposed in a systems approach was applicable to walking with a short leg brace. Thirty healthy young adults participated in this study. Step rate and Physiological Cost Index during walking on a treadmill were measured while the participants were wearing (a) short leg brace(s) on one leg or on both legs. Walking velocities were 33.3–100.0 m/min for males and 16.7–83.3 m/min for females. The optimal energy-saving gait when using a short leg brace exhibited deviation from the normal gait pattern, because step length was smaller and the step rate was higher. When the velocity exceeded 90 m/min, however, step length was longer and step rate was lower in short leg brace gait than in normal gait. We thought some modifications to the hybrid model were necessary to explain our results.

Key words: Constraints of walking movement, Energy cost, Short leg brace

INTRODUCTION

The purpose of this study was to obtain guidelines for gait exercise when a gait-disabled person wears a short leg brace (SLB). To achieve this goal, walking movement associated with the use of an SLB was investigated in terms of gait pattern and energy cost.

In Japan, physical therapy tends to focus on the reacquisition of movement patterns that healthy persons naturally perform—so-called “normal movement patterns”—and frequently targets the “normal gait pattern,” even when disabled people are performing gait exercise using an SLB. However, the use of an SLB constrains walking movement. In terms of energy cost, such a constraint functions favorably in stroke patients, but does not in healthy persons1). We thus speculated that walking movement with an SLB attached is different from a normal gait pattern.

We previously conducted studies to find gait patterns to save energy when an SLB was worn2,3). In those studies, we compared gait of 32 healthy participants under two walking conditions: when conventional shoes were worn, and when SLB was worn on one side. Walking tests were performed at the same walking speed on a treadmill and energy cost and the walking cycle were measured under the two walking conditions. The results indicated that gait with an SLB was associated with a longer swing phase time than conventional gait in 11 out of the 32 participants, and that energy cost was also lower for these 11 participants than for the other participants when SLB was worn. When acceleration through the long axis of the leg was measured during walking, the group in which the
swing phase was shorter exhibited faster acceleration and deceleration during the swing phase. We attributed this phenomenon to the fact that the participants used muscular activity when they swung their leg quickly during the short-term swing phase, resulting in an increased energy cost.

Based on the results of our previous studies, the purpose of the present experiment was to determine (1) whether the pattern of gait with SLB definitely deviated from that of normal gait and (2) whether the hybrid mass-spring pendulum model proposed in a systems approach was applicable to walking with SLB.

To clarify the first issue, we investigated how the walking ratio (obtained by dividing step length by step rate), which is recognized as being an invariant of normal gait, changed in gait using an SLB. In our previous studies, because the walking velocity was the same, changes in the stride time should have directly influenced the walking ratio. However, although the duration of the swing phase definitely varied with use of an SLB, stride time varied in association with changes in swing phase time in one study, but not in another. Such a discrepancy led us to conduct a more detailed analysis of the walking ratio in this study. Sekiya et al. has shown a walking ratio of 0.0064 for males and 0.0063 for females, in healthy Japanese people walking freely. These values are essentially consistent with those reported overseas in the relevant literature. Therefore, in this study, we analyzed the walking ratio using Sekiya’s value as a standard.

Next, we investigated the application of the hybrid mass-spring pendulum model, proposed by Holt et al., to the gait pattern when SLB was worn. In this hybrid model, the leg is characterized as a simple pendulum during walking. In the swing phase, the leg is a pendulum with the hip joint as its axis, and in the stance phase, it becomes an inverted pendulum with the ankle joint as the axis. The hybrid model hypothesizes the time ratio of the swing phase to the stance phase to be 1:1, so the duration of one cycle of the hybrid pendulum corresponds to stride time in walking. In a virtual space with no axial friction or air resistance, the cycle of the pendulum would be determined based on the length of the pendulum alone. However, since conditions where axial friction and air resistance are zero do not exist in the terrestrial environment, small forces need to be added periodically to keep the pendulum swinging consistently with a periodic cycle. The hybrid model was devised to take these factors into account. As shown in Fig. 1, $k$ is a spring constant, adding a periodic force to the pendulum. Tension in the soft tissues stretched by joint movement while walking (mainly tension induced by stretching of the triceps surae of the lower leg) has been shown to act as a spring in vivo.

The resonant cycle of this hybrid model (see the calculation expressions in the section “Methods”) has been previously shown to be capable of predicting stride time associated with natural gait in healthy persons. Energy cost was lowered when walking at a stride time closer to the resonant cycle of the hybrid model. The hybrid model was also shown to be capable of predicting the prolongation of stride time when a weight was attached to each ankle joint. The stiffness of the affected leg in children with spastic hemiplegic cerebral palsy was shown to increase the spring constant in the hybrid model.

Assuming that the spring of the triceps surae accelerates the cycle of the pendulum in accordance with the hybrid model, we hypothesized that if the ankle joint is fixed at a right angle with SLB, as in
our previous studies, the triceps surae would not be stretched and as a result would not function as a muscle spring. Accordingly, the resonant cycle of the hybrid pendulum would appear to be longer than the conventional cycle. A quicker swing of the leg than in the resonant cycle of the hybrid pendulum would require extra muscular activity, resulting in greater energy cost while walking. In fact, Ohashi showed that stride time was longer in participants who achieved gaits with lower energy costs than others when SLB was worn\(^2\). In our previous study, however, the tests were performed under limited walking speeds, and the reproducibility of the results was insufficient. This study was thus designed to investigate whether this hypothesis of the hybrid model can adequately explain the relationship between gait pattern and energy cost in gait with SLB over a wide range of walking speeds.

Spin rate (or its inverse, step rate) is an important variable to determine the walking ratio as well as to evaluate the applicability of the hybrid model. Thus, we used walking velocity, step length, and step rate as variables for gait pattern in this study. The participants were healthy volunteers. The walking tests were performed on a treadmill, with the participants wore (an) SLB(s) on either one or both sides, and the SLBs were fixed at a right angle ankle joint. In Japan, SLB is frequently fixed with a right angle ankle joint in physical therapy for hemiplegic stroke patients. The final aim of our study was to obtain guidelines for gait exercise when SLB is used. Since physical functions are generally limited in stroke patients, the data obtained from such patients are difficult to analyze. We thus considered it necessary to clarify the potential constraints on walking movement when a healthy person wears an SLB, to define a baseline for the use of SLB. When an SLB is used unilaterally, stride time is likely to differ between right and left according to values predicted by the hybrid model. However, if walking tests are performed on a treadmill, stride times on both sides should be the same. This raises a motor control issue concerning as to how bilateral stride times are adjusted. To examine this issue, we compared walking movement in gait with the unilateral versus bilateral use of SLB in this study.

**METHODS**

**Participants**

Fifteen healthy males (average age; 21.1 ± 4.4 years) and 15 healthy females (average age; 21.9 ± 2.5 years) participated in this study. The purpose, method, and how the results of this study would be dealt with were explained, and consent was obtained from each participant prior to participation in the study.

**Measurement items and apparatus**

Step rate and energy cost during walking were measured while the participants were wearing a SLB on the right side and a conventional shoe on the left side, or while wearing SLBs on both sides. A tape switch (18 mm × 45 mm), used as a foot-switch to measure step rate, was attached from the rear outer heel to the center of the heel of the SLB sole on the right side. Signals generated by the foot-switch were fed into a recorder (Graphtec, Thermal Arraycorder WR7700). Energy cost during walking was measured using the Physiological Cost Index (PCI) proposed by MacGregor\(^{10}\). PCI was obtained by subtracting heart rate at rest from that after walking for 3 min at a constant speed, and this difference was then divided by walking velocity (m/min). Heart rate was measured using an electrocardiograph (Fukuda Denshi, Dynascope 3100) and ECG was recorded using the thoracic bipolar lead method.

**Procedures**

After being fitted with a foot-switch and ECG electrodes, the participant was instructed to sit in a chair and rest until the ECG monitor indicated a stable heart rate. The ECG was recorded after the heart rate stabilized. The inverse of the mean R-R interval recorded during 6 sec was regarded as the resting heart rate.

The participants were instructed to walk on a treadmill (1 m [width] × 2.5 m [length]) under the walking conditions described below. The duration of walking in each trial was 3 min and 10 sec. Step rate and heart rate were measured during the 6-sec interval between 2 min 57 sec and 3 min 3 sec after the beginning of walking.

**Walking conditions**

In the first session, each of the participants wore an SLB on the right side and a conventional shoe on
the left. In the second session, they wore SLBs on both sides. They walked four times in each session at velocities of 33.3, 55.0, 76.7, and 100.0 m/min for males and 16.7, 40.0, 60.0, and 83.3 m/min for females.

The participants were instructed to sit at rest between the first and second sessions and between walking trials at the different velocities. No session or trial was performed until the resting heart rate was secured.

Calculation of the resonant cycle of the hybrid model

Holt’s method for calculating the resonant cycle of the hybrid model is shown below. Resonant cycle, \( \tau \), of the hybrid model, as shown in Fig. 1, is expressed by equation (1)

\[
\tau = 2 \pi \times \left( \frac{mL^2}{(mLg + kb^2)} \right)^{1/2}.
\]

where the numerator in the square root is the moment of inertia and the denominator is the sum of the torque of the pendulum due to gravity and the torque of the spring due to stiffness of the leg. When the ratio of \( kb^2 \) to \( mLg \) is replaced by \( u \),

\[
kb^2 = u \times mLg \quad \text{equation (2)}
\]

When equation (2) is substituted into equation (1),

\[
\tau = 2 \pi \times \left( \frac{mL^2}{((u + 1)mLg)} \right)^{1/2}
= 2 \pi \times \left( \frac{L}{((u + 1)g)} \right)^{1/2} \quad \text{equation (3)}
\]

Then, when \( u+1 \) in equation (3) is replaced by \( n \),

\[
\tau = 2 \pi \times \left( \frac{L}{ng} \right)^{1/2} \quad \text{equation (4)}
\]

Equation (4) is a general expression of the resonant cycle in the hybrid model.

In this hybrid model, the ratio of \( kb^2 \) to \( mLg \) in natural walking in healthy persons is known to be 1\(^4\). Thus, when healthy persons walk naturally, \( n=2 \) in equation (4). \( L \) in equation (4) is obtained by dividing the moment of inertia of the leg by the product of the mass of the leg and the radius of gyration. Anthropometric data of the participants are required to calculate \( L \).

Weight, leg length, thigh length, and below-knee (lower leg and foot) length were measured according to Winter’s anthropometry\(^{13} \) to calculate weight, center of mass, and radius of gyration of the thigh and of the below-knee segment. Weight and center of mass of the SLB were measured by a simple scale-and-lever apparatus. The moment of inertia about the center of gravity of the SLB was calculated from the difference between the cycle of a pendulum table by itself and the cycle when the SLB was mounted on the pendulum table\(^{12} \).

RESULTS

Step length, step rate, and walking ratio

Walking variables are known to be generally dependent on leg length. In this study, leg length was 85.3 \( \pm \) 4.0 cm for males and 78.2 \( \pm \) 4.0 cm for females, with significant differences between males and females (\( df=28, t=4.815, p<.0001 \)). Thus it was necessary to adjust the walking variables for males versus females. We adopted a method for adjusting the walking variables by using leg length in the following equations\(^6,13 \):

Adjusted step length:

\[
\text{Adj-SL} = \frac{\text{step length}}{\text{leg length}} \times \frac{\text{average leg length}}{\text{average leg length}}
\]

Adjusted step rate:

\[
\text{Adj-SR} = \frac{\text{step rate}}{\text{leg length}} \times \left( \frac{\text{average leg length}}{\text{leg length}} \right)^{1/2}
\]

Adjusted walking velocity:

\[
\text{Adj-V} = \text{Adj-SL} \times \text{Adj-SR}
\]

Adjusted walking ratio:

\[
\text{Adj-WR} = \frac{\text{Adj-SL}}{\text{Adj-SR}}
\]

Data obtained from this experiment were divided into four groups according to gender and brace condition (unilateral or bilateral use of SLB) to compare the gait patterns associated with walking velocity among the groups. There were significant correlations between Adj-SL and Adj-V in all conditions (all: \( r>.9, p<.0001 \)), as expressed by linear regression equations. Additionally, significant correlations between Adj-SR and Adj-V were found in all the groups, as expressed by quadratic regression equations (all: \( r>.8, p<.0001 \)). Figures 2 and 3 show changes in Adj-SL versus Adj-V and changes in Adj-SR versus Adj-V, respectively.

Regression equations of Adj-SL and of Adj-SR versus Adj-V were significant in all individual participants (\( p<.05 \)). Thus, 2 \( \times \) 2 (gender \( \times \) brace condition) analyses of variance with repeated measures were performed on slope of linear regression for Adj-SL and on the quadratic terms in regression equations for Adj-SR. The results of the quadratic terms in regression equations for Adj-SR versus Adj-V revealed significant main effects of gender (\( df=1/28, F=12.502, p<.01 \)) and brace
condition \((df=1/28, F=5.072, p<.05)\). More specifically, the regression curve had greater curvature in females than in males, and was greater with bilateral use of SLBs than with unilateral use.

Next, we investigated changes in Adj-WR as a function of Adj-V. The mean values and standard deviations of Adj-WR at each Adj-V over the four conditions are given in Table 1. When regression analysis of Adj-WR associated with Adj-V was conducted for individual participants, no significant regression expression was obtained. We thus analyzed Adj-WR by gender in 2 × 4 (brace condition × Adj-V) analyses of variance with repeated measures. Fisher’s PLSD \((p<.05)\) was used for post hoc testing. Significant main effects of Adj-V were found in both males and females \((df=3/84, F=30.547, p<.0001 \text{ for males}; df=3/84, F=17.167, p<.0001 \text{ for females})\). Post hoc testing revealed significant differences in Adj-WR among all velocity conditions in male participants. In female participants, there were significant differences in Adj-WR between the 85.2 m/min velocity condition and the other three velocity conditions, and between the 17.1 m/min velocity condition and the 61.4 m/min velocity condition. There was neither significant main effect in the brace condition nor interaction in either males or females.

Value for \(n\) in the hybrid model

In the hybrid model, the \(n\) value used in the calculation of pendulum cycle time is an important variable influencing step rate. In walking at a step rate that healthy individuals freely choose, usually defined as the optimal step rate to save energy cost, \(n\) takes on a value of 2. When walking at \(n\)=2, the step rate is lower than in optimal gait; conversely, in walking at \(n>2\), it exceeds the optimal step rate.
Thus, in this study, we have calculated $n$ under various walking conditions and investigated the gait pattern with SLB(s) by using these values as variables. In calculating the $n$ value in the hybrid model, actual gait values are used without adjustment. To avoid confusion between actual values and adjusted values for walking variables, pre-adjustment actual values are indicated by the following abbreviations: walking speed, Act-V; step length, Act-SL; and step rate, Act-SR.

The $L$ value was calculated when the hybrid model was applied to each participant in this study, and substituted into equation (4). Furthermore, stride time was calculated from Act-V and Act-SR in each walking trial, and substituted for $\tau$ in equation (4) to calculate $n$ in each walking trial for individual participants. Change in $n$ over Act-V was expressed in the form of a quadratic regression equation. $R^2$ for the regression equation showed slight scattering in the male bilateral SLB condition ($R^2=.6578$). In the other three groups, $R^2$ exceeded .84. Figure 4 shows the relationship between Act-V and $n$ in each group. The quadratic regression equations were significant for changes in $n$ over Act-V in all individual participants ($p<.05$). Thus a 2 × 2 (gender × brace condition) analysis of variance with repeated measures was conducted on the quadratic terms in the regression equations. Although the main effect of gender was not significant, brace condition was found to be significant ($df=1/28, F=13.052, p<.01$); that is, the

<table>
<thead>
<tr>
<th>Table 1.</th>
<th>Adjusted walking rate in each walking condition</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Adjusted Walking Velocity (m/min)</td>
</tr>
<tr>
<td>Condition</td>
<td>32.6</td>
</tr>
<tr>
<td>Male unilateral SLB</td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>0.0046</td>
</tr>
<tr>
<td>SD</td>
<td>0.0008</td>
</tr>
<tr>
<td>Male bilateral SLB</td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>0.0054</td>
</tr>
<tr>
<td>SD</td>
<td>0.0017</td>
</tr>
<tr>
<td>Female unilateral SLB</td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>0.0051</td>
</tr>
<tr>
<td>SD</td>
<td>0.0021</td>
</tr>
<tr>
<td>Female bilateral SLB</td>
<td></td>
</tr>
<tr>
<td>M</td>
<td>0.0045</td>
</tr>
<tr>
<td>SD</td>
<td>0.0013</td>
</tr>
</tbody>
</table>

Fig. 4. Relationships between active walking velocity and $n$ value in males (the upper graph) and in females (the lower graph).
curve in the unilateral SLB condition was smaller than in the bilateral SLB condition.

**Optimal gait pattern to save energy**

When actual walking velocity is defined as an independent variable and PCI as a dependent variable, the relationship between the two variables is a quadratic function. The velocity that produces the lowest PCI has been shown to approximate the velocity of free walking\textsuperscript{10,14,15}. In this analysis, walking velocity is a pre-adjustment actual value. In this study, quadratic regression showed significant correlations between PCI and Act-V for all participants, regardless of unilateral or bilateral use of the SLB ($p<.05$). Quadratic regression analysis additionally revealed significant correlation between Act-SR and Act-V, irrespective of brace condition ($p<.05$). The Act-V producing the lowest PCI value in each participant, i.e., the gait velocity optimal for saving energy, was then obtained from these quadratic functions. Act-SR and Act-SL were calculated from the optimal energy-saving gait velocity and defined as the optimal gait pattern.

Walking variables for optimal gait were adjusted for leg length according to the aforementioned method to compare the optimal gait pattern between males and females and between brace conditions. Adj-V, Adj-SL, Adj-SR, and Adj-WR for optimal gait in each condition are given in Table 2. When $2 \times 2$ (gender $\times$ brace condition) analyses of variance with repeated measures were conducted with respect to Adj-V, Adj-SL, and Adj-SR, there were significant gender differences in Adj-V and Adj-SL ($df=1/28$, $F=13.084$, $p<.01$ for Adj-V; $df=1/28$, $F=11.396$, $p<.01$ for Adj-SL). That is, Adj-V was faster and Adj-SL was larger in males than in females. For Adj-V, a significant main effect of the brace condition was found ($df=1/28$, $F=4.985$, $p<.05$), revealing that Adj-V was slower with bilateral use than with unilateral use of the SLB. Analyses of variance for Adj-V and Adj-SL did not show any significant interaction between gender and brace condition. For Adj-SR, there were neither significant main effects nor interaction.

The Adj-WR value has been shown to be constant over a wide range of walking velocities in healthy persons\textsuperscript{6}. Tests were thus conducted on differences between the Adj-WR values obtained in this study and the hypothetical mean value. Hypothetical mean values have been defined as .0064 for males and .0063 for females according to the data obtained by Sekiya et al.\textsuperscript{6}. Although Adj-WR in the male bilateral SLB condition did not significantly differ from the hypothetical mean value, Adj-WR was significantly smaller than the hypothetical mean value in the other three conditions (all; $p<.0001$).

We then calculated the $n$ value of the hybrid pendulum for the optimal energy-saving Act-V in all participants. In the hybrid model, $n$ is reportedly fixed at 2 in conventional optimal energy-saving gait of healthy people\textsuperscript{7}. We compared $n$ for the optimal Act-V determined in each condition with this theoretical mean value of 2. As indicated in Table 3, $n$ did not differ significantly from the hypothetical mean value for males, when SLB was

<table>
<thead>
<tr>
<th>Condition</th>
<th>Adj-V (m/min)</th>
<th>Adj-SL (m)</th>
<th>Adj-SR (steps/min)</th>
<th>Adj-WR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Male unilateral SLB</td>
<td>M 71.7</td>
<td>0.629</td>
<td>114.0</td>
<td>0.0055</td>
</tr>
<tr>
<td>SD 7.35</td>
<td>0.0477</td>
<td>7.62</td>
<td>0.00056</td>
<td></td>
</tr>
<tr>
<td>Male bilateral SLB</td>
<td>M 67.2</td>
<td>0.628</td>
<td>107.7</td>
<td>0.0060</td>
</tr>
<tr>
<td>SD 7.91</td>
<td>0.0746</td>
<td>13.22</td>
<td>0.00133</td>
<td></td>
</tr>
<tr>
<td>Female unilateral SLB</td>
<td>M 63.9</td>
<td>0.591</td>
<td>107.9</td>
<td>0.0055</td>
</tr>
<tr>
<td>SD 8.64</td>
<td>0.0522</td>
<td>8.14</td>
<td>0.00048</td>
<td></td>
</tr>
<tr>
<td>Female bilateral SLB</td>
<td>M 60.1</td>
<td>0.557</td>
<td>108.2</td>
<td>0.0052</td>
</tr>
<tr>
<td>SD 4.97</td>
<td>0.0452</td>
<td>7.96</td>
<td>0.00068</td>
<td></td>
</tr>
</tbody>
</table>

unilaterally worn or bilaterally worn. In female groups, however, \( n \) was significantly smaller than 2 when an SLB was worn unilaterally (\( p<.05 \)) but not when SLBs were worn bilaterally.

## DISCUSSION

### Walking ratio for gait with SLB(s)

In this study, Adj-SL values for Adj-V values were plotted on essentially the same regression line, regardless of gender of participant and of unilateral or bilateral use of SLBs. Adj-SR values for Adj-V values were likewise plotted on virtually the same quadratic regression curve in three conditions, the exception being when male participants wore SLB on one leg. For the latter condition, the curvilinear term in the quadratic regression equation was small enough to make the regression curve indistinguishable from the straight line of linear regression (\( R^2=.8787 \) in quadratic regression; \( R^2=.866 \) in linear regression). Since the pattern in Adj-SL for Adj-V was generally different from that in Adj-SR for Adj-V in this study, the principle of normal gait, i.e., walking ratio being constant in the free walking of healthy persons, did not fit our results. According to Sekiya et al.\(^6\), the walking ratio of a healthy person with a natural gait is .0064 in males and .0063 in females. The values of walking ratio measured in this experiment were generally smaller than Sekiya’s values. According to Zarrugh et al., when walking tests are performed on a treadmill, the walking ratio ranges from .0069 to .0072 in males and from .0061 to .0068 in females\(^1\(^6\). Sekiya’s values are not larger than Zarrugh’s values, and it seems unlikely that the walking ratio would vary greatly under such specific walking conditions as treadmill walking, so Sekiya’s values were used as hypothetical mean values for comparison with the walking ratios obtained under the four walking conditions in this study. The walking ratio in the male unilateral SLB condition turned out to be similar to that of normal gait only when fast Adj-V variables (75.1 m/min and 98.0 m/min) were adopted and was otherwise smaller than that of normal gait. In females, the walking ratio was smaller under all walking conditions than normal gait, regardless of unilateral or bilateral use of SLBs, except that it was similar to normal gait when Adj-V was the fastest (85.2 m/min).

When step rate is relatively large for a given step length, the walking ratio decreases. Thus, changes in Adj-SL and Adj-SR depending on the Adj-V were examined in participants wearing SLBs to determine how these values would differ from those found in a normal gait. Data obtained in this study are compared with data obtained on normal gait by Sekiya et al.\(^6\) in Fig. 5. The graph of step length shows a quadratic regression curve when plotted against walking velocity in normal gait. As illustrated in Fig. 5, step length is smaller in SLB gait than in normal gait when the walking speed is low, but the curves intersect when the velocity is 90 m/min. Beyond 90 m/min step length is larger in SLB gait than in normal gait. The graph of step rate shows a quadratic regression curve in both SLB gait and normal gait. However, magnitudes of the quadratic terms in the regression equations become smaller in the following order: male unilateral SLB > normal gait > male bilateral SLB, female unilateral SLB, and female bilateral SLB (but because the quadratic terms are negative values, the degree of curvature progressively increases). Step rate is larger in SLB gait than in normal gait when the walking speed is low. The regression curves again intersect at 90 m/min between normal gait and SLB gait for all conditions, except for the condition of males with unilateral SLB. When the velocity exceeds 90 m/min, step rate is smaller in SLB gait than in normal gait. Step rate in the condition of males with unilateral SLB exhibited no intersection with that of normal gait and was always larger than

### Table 3. \( n \) value in each walking condition

<table>
<thead>
<tr>
<th>Gender</th>
<th>Male</th>
<th>Female</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Unilateral SLB</td>
<td>Bilateral SLB</td>
</tr>
<tr>
<td>M</td>
<td>2.067</td>
<td>1.860</td>
</tr>
<tr>
<td>SD</td>
<td>0.2837</td>
<td>0.4377</td>
</tr>
</tbody>
</table>

\( n \) values only in the female unilateral SLB condition was smaller than the theoretical value (\( n=2 \)); * \( p<.05 \).
that of normal gait. It may thus be concluded that a decreased walking ratio in this study was attained by combining a small step length with a large step rate in SLB gait when the velocity was less than 90 m/min.

Walking ratio and $n$ value in optimal energy-saving gait

In this study, optimal energy-saving gait with use of the SLB was achieved at a walking velocity ranging from 60.1 m/min to 71.7 m/min. Step length was smaller and step rate was higher with the SLB gait than in normal gait within the above velocity range. Thus, the walking ratio for the optimal energy-saving gait was smaller in SLB gait than in normal gait. However, the walking ratio obtained in the condition of males with bilateral SLBs did not significantly differ from the hypothetical mean value, because it varied greatly from participant to participant. When the hybrid model was applied to the gait pattern with use of the SLB, on the other hand, the $n$ value for gait in male participants did not significantly differ from the normal value of 2. In female participants, $n$ was smaller than 2. According to the hybrid model, gait exhibiting $2<n$ entails a smaller step rate than the optimal energy-saving gait for healthy persons. However, the results of this study revealed that the actual measured step rate was higher than that of normal gait, which contrasts with the hypothesis of the hybrid model. This issue is discussed in detail below.

Limitations of the hybrid model

The hybrid model is an abstract model of movement of the leg during the swing phase. According to the hybrid model, movement of the leg during the stance phase is explained by the theory of the inverted pendulum. However, there are some salient issues of the model that need to be dealt with, especially as they relate to how the inverted pendulum operates during the stance phase. This can be explained by the hypothesis that the time ratio of swing phase to stance phase is 1:1. If the time ratio of the swing phase to the stance phase is 1:1, the following formula is derived; stride time = swing time in the right leg + swing time in the left leg. The pendulum cycle in the hybrid model corresponds to stride time. Since there is the double support phase in actual walking, it is impossible for the time ratio of swing phase to stance phase to be exactly 1:1. However, setting this issue aside, it has been frequently reported that stride time can be predicted according to this hybrid model\(^4\,^5\,^7\,^9\). According to the theory of the hybrid model, $n$ is automatically adjusted to 2 for optimal energy-saving gait of healthy people. We first hypothesized the following idea by adopting the hybrid model. When the hybrid model is applied to SLB walking, the principle of $n=2$, which is supposed to produce an optimal energy-saving gait, breaks down; instead, the optimal energy-saving gait is realized by a stride time approximating $n=1$, i.e., gait with a smaller step rate. This can be
explained by the fact that the muscle spring of the triceps surae, which is thought to influence the hybrid pendulum in normal gait, does not function, because the ankle joint is fixed by the SLB. In our previous studies\textsuperscript{2,3}, a gait pattern that decreased energy cost was attained among participants who wore SLBs by modifying the gait pattern to prolong swing phase. The hybrid model thus appears to offer a reasonable explanation for movement of the leg during swing phase in SLB gait. However, in this study, when stride time and step rate were investigated, optimal energy-saving gait with an SLB was attained when step rate was higher than that of normal gait. When we investigated constraints to movement of the leg when using SLB, the following problems were raised when the hybrid model was applied to walking with the SLB.

One problem arising in the application of the hybrid model to walking with SLB is that the muscle spring constant $k$ is decreased. Another problem is that the fixation of the ankle joint with SLB interferes with the swing of the inverted pendulum during stance phase. When an SLB is worn, swing of the inverted pendulum is possible within the phase between heel contact and mid stance because the rear foot functions as an axis of the pendulum. However, axial rotation of the rear foot stops at mid stance because dorsiflexion of the ankle joint is restricted. The forefoot must subsequently function as an axis of rotation for heel off. In the movement of the leg during stance phase, fixation of the ankle joint thus gives rise to two inverted pendulums, the first in which the rear foot functions as an axis, and the second in which the forefoot functions as an axis. As illustrated in Fig. 6, because the swing of the second inverted pendulum begins at the position where the pendulum resists gravity, it is difficult to begin the swing of the second inverted pendulum unless the walking velocity is adequately fast. Unless the second pendulum is already in motion during gait at a slow velocity, forward movement of the pelvis stops at the point where the first pendulum stops, i.e., where the leg becomes almost vertical, and step length of the contralateral leg is limited as a result. If the walking velocity is fast, on the other hand, it becomes easy to swing the second inverted pendulum resulting in a larger step length. When walking velocity is fast, however, the movement of the leg, acting as a pendulum, must likewise be fast during swing phase. Thus, if the swing of the

![Fig. 6. The schema of two inverted pendulums in SLB gait during stance phase. The first in which the rear foot functions as an axis during the phase between heel contact and mid stance (a), and the second in which the forefoot functions as an axis during the phase between mid stance and toe off.](image-url)
pendulum is insufficient because of an inadequate spring constant $k$, an increase in step rate cannot be achieved. Only when the movement of the inverted pendulum during stance phase with SLB is taken into account in the theory of the hybrid model can the results obtained in the present study be rationally explained. Based on the results of this study, one cannot neglect how wearing SLB affects movement of the leg during stance phase. The original hybrid model by itself, formulated with a normally functioning ankle in mind, is unlikely to lead to a full explanation of changes in gait with SLB, because the influence of the movement of the leg during stance phase is inadequately taken into account.

**CONCLUSION**

The results of this study revealed a certain deviation of the movement pattern in gait with SLB from that of normal gait. More specifically, step length shortened and step rate increased. Such deviation was more marked when walking speed was slower. The principle of a constant walking ratio within a normal gait pattern could not be applied to gait with SLB. When SLB was worn, walking ratio increased with an increase in walking velocity. The walking ratio in gait with SLB reached that of normal gait when walking velocity exceeded 90 m/min. The velocity required for optimal energy cost in gait with SLB was between 60.1 m/min and 71.7 m/min. It thus appears to be fair to say that even the optimal energy-saving gait when using SLB exhibits deviation from the normal gait pattern because step length is smaller and the step rate is higher.

The hybrid mass-spring pendulum model is used to predict the step rate required for optimal energy-saving gait in healthy people. Its applicability, however, was shown in the present study to be inappropriate for the gait pattern when SLB is worn, because the hybrid model is a conceptual model of leg movement during swing phase without taking into account the influence of leg movement during stance phase. In this study, the ankle joint was fixed by the SLB, resulting in a specific constraint to the inverted pendulum movement of the leg during stance phase. This resulting pattern could not be explained by the hybrid model.

Our results have clinical implications regarding gait exercise for patients wearing SLB. In this study, the SLB gait pattern in female participants definitely differed from normal gait, irrespective of unilateral or bilateral use of SLBs, but no differences in gait pattern in females were seen between the unilateral and bilateral SLB groups. This suggests that when females wore SLB on one leg, the other leg did not function normally either. Unilateral use of SLB imposes an appreciable burden on females, regardless of age or the state of health. In male participants, on the other hand, change in the step rate associated with walking velocity notably differed between the unilateral and bilateral SLB conditions. When males wore SLB on one leg, step rate increased in an almost linear fashion, even when walking speed was very high. Thus, even when the walking velocity exceeded 90 m/min, step rate was still higher than in normal gait. On the other hand, gait with unilateral use of SLB for male participants was performed with smaller step length because of the constraint of leg movement during stance phase. So a gait pattern that compensated for slow walking velocity was achieved through increasing step rate, in spite of constraint of leg movement during swing phase by use of the SLB. The effort to produce such a gait pattern did not increase energy cost in young healthy males. However, the acquisition of this gait pattern (i.e., compensating for shorter step length by increasing step rate to maintain walking velocity) was restricted to unilateral use of the SLB. The gait pattern in unilateral use of SLBs was the same for males as for females.

This has important implications for unilateral use of SLB in stroke patients in clinical settings. In view of age and exercise ability in such patients, it seems difficult for them to be subjected to the gait pattern seen in the male participants in this study. In particular, fixation of the ankle joint by the SLB greatly constrains movement of the leg during stance phase. Even if hemiplegic patients were to have the same ability to exercise as young healthy males, step length would have to be smaller than that of healthy persons as long as dorsiflex of the ankle joint were not possible. In this study, the constraints of walking movement in use of the SLB became less prominent when the participants walked somewhat faster. That is, the gait pattern with use of the SLB was naturally closer to the normal gait pattern at higher walking speeds. Thus, instructing patients unable to walk at high speeds to
walk with a normal gait pattern may place a great burden on such patients.

REFERENCES


