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Paper:

Analysis of Trunk Stabilization Effect by Passive Power-Assist Device

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We are developing a passive power assist supporter called *Smart Suit Lite*. Smart Suit Lite, a compact, lightweight device, uses the force of elastic belts both to support muscles and to stabilize the torso in a way similar to the corset. It is reported that the corset stabilizes posture, and decreases intradiscal pressure by applying pressure around the pelvis. The mechanism behind this remains to be fully elucidated, however, and has not been used actively for power assist devices. This study aims to construct a mechanical model of trunk stabilization generated by assistive devices. In this paper, we propose modeling based on basic experiments measuring motions under static conditions.

Keywords: trunk stabilization, power assist, biomechanics

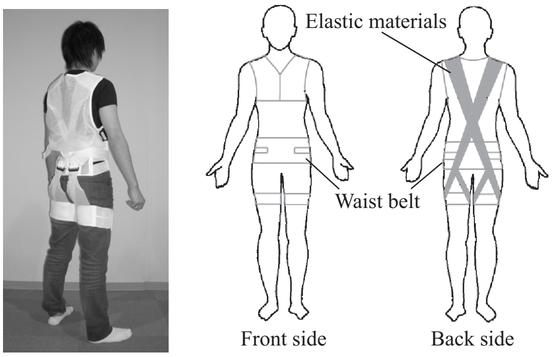
1. Introduction

The physical load caused by the heavy labor such as nursing care has become a major problem requiring that much effort be spent on developing power suits to assist muscles and motion. Sankai et al. developed HAL, an exoskeletal robot suit that uses many sensors and motors [1]. Developed to extend and amplify human physical functions, the effect of HAL in reducing the level of muscle activity has been verified and is currently used in rehabilitation programs. Yamazaki et al. developed a passive suit-type back muscle supporter that uses only the passive force of elastic materials [2]. Developed to reduce the load on muscles, it was designed with magnitude of support force and interference with the movements. Many other power assist suits also have been developed focusing on reducing muscle load [3–5].

We are developing a passive power assist supporter called Smart Suit Lite (SSL) for preventing damage to the back. SSL is a compact, lightweight harness that uses elastic materials and is based on the concept of "KEIROKA," removing the burden on the body – rather than attempting to increase strength. Specifically, it supports the lumbar region by reducing the load on muscles and intervertebral disks. To achieve these effects simultaneously, the assist force is transmitted in part to the torso and lower limbs while remaining assist force is transmitted to the lumbar region.

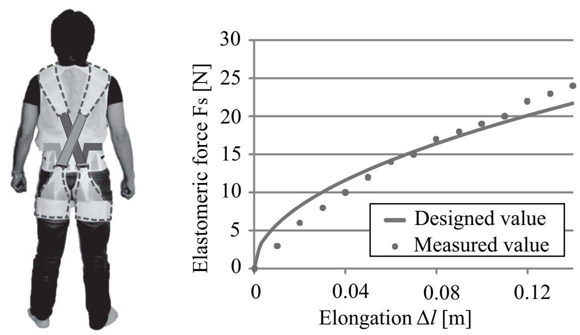
Previously, we have proposed a design method for SSL based on an analysis of the muscle assist effect, and developed an SSL to reduce the muscle load during nursing care [6]. We have confirmed experimentally that the SSL for nursing care is capable of reducing the muscle activity in the lumbar region by an average of 24.4%. In the present study, we employ the same SSL to evaluate and model the trunk stabilization effect, which is the other effect beside reduction of muscle activity.

Among harness devices that possess a trunk stabilization effect, the lumbar corset is widely used to treat or prevent lumbago (lower back pain). It is thought that, by increasing the pressure surrounding the pelvis, the corset increases lumbar support or the intra-abdominal pressure, which reduces the intradiscal pressure and stabilizes posture [7]. Yet, it is difficult to take direct measurements of the intradiscal pressure or joint moments, so its effects have not been adequately evaluated quantitatively. This study proposes a dynamic model of the trunk stabilization effect focusing on the changes in lumbar support, i.e., the motional changes due to assist, aimed at its application to design of assist devices. In this paper, we represent the harness wearer using a link model, estimate the lumbar joint stiffness by motion measurements, and propose a method of constructing a model of the trunk stabilization effect.



(a) Prototype (b) Schematic of elastic belts [6]

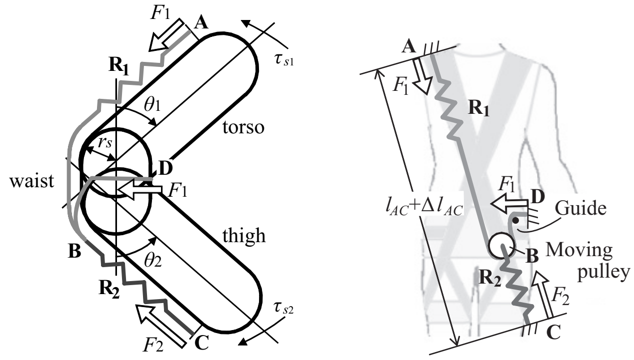
Fig. 1. Smart suit lite (SSL).



(a) Placement of elastic belts

(b) Comparison between designed value of elastic property and measured value [6].

Fig. 3. SSL for caregiving movement.



(a) Lateral view (b) Back view

Fig. 2. SSL assist mechanism [6].

2. Smart Suit Lite

2.1. Assist Mechanism and Design

This section describes the configuration of assist device SSL and its effects. **Fig. 1** shows the SSL prototype and **Fig. 2** diagrams the assist mechanism. Elastic material $\mathbf{R_1}$ on the upper body and $\mathbf{R_2}$ on the lower limbs are connected by moving pulley B. In addition, $\mathbf{R_1}$ is connected to the waist belt at point D after turning up at point B. Assuming that the initial length of the path between A and C to be the natural length of elastic belt, Δl_{AC}, the change in path length when the wearer changes posture, generates elastomeric force F_1 at A and force F_2 at C. These forces generate assistive torques τ_{s1} and τ_{s2}, which respectively work to extend the lumbar joint θ_1 and hip joint θ_2. Assist torque reduces wearer's joint torque and lightens the load on muscles driving the joint. At the same time, elastic force F_1 acts on the waist belt at point D. F_1 tightens the belt, and increases the compression exerted on the lumbar.

Combined torque τ_{s12} and elastomeric force F_1 are given as follows:

$$\begin{aligned}\tau_{s12} &= \tau_{s1} + \tau_{s2} \\ &= \frac{2}{5} r_s k_s \Delta l_{AC} + \frac{4}{5} r_s k_s \Delta l_{AC} = \frac{6}{5} r_s k_s \Delta l_{AC} \quad . \quad (1)\end{aligned}$$

$$F_1 = \frac{2}{5} k_s \Delta l_{AC} \quad \ldots\ldots\ldots\ldots \quad (2)$$

where r_s is the moment arm of elastic material, and k_s the coefficient of elasticity. The above expressions are derived from the force balance between elastic materials with the pulley configuration. Extended length Δl_{AC} of the entire line from shoulder to leg is divided by ratio 1 : 4 between upper and lower elastic material. Details of the assist and design are presented in a previous paper.

The lumbar load is known to increase more as the waist bends [8]. Elongation of elastic material increases the greater the posture change of the wearer. Thus, SSL exerts greater assist in postures that cause greater lumbar load.

We previously designed SSL considering the muscle assist effect. We used the results of muscle activity analysis from using musculoskeletal simulator, SIMM (MusculoGraphics, Software for Interactive Musculoskeletal Modeling), and assist analysis based on a geometrical model. Setting 25% reduction in the activity level of back muscles as the goal, and targeting representative nursing care movement, we designed the elastic material arrangement shown by broken lines in **Fig. 3(a)** and elastic properties expressed as follows:

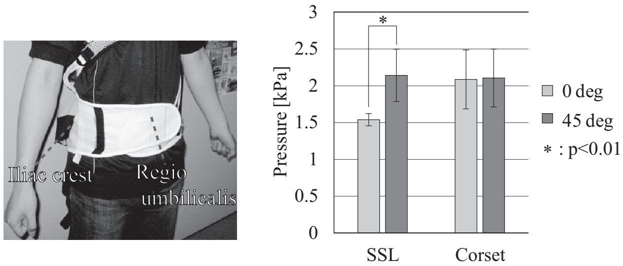
$$\hat{F}_s(\Delta l) = a_1 \Delta l^{a_2+1} \quad \ldots\ldots\ldots\ldots \quad (3)$$

where Δl is elastic belt elongation, $\hat{F}_s$ is target elastic force, and coefficients are $a_1 = 58.1$ and $a_2 = -0.5$. Design values and measurements of elastic properties are shown in **Fig. 3(b)**. We evaluate and model trunk stabilization effects using the same SSL.

2.2. Tightening Force Generated by Posture Change

In this section, we describe tightening force exerted by SSL on the wearer. Past studies have shown experimentally that lumbar corsets reduce the load during physical labor. Some studies have also pointed out negative aspects, e.g., that long-term use may cause trunk muscles to atrophy due to their assist effect [9]. Meanwhile, SSL does not generate tightening when the wearer is standing upright but provides assist only when the user changes posture.

We conducted an experiment to measure tightening force and verified that tightening of SSL varies with the wearer's posture. **Fig. 4(b)** shows tightening pressure



(a) Sites where body pressure was measured (b) Change in body pressure against posture

Fig. 4. Measured value of trunk tightening pressure.

around the pelvis for four subjects (male adults, height 1730±96 mm, weight 64.5±8.0 kg) when they wore SSL or a commercially available soft corset and bent from the waist at angles of 0° and 45°. Using a pressure measurement device (Pressure scanning aid Cello/CR-270, CAPE), pressure between the harness and body at regio umbilicalis and iliac crest, shown in **Fig. 4(a)**, were measured and average values determined. When the corset was worn, there was very little change in body pressure for postural change. When wearing SSL, body pressure increased an average of 34.6% in the bent posture compared to upright posture, where the difference had a significance level of 1%. This result thus verified experimentally that tightening of SSL changes with posture.

3. Optimal Design of SSL Based on Motion

We propose an optimal design (shown in **Fig. 5**) for SSL using a musculoskeletal dynamics simulator. After selecting the motion targeted for assist, the arrangement and properties of elastic material are designed to obtain a targeted assist effect from motion analysis. This method enabled us to design the optimal SSL for different types of work by measuring target motion. There are already several simulators used to analyze human muscle force from movement, but it also remains necessary to incorporate assist effects in the model to design SSL.

As stated in Section 2.1, we previously proposed and evaluated a model of the muscle force assist effect and a design method based on it. In the next section, we propose a method for constructing a model of trunk stabilization whose effect is expressed in dynamic parameters of a human link model. In Section 5, we verify its validity in motion measurement experiments using multiple subjects.

4. Modeling of Changes in Body Characteristics

4.1. Trunk Stabilization Effect

This section describes a method for representing the trunk stabilization effect based on body feature parameters.

Human beings support loads on the upper body using both the vertebral column and the abdominal cavity. Morris et al. experimentally showed that load on intervertebral disks is reduced when intra-abdominal pressure is raised [10]. Abdominal pressure increases when deep trunk muscles contract simultaneously. Matsumaru et al. proposed a model of reducing the load on the lumbar vertebrae based on estimating abdominal pressure [11]. It was also confirmed that pressure around the trunk exerted by a corset increases abdominal pressure similar to that by the trunk muscles [12]. Cholewicki et al. showed that increasing abdominal pressure by using a waist belt and by muscle activity both increase trunk stiffness, and concluded that stabilization based on a waist belt is a passive mechanism unaccompanied by increased muscle activity [13]. Based on these studies, we postulated that when trunk stiffness is increased by tightening and the posture is stabilized, the lumbar load is reduced.

We use a rigid two-link model to represent the human body, shown in **Fig. 6**, to model trunk stabilization. For simplicity, we assume that the lumbar joint possesses flexion only as its degree of freedom. θ denotes the angle of forward trunk flexion relative to the lower limbs. The mass of the upper body is denoted by m, the distance from the lumbar joint to the center of gravity of the upper body by l_c, and pressure exerted by the assist harness on the lumbar region by P. Natural swaying occurs when a person intends to keep posture, so we assume that this oscillation is expressed by stiffness k of the lumbar joint. Note that joint stiffness k in this case is not a physical property of the joint itself, but is observed apparent stiffness of the system, which includes human muscle activity and effects of abdominal pressure. In the sections that follow, we examine a method for representing trunk stability using joint stiffness.

4.2. Estimation of Lumbar Joint Stiffness

We estimate the lumbar joint stiffness of the human link model described in the previous section from motion measurements. The subject is a healthy male (age 24, height 1.893 m, weight 77.6 kg); holding a 6 kg weight. He maintained his posture for 30 seconds as he bent forward from the waist with the back kept straight. He was instructed to keep his feet close together and to look at a marked point that was set 3 m in front of him. The motion was measured by an optical motion capture system (EvaRT 4.3.57; Motion Analysis Inc.) and six cameras (HAWK-200RT; Motion Analysis Inc.) and recorded at a sampling frequency of 120 Hz.

The same experiment was carried out when the subject wore a soft corset for comparison. Two conditions were used for corset tightening. For the first condition, the corset was put on as usual so that tightening pressure remained the same when the posture was changed. For the second condition, the strength of tightening was adjusted to forward flexion angle so that tightening force would be the same as SSL. Measurement results of tightening are shown in **Fig. 7**. Measurements were taken for

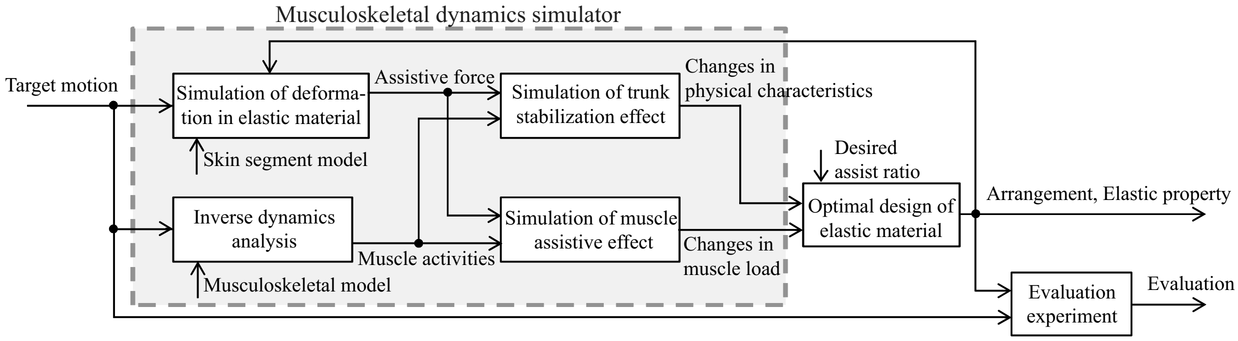


Fig. 5. Design flow of elastic materials for SSL.

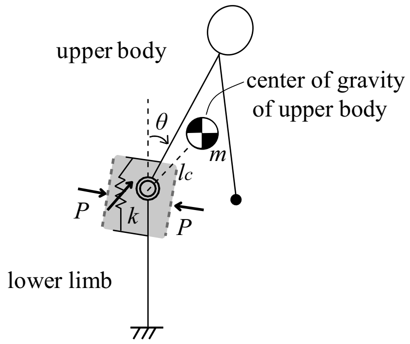


Fig. 6. Dynamic model for analyzing trunk stabilization.

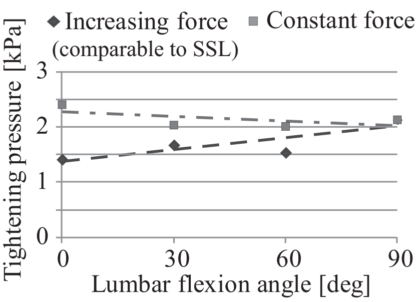


Fig. 7. Applied body pressure.

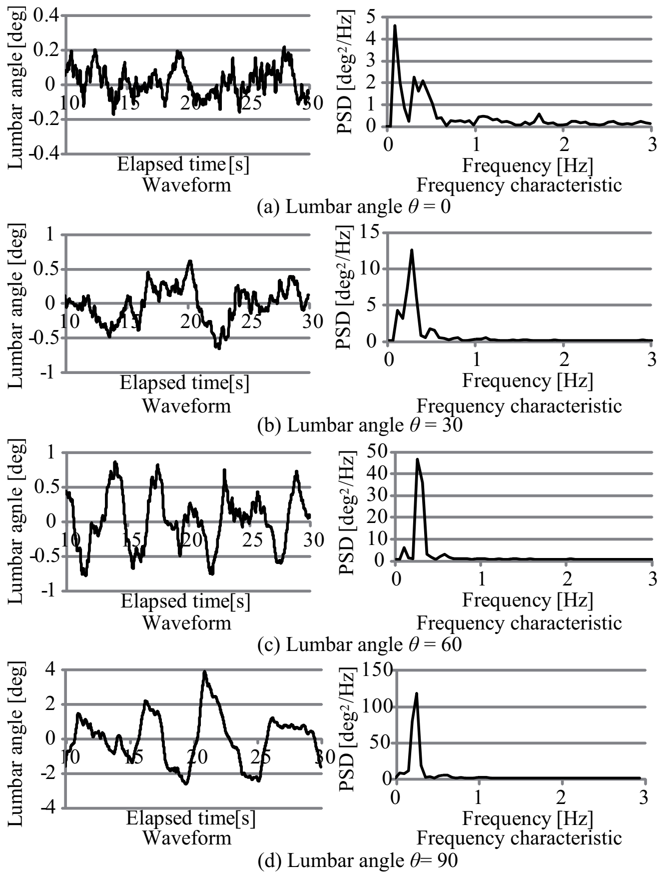


Fig. 8. Example of 0.1–10.0 Hz lumbar angle component and its frequency characteristic.

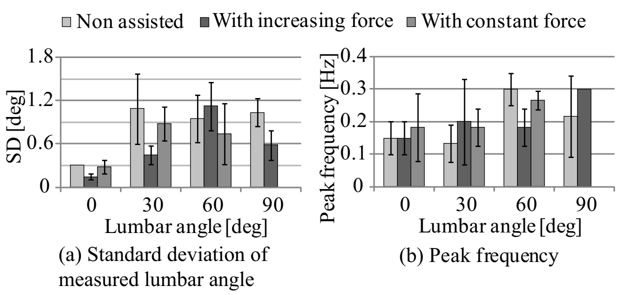


Fig. 9. Standard deviation and peak frequency of lumbar joint sway.

11 conditions by combinations of the assist conditions (*non-assisted*, *with increasing force*, and *with constant force* whose 90° flexion omitted because tightening was the same as when force was increased), and flexion angle (0°, 30°, 60° and 90°), with each condition measured three times in random sequence.

The lumbar flexion angle was processed using a 0.1–10.0 Hz band pass filter. The time-varying flexion angle and frequency components at non-assisted condition are shown in **Fig. 8**. Standard deviation of the forward lumbar flexion angle over 30 seconds is shown in **Fig. 9(a)**. With the exception of 60° flexion, standard deviation is lower when the corset is worn, indicating that the corset serves to control oscillation. In the 60° flexion "with increasing force," tightening pressure did not increase linearly, as shown in **Fig. 7**, and there was no posture stabilization.

Table 1. Moment of inertia.

Lumbar angle [deg]	Moment of inertia [kg·m^2]
0	2.24
30	3.40
60	4.98
90	7.23

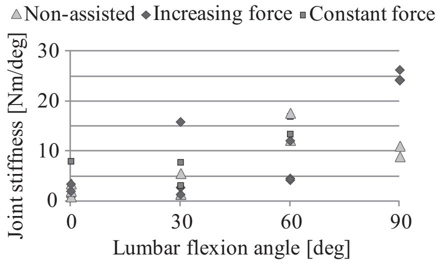


Fig. 10. Relationship between estimated joint stiffness and lumbar angle.

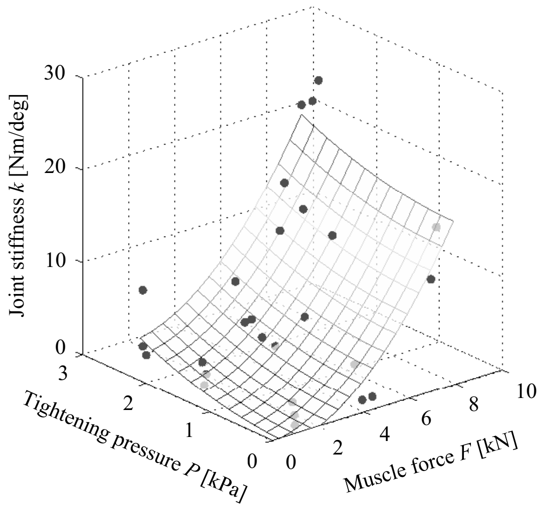


Fig. 11. Relationship between muscle force, tightening pressure, and joint stiffness. The point cloud is values estimated by measurement. The curved surface is the proposed model.

In experiments, tightening force in the direction tangential to the body was increased by adjusting the corset, but the force in the normal direction was affected by factors such as the range of body-corset contact and the hardness of the body surface. In the future, we must develop a pressure distribution model to accurately predict SSL tightening pressure.

We next estimated joint stiffness. Here, we assume that k [N·m/deg] is the lumbar joint stiffness shown in **Fig. 6** against rotation in the flexion direction. We also assume that the upper-body link executes continual free oscillation expressed by $I\ddot{\theta} + k\theta = 0$ relative to the lower-limb link. The relationship between frequency f_p of this oscillation, moment of inertia I of the upper body, and stiffness k is expressed by $2\pi f_p = \sqrt{k/I}$, so k is estimated by the following expression:

$$k = 4\pi^2 f_p^2 I. \quad \quad (4)$$

Moment of inertia I is obtained by $I = ml_c^2$, from upper body mass m and distance l_c between the lumbar joint and the center of mass. m is the sum of 51.5% of the subject's body weight and 6 kg of the weight. l_c is calculated from motion measurement results because it varies with posture. **Table 1** presents the average moment of inertia for each posture. The peak frequency determined from FFT analysis is used as frequency f_p. Average peak frequency is shown in **Fig. 9(b)**. In trials in which posture was stabilized by tightening, peak frequency tends to increase. We confirmed that a negative correlation with a correlation coefficient of -0.5 exists between peak frequency and standard deviation. This suggests that trunk tightening increased lumbar stiffness and that posture was stabilized by finer posture control. Estimated joint stiffness is shown in **Fig. 10**.

4.3. Modeling of Lumbar Joint Stiffness

This section focuses on the construction of a model of joint stiffness. Based on prior studies cited in Section 4.1, we make the following two hypotheses:

1. The increase in trunk stability (stiffness) is achieved by simultaneous contraction of the trunk muscles.
2. Tightening applied externally increases trunk stability independently of muscle activity.

From these hypotheses, we propose the following model equation for joint stiffness,

$$k^* = k_0 + a_F F^\alpha + a_P P^\beta \quad \quad (5)$$

where F is a subject's muscle force and P is tightening pressure around the lumbar region exerted by the harness. It has coefficients a_F and a_P, and exponents α and β, and constant term k_0. These parameters were determined using results described in the previous section. It is difficult to directly measure deep trunk muscles, so we used a musculoskeletal model simulator SIMM. Conducting inverse dynamic analysis using measured motion as input, the sum of estimated muscle exertion of the erector spinae group and abdominal rectus is used as muscle force F [kN]. For tightening pressure P [kPa], measurement shown earlier in **Fig. 7** were used.

Since joint stiffness exhibit quadratically increasing tendencies with tightening pressure and muscle force, we used $\alpha = \beta = 2$ experimentally. Based on multi-regression analysis of Eq. (5), other coefficients were determined to be $k_0 = -1.36$, $a_F = 0.31$ and $a_P = 0.71$. **Fig. 11** shows estimated values of joint stiffness k plotted against muscle force F and tightening pressure P and the curved surface of the model equation. The coefficients of the terms for tightening pressure and the muscle force are both positive, which agrees with our hypotheses

Table 2. Subjects for verification experiments on postural stabilization.

Subject	Age	Height [m]	Weight [kg]
A	24	1.89	77.6
B	42	1.80	73.8
C	25	1.65	56.5
D	27	1.72	64.0
E	26	1.66	60.0

that joint stiffness increases with an increase in tightening pressure and with an increase in the muscle force. The coefficient of determination of the regression equation is 0.73, p-value of the analysis of variance is 7.31×10^{-8}, and the effect of tightening on joint stiffness is $p = 0.09$, indicating that Eq. (5) is statistically valid.

5. Relationship Between Joint Stiffness and Postural Stability

Based on motion measurement experiments in which subjects wore SSL, we investigated the relationship between an increase in trunk joint stiffness and postural stability. Subjects were five male adults who were instructed to hold a 6 kg weight and maintain their posture for 30 seconds, as in experiments in the previous section. The age, height, and weight of subjects are listed in **Table 2**. Using SSL in Section 2.1, the chest vest, waist belt, and elastic materials were adjusted for each subject so that no tightness or looseness existed when the subject stood erect. Setting six conditions – combination of assist condition (with/without SSL) and lumbar flexion angles (30°, 60°, and 90°), measurements were taken for each condition three times in random sequence. Motion was captured by an optical motion capture system, and the body sway was recorded with a force plate (Kistler 9286A, 9865EiY28) as center of foot pressure (COP) at a sampling frequency of 120 Hz. COP were processed using a 0.02–10 Hz band pass filter.

We first discuss changes in postural stability for the entire body when wearing SSL. We used the standard deviation of COP to evaluate postural stability because it is generally used as an evaluation index [14]. Because the posture being measured tends to fluctuate in the forward and backward directions, we deal with only the sagittal direction. Results in **Fig. 12** show that standard deviation decreased when SSL was worn by subject A at flexion postures of 30°, 60° and 90°; for subject B at a flexion of 90°; and for subject E at flexion postures of 30° and 90°. Postures in which the body sway was reduced are those in which standard deviation without SSL was greater than other postures. These results suggest that such postures, exhibiting relatively low stability, have low stiffness, so postural stability is increased by wearing SSL.

Next, to touch on lumbar joint stiffness and stability, joint stiffness in the lumbar region is determined from Eq. (4). Lumbar stability was evaluated by the standard deviation of the lumbar flexion angle over 30 seconds. Results for all subjects and postures are plotted in **Fig. 13**. Regression analysis of all data gave the correlation coefficient between joint stiffness and standard deviation of swaying as −0.50, a negative correlation, with a significance level of 1%. Variance analysis of the regression equation yielded p-values of 0.13×10^{-2} for data when SSL was worn and 0.12×10^{-2} for data when SSL was not worn, indicating that lumbar stiffness contributes to postural stability under both condition. Posture thus tends to be more stable when the lumbar region has high stiffness regardless of whether the subject is assisted or not. This suggests that it is valid to use joint stiffness to express trunk stabilization. The effect of SSL is manifested

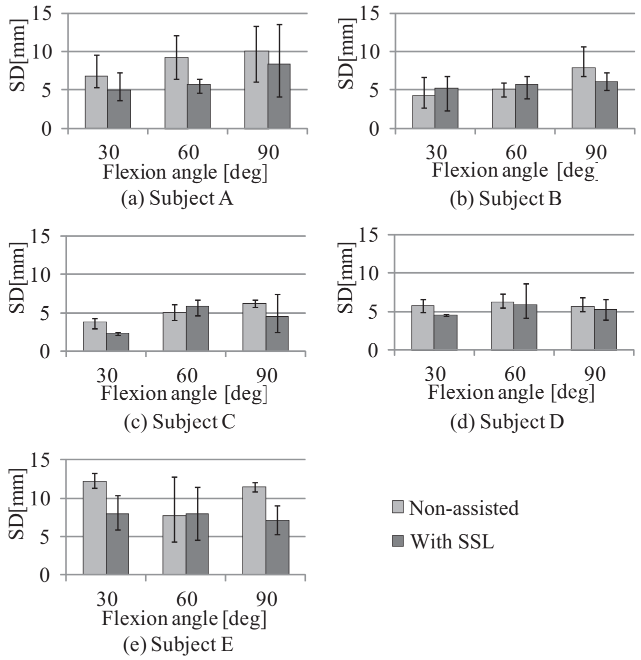


Fig. 12. Sagittal standard deviation of COP.

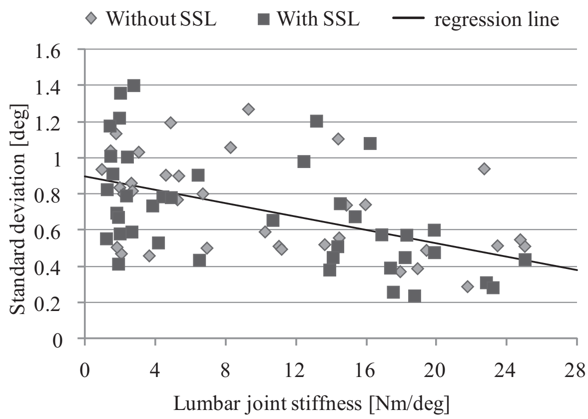


Fig. 13. Relationship between lumbar joint stiffness and postural stability.

in subjects who have low stability, so it may be the case that trunk stabilization is achieved by strengthening joint stiffness in those with low stability to make it approach that of highly stable people.

6. Conclusions

We have proposed a method for constructing a model of trunk stabilization generated by SSL based on the motion measurement. Devices such as the lumbar corset that apply pressure around the pelvis increase the support of the lumbar region and stabilize posture. We defined this as the trunk stabilization effect. Based on the hypothesis that minute movements change when assisted, we conducted experiments measuring the motion of subjects while they statically maintained a posture and obtained the following conclusions:

1. Using the estimated stiffness of the lumbar joint of the human link model, we have proposed a method for constructing a model expressing the relation between trunk muscle force, tightening pressure surrounding the lumbar region, and joint stiffness. We believe that this model is applicable to simulating the trunk stabilization for arbitrary assist force.

2. We have found a negative correlation between lumbar joint stiffness and the magnitude of postural sway, suggesting that joint stiffness is a valid parameter for expressing the trunk stabilization effect.

In the future, we hope to extend this model to deal with three-dimensional motion and also to one with universality that encompasses individual differences.

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Brief Biographical History:
2005- Assistant, Department of Health and Sport, Niigata University of Health and Welfare
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Main Works:
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Membership in Academic Societies:
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Main Works:
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- Japanese Society for Clinical Biomechanics
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