



Article

Modeling Joint Stiffness Change by Pelvic Tightening Based on Pelvic Alignment

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Abstract: This paper aims to develop a regression model that explains the relationship between changes in lumbar joint stiffness and pelvic alignment (posture or shape of the bones of the pelvis and lumbar spine) due to pelvic tightening. The proposed model is based on the hypothesis that lumbar joint stiffness increases with changes in pelvic alignment. The proposed model is based on experimentally measured stiffness values and pelvic alignment data sets. The stiffness of the lumbar joint was estimated by motion analysis using a motion-capture system. Ninety-six volunteers participated in the experiment to estimate stiffness values, and the validity of using lumbar joint stiffness as the output of the model was examined. The pelvic alignment was measured through X-ray images. Pelvic alignment was measured using radiographic images, and 25 volunteers participated. The Results section states that the amount of change in the posture of the sacrum relative to the pelvis and the curvature of the lumbar spine contributes to the change in lumbar joint stiffness. Future work will include FEM analysis to validate the overall hypothesis and the validity of applying the model to a group other than those who participated in the development of the model.

Keywords: joint stiffness; pelvic alignment; pelvic tightening



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1. Introduction

1.1. Background

Lumbar assistance by pelvic tightening is commonly applied as orthosis such as pelvic belts and corsets. Its assisting effect is widely recognized [1–3] and has been used conventionally. For example, Lee et al. [4] investigated the mechanical effects of pelvic tightening on the human body using radiographic images and motion-capture measurement. Investigating changes in joint stiffness due to tightening, Mens [5] and Damen [6,7] reported the stabilization of the sacroiliac joint based on ultrasonic measurements. Moreover, Snijders [8] reported a decrease in transversus abdominis muscle activity due to the trunk-assisting effect of the pelvic belt. Furthermore, our previous study elucidated the mechanism of reducing lumbar burden using pelvic belts and presented that lumbar burden reduction is the result of the change of the lumbar joint stiffness [9,10].

However, it is not clarified how the stiffness of the lumbar spine joints is changed. No precedent proves the process of lumbar support by pelvic belts based on mechanics, and previous studies only reported on hypotheses based on measurements [11–13]. The assistance effect of pelvic belts varies among individuals [2,9,14], and in the majority of cases, the lumbar burden is reduced by tightening, while in some cases the lumbar burden is unchanged or increased. Moreover, while the pelvic belt is expected to reduce the lumbar burden, there are concerns about impaired blood circulation and muscle atrophy due

to prolonged wear [15,16]. Pelvic belts should only be prescribed to patients for whom they are expected to be sufficiently effective, and a means of ascertaining which patients will have reduced lumbar burden is needed. Therefore, it is necessary to elucidate the mechanism of the change in lumbar joint stiffness (elastic coefficient) due to tightening and formulate its assistance effect. This paper aims to develop a regression model that explains the relationship between changes in lumbar joint stiffness and pelvic alignment (posture or shape of the bones of the pelvis and lumbar spine) due to pelvic tightening. The study considers the acting force on the pelvis by the tightening and proposes a hypothesis of a burden-reduction mechanism. The hypothesis predicts an increase in lumbar joint stiffness with the changes in pelvic alignment. This study also selects the alignment features that contribute strongly to the change in joint stiffness and inputs them into the model.

Prevenient studies demonstrated the effect of pelvic tightening on the posture and shape of the pelvis, in addition to the changes in motion and mechanical characteristics of the body [17]. Klima et al. [18] reported an enlargement of the upper pelvic opening, anteversion of the sacrum, and vertical translation of the sacrum relative to the pelvis using the pelvis and sacrum of a cadaver. Therefore, this study expects that changes in the shape and posture of the pelvis (i.e., alignment) contributed to the increase in frictional resistance on the surface of the sacroiliac joint or change in the elastic equilibrium point of lumbar due to the extension of the intervertebral ligaments, and caused an increase in joint stiffness.

1.2. Hypothesis about Lumbar Burden Reduction

Based on previous work [9,10], this study proposes the following hypotheses of the mechanism for reducing lumbar burden by pelvic tightening:

- (1) Changes in pelvic alignment due to tightening;
- (2) Increased lumbar joint stiffness due to changes in alignment;
- (3) Reduction in lumbar joint flexion due to the change in joint stiffness [10];
- (4) Reduction in lumbar joint torque (lumbar burden) [9].

The details of the processes presented in (1) to (4) are described below. The degree of freedom on the sagittal plane of the spine, sacroiliac joint, and hip joint is represented by the two-dimensional, three-link model (Figure 1). The pelvic belt is worn below the superior anterior iliac spine and above the greater trochanter, such that the sacrum and hip bone can be tightened.

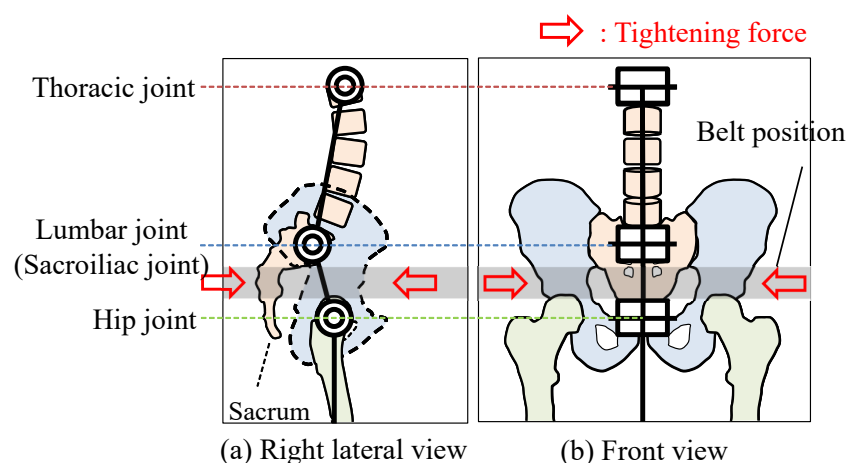


Figure 1. Hypothesis on reduction of the lumbar burden by tightening the pelvis: (a) Lateral view of the pelvis and lumbosacral vertebrae; (b) front view of the pelvis and lumbosacral vertebrae.

First, the compressive force from the front–back direction of the human body by pelvic tightening is considered, as shown in Figure 1a. The tightening force acts on the sacrum from the back and the sacrum is pressed against the pelvis. It causes backward tilting of the sacrum, as shown in Figure 2a. In addition, the lumbar vertebral body changes its posture with the tilt of the sacrum, such that the lordotic curve of the lumbar spine is relieved [17]. The belt also compresses the pelvis from the lateral direction of the human body, which causes a change in the shape of the pelvis; that is, the upper part of the pelvis closes, whereas the lower part widens, as shown in Figure 2b [17,18].

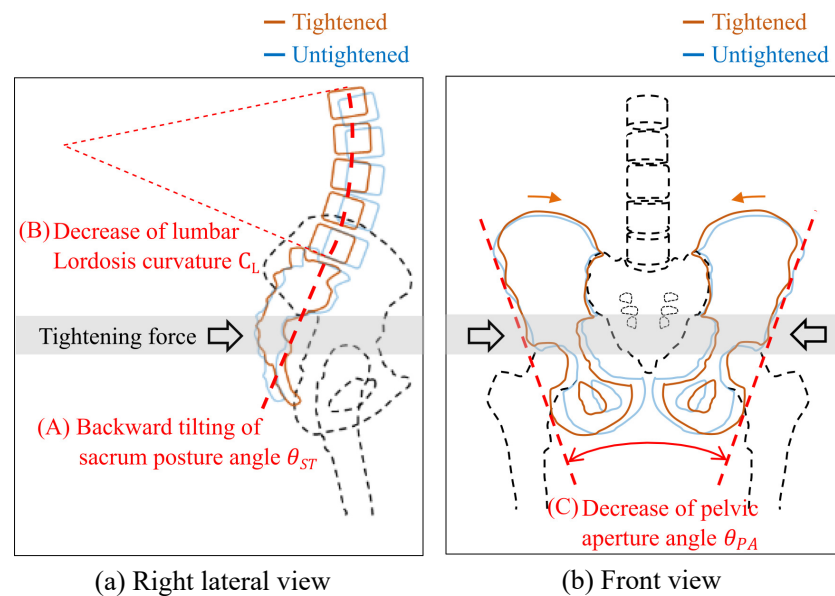


Figure 2. Change of the lumbosacral alignment by tightening: (a) Backward tilting of sacrum posture angle θ_{ST} and alleviation of lumbar curvature C_L ; (b) close of the pelvic aperture angle θ_{PA} .

This study inferred that such changes in the alignment of the lumbosacral vertebrae and pelvis contribute to changes in lumbar joint stiffness during forward bending. The compression of the sacrum to the pelvis indicates compression on the articular surface of the sacroiliac joint. The compression of the articular surface increases frictional resistance on the movement of the joint and increases the stiffness of the sacroiliac joint (lumbar joint in Figure 1). The closure of the upper part of the pelvis is also considered to compress the surface of the sacroiliac joint and increase its stiffness. This study also expects that relaxation of the lumbar lordosis curve changes the elastic equilibrium angle of the lumbar joint due to the extension of the intervertebral ligaments such as ligamenta flava, interspinal ligaments, and supraspinal ligaments, which are attached to each vertebral body.

An increase in lumbar joint stiffness suppresses the flexion of the lumbar joint, and hip joint flexion replaces the flexion of the lumbar joint (Figure 3). As a result, the moment arm from the lumbar joint to each center of gravity in the upper body is shortened, whereas the torque exerted at the lumbar joint to support the upper body is reduced. Moreover, the torque at the lumbar joint is exerted by the contracting force of the back muscles such as the erector spinae muscles. The back muscles are attached along the vertebral body, and the contraction force of these muscles mainly contributes to the compression of the intervertebral disc [19]. In other words, a decrease in the torque exerted by the lumbar joint indicates a reduction in the compressive force on the intervertebral disc (lumbar burden).

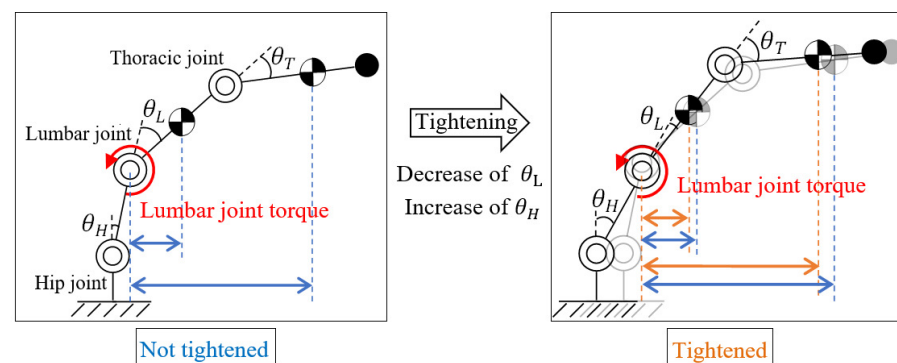


Figure 3. Posture correction through pelvic tightening.

Our previous study elucidated the changes in lumbar joint stiffness described in (3) of the hypothesis [10] and the lumbar burden reduction due to the posture correction described in (4) [9] based on motion measurements and the mechanical model. On the other hand, the change in pelvic alignment described in hypothesis (1) has long been verified by experiments using skeletons extracted from dead bodies [18] and by FEM analysis [17]. While FEM analysis is commonly used to evaluate lumbar strain and pelvic alignment [20–22], and has the advantage of directly evaluating changes in frictional resistance at the sacroiliac joint by analyzing stresses produced when forces are applied to the sacrum, these methods and models do not consider the volume of subcutaneous fat and muscle. Because the skeletal compression from belt tension is relieved by the subcutaneous fat and muscle attached to the pelvic surroundings, these methods contain modeling errors. Particularly in FEM analysis, the friction coefficient of the joint surface and the elastic equilibrium point must be set appropriately to evaluate the joint stiffness value, making precise analysis difficult. In contrast, this study is based on actual measurements of joint stiffness values and pelvic alignment in living bodies, and can model the phenomena that commonly occur in volunteers. Therefore, as a phenomenon supporting hypotheses (1) and (2), this study models the increase in joint stiffness due to changes in pelvic alignment.

2. Materials and Methods

2.1. Volunteers

A total of 96 volunteers participated in the lumbar and hip-joint stiffness measurement. Participants consisted of 52 males and 44 females; their age ranged from 20 to 60 years old. The mean and standard deviation of height was 1.67 ± 0.08 m, weight was 63.4 ± 11.6 kg, age was 36.0 ± 13.3 , and BMI was 22.8 ± 3.5 . These volunteers were recruited as healthy Japanese adults who did not have lower-back pain. Volunteers signed a consent form of their own volition after receiving an explanation of the measurements before the experiment. Data were not used for any purpose other than the study described to the volunteers, and data were recorded anonymously so that they could not be linked to individual volunteers. All experiments were conducted during the daytime and took approximately 3 h per person.

In the pelvic alignment measurements, participants included a total of 24 adults (20 males and 4 females) in standing posture and 25 adults (20 males and 5 females in their 20s and 60s) in forward bending. The number of volunteers in the standing position is small because volunteers with unclear imaging of the hip bone, which is necessary for alignment measurement, were excluded. These volunteers were part of the 96 volunteers who participated in the measurement of joint stiffness and who consented to exposure by radiography. The mean and standard deviation of height was 1.70 ± 0.05 m, weight was 66.9 ± 10.3 kg, age was 39.5 ± 13.5 , and BMI was 23.4 ± 3.1 . Pelvic alignment measurements were also taken during the daytime and took approximately 30 min per person.

The experiments in this study were approved by the institutional review board (Ethics Review Committee concerning Engineering Research with Human Subjects in the Graduate School of Information Science and Technology, Hokkaido University, No. 352).

2.2. Estimation of Joint Stiffness

The estimation of joint stiffness was based on motion analysis using a motion-capture system (MAC3D System, Motion Analysis Corp., 6085 State Farm Drive Suite 100, Rohnert Park, CA 94928, USA). The motion to be measured was the extension of the upper body from the forward bending posture to the upright posture, as shown in Figure 4a. During this motion, the volunteer did not hold any load at hand. The initial forward bending posture of the target motion was made so that the tilt angle on the body surface near the 12th thoracic vertebra was 45 degrees. Before the experiment, each volunteer performed a 45 degree forward bending posture and recorded the position of their hand. During measurement, a table was placed at the recorded height, and the initial posture of the motion was defined by the volunteer touching the table. Volunteers were instructed not to bend the knee joint during the movement, which allowed all volunteers to perform a constant movement, regardless of leg length. The target motion was measured 10 times in each of the conditions of not wearing the pelvic belt and wearing the pelvic belt and applying a tightening force. In the pelvic-belt-wearing condition, the belt tension was set at 80 N. The belt tension was adjusted by Active Corset [9,10]. It is known that the assistance effect of the pelvic belt saturates at a tension of 50 N [23,24], but it was expected that some volunteers with greater body weight or height would require a higher tightening force. Therefore, in order to provide sufficient assistance to all volunteers, the belt tension was set at 80 N, the upper limit of the Active Corset's output. The width of the belt was 50 mm, and the position of the belt was below the superior anterior iliac spine and above the greater trochanter.

Figure 4b of the mechanical model that represented the motion of the body on the sagittal plane is used to estimate the stiffness values k_L and k_H of the lumbar and hip joints. The joint stiffness values k_L and k_H are obtained from the motion equation (Equation (1)) of the mechanical model. The subscript x in Equation (1) represents each joint and takes either H (hip joint) or L (lumbar joint). Additionally, θ_x denotes the flexion angle of joint x , whereas θ_{x0} pertains to the elastic equilibrium point of joint x . The joint angle $\theta_{x(i)}$ was measured directly from the motion capture, and the angular velocity $\dot{\theta}_{x(i)}$ and angular acceleration $\ddot{\theta}_{x(i)}$ were obtained by time differentiating the joint angle.

$$M(\theta_{x(i)})\ddot{\theta}_{x(i)} + c(\theta_{x(i)}, \dot{\theta}_{x(i)}) + g(\theta_{x(i)}) + k_x(\theta_{x(i)} - \theta_{x0}) = \tau_{mx(i)} \quad (i = 1, 2, 3, \dots, s) \quad (1)$$

The suffix i represents the time series index of the measured data, and s represents the measured data point. Equation (1) consists of an inertial term $M_x(\theta_{x(i)})\ddot{\theta}_{x(i)}$, a centrifugal/Coriolis force term $c(\theta_{x(i)}, \dot{\theta}_{x(i)})$, a gravity term $g(\theta_{x(i)})$, an elastic term $k_{x(i)}(\theta_{x(i)} - \theta_{x0(i)})$, and a muscle exertion torque term $\tau_{mx(i)}$. All these terms can be subsequently calculated, except for the muscle exertion torque $\tau_{mx(i)}$ directly from the joint angle $\theta_{x(i)}$, angular velocity $\dot{\theta}_{x(i)}$, and angular acceleration $\ddot{\theta}_{x(i)}$.

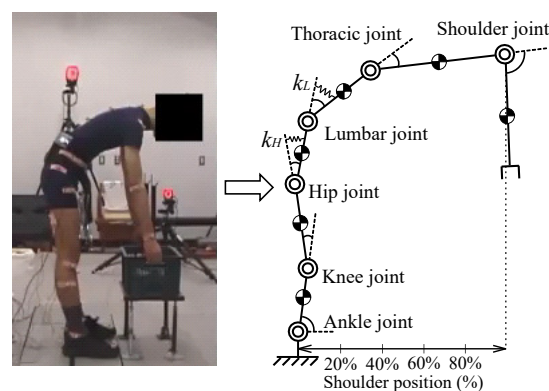
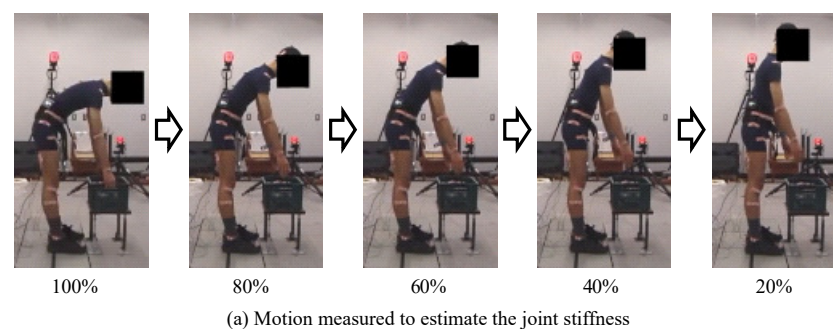
The muscle exertion torque $\tau_{mx(i)}$ was eliminated using Equation (2). This is because the latency of the spinal reflex of the erector spinae muscles for postural maintenance was at least 30 ms. The reason for Equation (2) is that there was no change in muscle activity of the lower-back between three consecutive samples. WATANABE [25] measured the latent time for the spinal reflex of erector spinae muscles by using electromyogram and reported that the latency was 70–90 ms in most cases, with a minimum of 30 ms. Thus, for at least 30 ms, the spinal reflex caused by the posture change during the measured motion does not change the back muscle torque. Furthermore, the change in joint angle during 30 ms was minimal, with an average of 0.008 rad for the hip joint and 0.007 rad for the lumbar joint among 96 volunteers who participated in the measurement. The change in posture during

this period was minute, and the change in muscle torque was expected to be sufficiently small for Equation (2) to hold.

$$\tau_{mx(i)} = \tau_{mx(i-3)} \quad (i = 1, 2, 3, \dots, s) \quad (2)$$

The measurement period of motion capture was 10 ms, and there was no change in muscle exertion torque between the three samples (30 ms). Therefore, Equation (3) can be obtained by taking the difference between $i = n$ and $i = n - 3$ in Equation (1) and applying Equation (2). In Equation (3), all terms on the right side can be calculated directly; thus, the stiffness values of the lumbar and hip joints can be obtained by the regression analysis using the least squares method [10].

$$\begin{aligned} k_x(\theta_{x(n)} - \theta_{x(n-3)}) = & M(\theta_{x(n-3)})\ddot{\theta}_{x(n-3)} - M(\theta_{x(n)})\ddot{\theta}_{x(n)} \\ & + c(\theta_{x(n-3)}, \dot{\theta}_{x(n-3)}) - c(\theta_{x(n)}, \dot{\theta}_{x(n)}) \\ & + g(\theta_{x(n-3)}) - g(\theta_{x(n)}) \quad (n = 1, 2, 3, \dots, s) \end{aligned} \quad (3)$$



(b) Mechanical model for estimating the joint stiffness

Figure 4. A motion measurement experiment was conducted to estimate the joint stiffness: (a) Target motion is the extension of the upper body from a forward bending position with the 12th thoracic vertebrae tilted at 45 degrees to a standing position; (b) lumbar joint and hip joint stiffness values are estimated by inputting the measured joint angles into a mechanics model.

The lumbar joint has non-linear elastic characteristics so that the process of the target motion was divided into several sections, and joint stiffness was calculated for each section. To divide target motion, a scale called shoulder position was introduced. Shoulder position expresses the degree of forward bending of the upper body and is calculated as the horizontal distance from the base of the model to the midpoint of both shoulders. Shoulder position reaches 0% at the standing posture during the extension of the model joint and 100% with forward bending at 45°, which is the initial posture of the target motion (Figure 4b). The lumbar (k_L) and hip (k_H) joint stiffness was calculated in four sections, in which the shoulder position values were 20–40%, 40–60%, 60–80%, 80–100%.

Additionally, this study calculates the joint stiffness ratio R using Equation (4). We have assessed that the pelvic belt increases the lumbar joint stiffness relative to the hip joint stiffness by evaluating the increase in joint stiffness ratio R [10].

$$R = \frac{k_L}{k_H} \quad (4)$$

The joint stiffness values estimated in this study include the stiffness values of any tissue acting as an elastic element in Equation (1). This includes resistance due to joint friction, the effect of trunk stiffness caused by increased intra-abdominal pressure from contraction of the rectus and transversus abdominis muscles, and the elasticity of the back muscle groups and lumbar intervertebral ligaments. Among these, the resistance due to joint friction was expected to change by correcting the alignment by the tightening force, and to affect the stiffness value of the lumbar joint k_L .

2.3. Measurement of Pelvic Alignment

This section describes the method used to measure pelvic alignment, which is the input information of the model. X-ray images have generally been used for the measurement of pelvic alignment features [4,26–32]; therefore, our study also used this methodology accordingly.

This study measured the following alignment information, which is known to be changed by tightening forces [17,18]:

- (A) Posture angle of the sacrum relative to the pelvis θ_{ST} (Figure 2a);
- (B) Curvature of the lumbar lordosis C_L (Figure 2a);
- (C) Pelvic aperture angle θ_{PA} (Figure 2b).

Regarding posture angle (A), when the pelvis is tightened, the tightening force acts on the bones from the belt in the front–back direction of the body, as shown in Figure 2a. The force acts on the lower part of the sacrum and tilts the sacrum backward with respect to the pelvis, with the lumbar joint as the axis of rotation [17].

Next, in term of the curvature (B), the study expects that the alignment of the vertebral bodies connected to the top of the sacrum changes due to the tilting of the sacrum. The lumbar spine curves toward the front of the body. However, when the sacrum tilts backward, lumbar spine lordosis is alleviated.

For aperture angle (C), on the coronal plane, the tightening force acts on the hipbone in the horizontal direction (Figure 2b). The force deforms the pelvis in a direction that reduces the distance between the left and right superior anterior iliac spines [17,18].

To measure the posture angle (A), an image of the sagittal plane taken from the left lateral side of the human body is used (Figure 5a,b). This measurement was performed for each of the standing and forward bending postures. To obtain the relative posture angle of the sacrum with respect to the pelvis, the inclination angles of the sacrum and pelvis are measured. A line that connects the superior anterior iliac spine and pubic symphysis is drawn to measure the pelvic inclination angle (θ_P). To measure the sacral inclination angle (θ_S), another line that connects the central part of the sacrum and sacral cornu is drawn. Based on θ_P and θ_S , the posture angle of the sacrum relative to the pelvis θ_{ST} is calculated using Equation (5).

$$\theta_{ST} = \theta_S - \theta_P \quad (5)$$

To obtain the curvature (B), images of the standing and forward bending postures taken from the sagittal plane were used as for the posture angle (A) measurement. Each of the five lumbar vertebrae is complemented by a quadrangle to determine the position of the barycenter (Figure 5a,b). The curvature of lumbar lordosis C_L is calculated as the inverse of the curvature radius r of the arc approximated from five barycenter points.

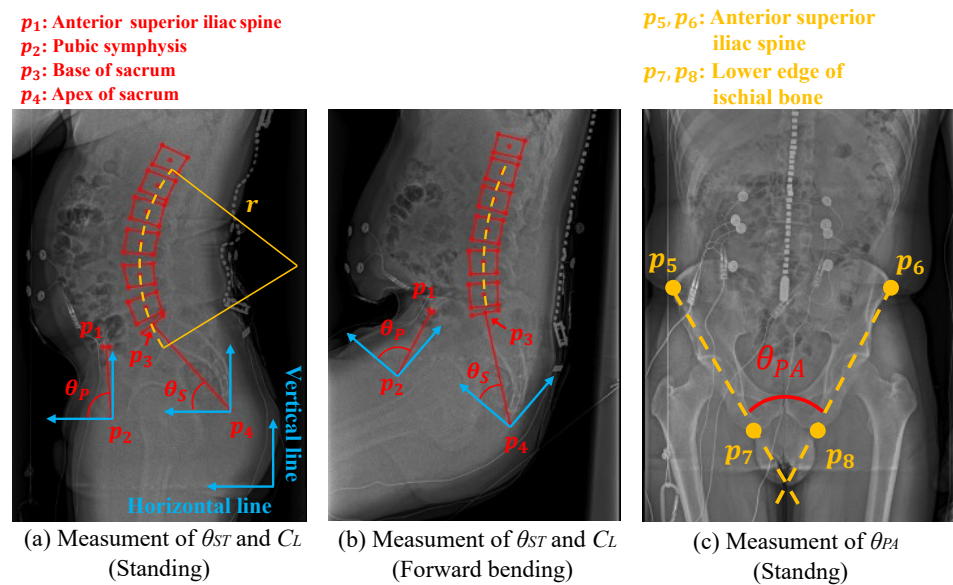


Figure 5. Measurement of pelvic alignment features using X-ray images: (a) Measurement of sacrum tilt relative to the pelvis θ_{ST} and lumbar curvature C_L in standing posture; (b) measurement of sacrum tilt relative to the pelvis θ_{ST} and lumbar curvature C_L in forward bending; (c) measurement of pelvic aperture angle θ_{PA} .

The alignment feature of aperture angle (C) was measured only during standing because the pelvis cannot be imaged from the front of the human body during forward bending (Figure 5c). To evaluate the opening in the upper part of the pelvis, the pelvic aperture angle is measured. A line connecting the superior anterior iliac spine and the lower end of the ischium is drawn on the left and right hipbones. The angle between the two lines is considered the pelvic aperture angle (θ_{PA}).

$$\Delta\theta_{ST} = \theta_{ST}^T - \theta_{ST}^U \quad (6)$$

$$\Delta C_L = C_L^T - C_L^U \quad (7)$$

$$\Delta\theta_{PA} = \theta_{PA}^T - \theta_{PA}^U \quad (8)$$

where, θ_{ST}^U : sacrum tilt angle relative to the pelvis without tightening; θ_{ST}^T : sacrum tilt angle relative to the pelvis with tightening; C_L^U : curvature of the lumbar spine without tightening; C_L^T : curvature of the lumbar spine with tightening; θ_{PA}^U : pelvic aperture without tightening; θ_{PA}^T : pelvic aperture with tightening.

Equations (6)–(8) are used to determine the variation of each alignment feature $\Delta\theta_{ST}$, ΔC_L , $\Delta\theta_{PA}$. The belt tension of pelvic tightening was 80N.

3. Results

3.1. Model Formulation

To model the change in lumbar joint stiffness based on the hypothesis, confirming that the increase in joint stiffness ratio R is accompanied by a change in lumbar joint stiffness is necessary. Therefore, the degree of change in lumbar joint stiffness is Δk_L , which is the numerator of the right-hand-side term in Equation (4), and the amount of change in the inverse of the hip joint stiffness value corresponding to the denominator, susceptibility $\Delta \frac{1}{k_H}$, is calculated using Equations (9) and (10). We evaluate the contribution of Δk_L and $\Delta \frac{1}{k_H}$ to the change in the joint stiffness ratio ΔR .

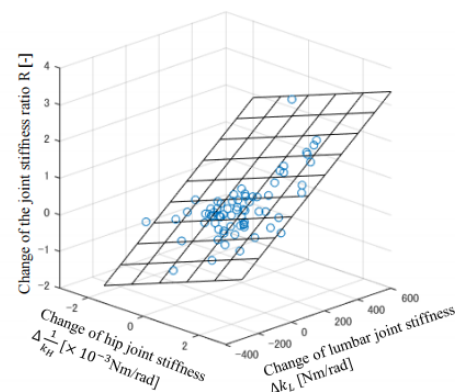
$$\Delta k_L = k_L^T - k_L^U \quad (9)$$

$$\Delta \frac{1}{k_H} = \frac{1}{k_H^T} - \frac{1}{k_H^U} \quad (10)$$

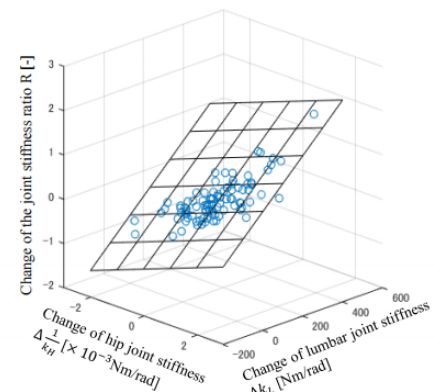
where, k_L^U : lumbar joint stiffness without tightening; k_L^T : lumbar joint stiffness with tightening; k_H^U : hip joint stiffness without tightening; k_H^T : hip joint stiffness with tightening.

We calculate the contribution rate of Δk_L and $\Delta \frac{1}{k_H}$ to the change in joint stiffness ratio ΔR from the standard partial regression coefficient obtained by multiple regression analysis. The contribution rates of Δk_L and $\Delta \frac{1}{k_H}$ are calculated in four sections, in which the shoulder position values were 20–40%, 40–60%, 60–80%, and 80–100%. In the section where the shoulder position is 0% to 20%, there is a discrepancy between the motion and the model due to oscillation of the acceleration term by swaying the volunteer's body.

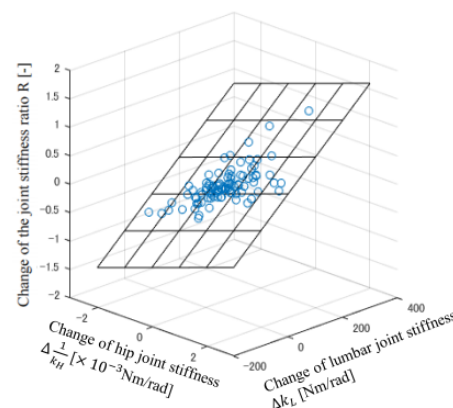
Table 1 shows the multiple correlation coefficients and the contribution rates of Δk_L and $\Delta \frac{1}{k_H}$ to the joint stiffness ratio change ΔR . Figure 6 shows scatter plots of joint stiffness ratio change ΔR for Δk_L and $\Delta \frac{1}{k_H}$ for 96 volunteers. The value of the multiple correlation coefficient is 0.9 or more in each section (Table 1). Thus, the study infers that the amount of change in joint stiffness ratio ΔR was caused by the change in stiffness of each joint. The contribution rate of Δk_L was 63.0% at a shoulder position value of 20–40%, 58.9% at 40–60%, 59.8% at 60–80%, and 54.4% at 80–100%. The contribution rate of Δk_L occupies the majority in all sections, and the joint stiffness ratio increased with the increase in lumbar joint stiffness, as hypothesized.



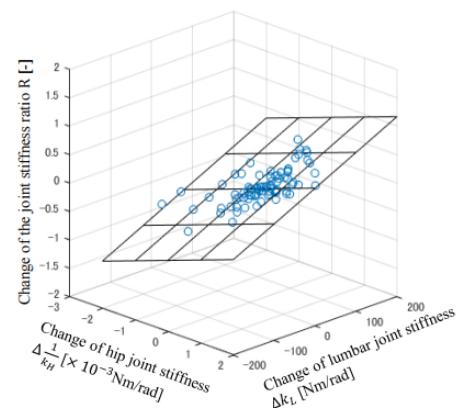
(a) Shoulder position = 20% to 40%



(b) Shoulder position = 40% to 60%



(c) Shoulder position = 60% to 80%



(d) Shoulder position = 80% to 100%

Figure 6. Correlation between the change in joint stiffness ratio R and $\Delta \frac{1}{k_H}$ and Δk_L : (a–d).

Table 1. Multiple regression analysis between the change in joint stiffness ratio ΔR and variance of joint stiffness Δk_L and susceptibility $\Delta \frac{1}{k_H}$.

Shoulder Position	20–40%	40–60%	60–80%	80–100%
Correlation coefficient	0.91	0.97	0.98	0.95
Contribution ratio of Δk_L	63.0%	58.9%	59.8%	54.4%
Contribution ratio of $\Delta \frac{1}{k_H}$	37.0%	41.1%	40.2%	45.6%

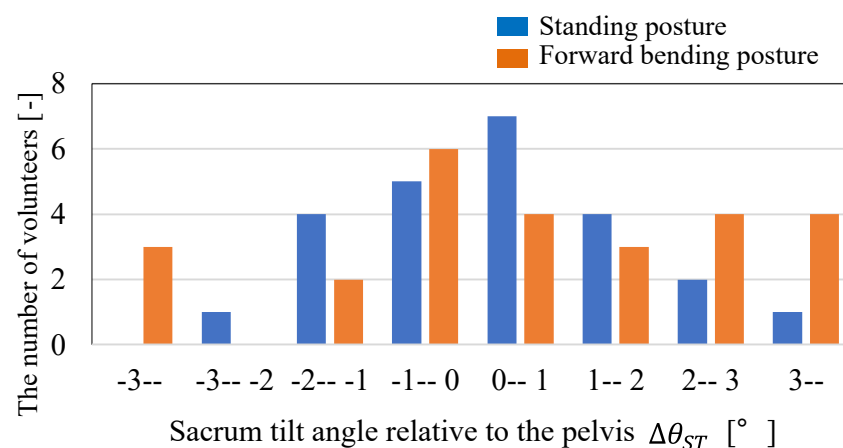
Conversely, Table 1 illustrates that $\Delta \frac{1}{k_H}$ has a certain level of contribution to the change in joint stiffness ratio ΔR . In other words, in volunteers whose ΔR value increased, hip joint stiffness tended to decrease at the same time as the increase in lumbar joint stiffness. The study infers that this reaction is caused by changes in the force of the muscle contraction for moving the hip joint. As the lumbar joint stiffness increases and the bending angle decreases, the bending angle of the hip joint increases to achieve the desired motion for body extension. At this time, the brain input changes the muscle exertion required to move the hip joint, which causes a change in the passive resistance of the joint.

The study focuses on the degree of change in lumbar joint stiffness, which is expected to increase due to changes in pelvic alignment and largely contributes to the change in joint stiffness ratio ΔR , as the output of the model. A regression model (Equation (11)) that expresses changes in lumbar joint stiffness and receives the changes in pelvic alignment as input is considered. The left-side value, which is the output of the model, represents the degree of change in lumbar joint stiffness. The coefficient α of the regression model should be determined.

$$\Delta k_L = \alpha x \quad (11)$$

3.2. Change in Pelvic Alignment

Figures 7–9 display a histogram of the number of people for each change in alignment feature. The sacral posterior tilt relative to the pelvis was confirmed in 14 of the 24 volunteers with valid data in the standing position, and 15 out of 25 volunteers in the forward bending position. The decrease in lumbar lordosis curvature was confirmed in 15 out of 24 volunteers in the standing position and 15 out of 25 volunteers during forward bending. The decrease in pelvic aperture angle, which was measured only during standing, was observed in 8 out of 24 volunteers.

**Figure 7.** Histogram of change in sacrum tilt angle relative to the pelvis.

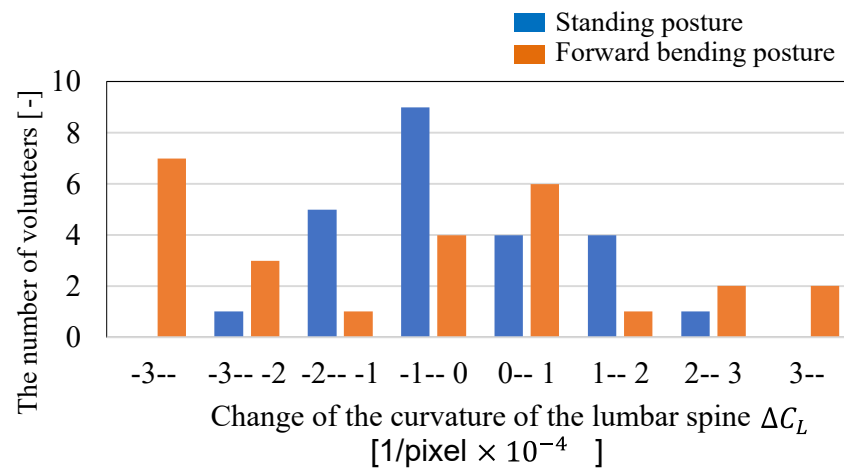


Figure 8. Histogram of change in curvature of lumbar spine.

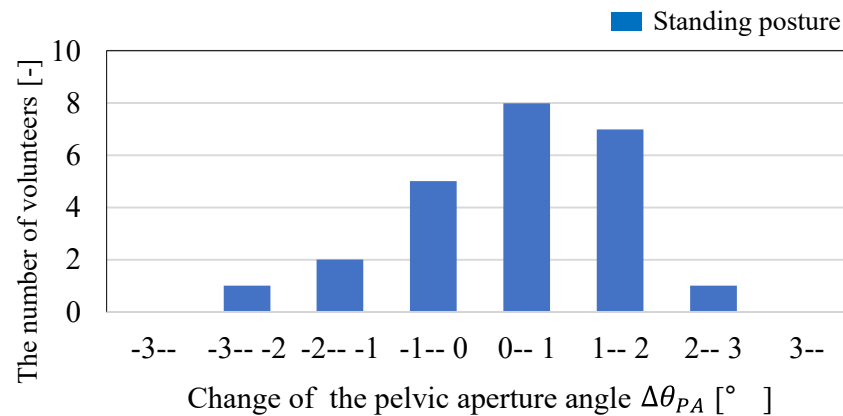


Figure 9. Histogram of change in pelvic aperture angle.

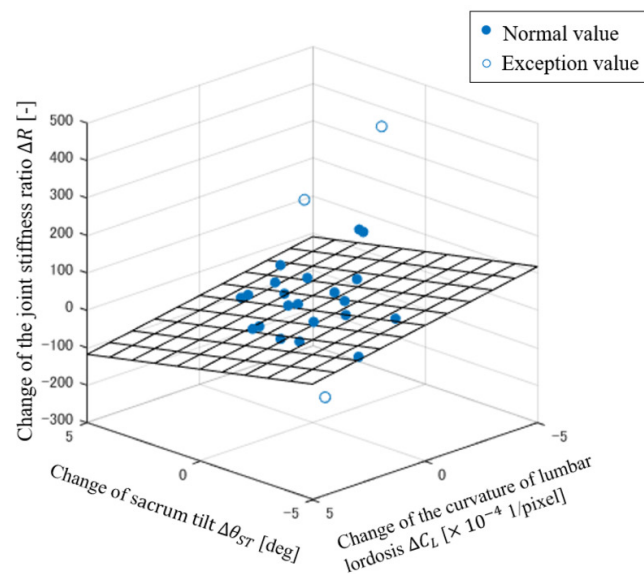
3.3. Modeling the Changes in Lumbar Joint Stiffness

The features to be used were selected as input to the model in Equation (11) from the pelvic alignment features of (A) to (C). Multiple regression analysis was performed using the change in lumbar joint stiffness as the objective variable and the degree of change in each alignment feature as the explanatory variable. The alignment feature is selected using the contribution rate calculated from the standard partial regression coefficient. (A) and (B) are used in the forward bending posture because the alignment features in (C) are measured only in the standing posture. Lumbar joint stiffness is estimated in the section that has a width of 10% in the shoulder position value. This section is within the range where the shoulder position value is from 0% to 30%, and lacks estimation failure due to oscillation of the body barycenter or fault of motion-capture measurement. In forward bending, the range where the shoulder position value was within 80% to 100% was used.

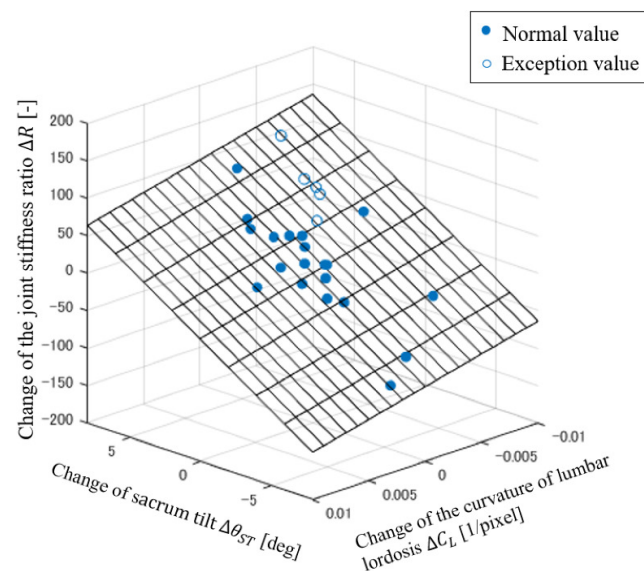
Table 2 provides the result during standing and forward bending with the partial regression coefficient of each change in alignment and the contribution rate of each variable. The value of the regression coefficient is attached with a symbol indicating the result of the *t*-test. During standing, the contribution rates to the change in lumbar joint stiffness are 47.3%, 43.2%, and 9.6% for changes in the sacral posture angle $\Delta \theta_{ST}$, lumbar lordosis curvature ΔC_L , and pelvic aperture angle $\Delta \theta_{PA}$. The study proposes that two features, $\Delta \theta_{ST}$ and ΔC_L , provide major contributions. During forward bending, $\Delta \theta_{ST}$ significantly contributes to the change in lumbar stiffness ($p < 0.01$) at a contribution rate of 85.8%. Conversely, the contribution rate of ΔC_L is small, at 14.2%. The study used the variations for sacral posture angle $\Delta \theta_{ST}$ and lumbar lordosis curvature ΔC_L , which have major contributions to the change in lumbar joint stiffness, as input to the model, and obtain Equation (12).

For Equation (12), model parameters α_1 and α_2 are determined by multiple regression analysis. The distribution of the data points used for model identification is illustrated by scatterplots for each posture (Figure 10). The data points whose residual errors from the regression plane are outside the 95% confidence interval were taken as exceptional values and excluded from the model identification. Table 3 displays the result of the regression analysis. In the standing position, the correlation coefficient was 0.28, whereas the values of parameters α_1 and α_2 were -2.62 and -1.08×10^5 , respectively. In the forward bending position, the correlation coefficient was 0.83, whereas the values of α_1 and α_2 were 12.49 and -3.60×10^3 , respectively.

$$\Delta k_L = \alpha_1 \Delta \theta_{ST} + \alpha_2 \Delta C_L \quad (12)$$



(a) Standing



(b) Forward Bending

Figure 10. The regression model represents the variation of lumbar joint stiffness Δk_L by inputting the variation of the sacrum tilt $\Delta \theta_{ST}$ and the lumbar lordosis C_L : (a) Relationship between stiffness change and pelvic alignment change in standing position; (b) relationship between stiffness change and pelvic alignment change in forward bending.

Table 2. Partial regression analysis between the variation of lumbar joint stiffness Δk_L and variation of pelvic alignment.

		Regression Coefficient (–)	Contribution Ratio (%)
Standing	$\Delta\theta_{ST}$	–24.50	47.3
	ΔC_L	-2.73×10^5	43.2
	$\Delta\theta_{PA}$	–7.29	9.6
Forward bending	$\Delta\theta_{ST}$	13.72 *	85.8
	ΔC_L	-3.65×10^4	14.2

*: $p < 0.01$.**Table 3.** Regression analysis between changes in lumbar joint stiffness Δk_L and sacral tilt angle $\Delta\theta_{ST}$.

	α_1	α_2	Correlation Coefficient (–)
Standing	–12.62	-1.08×10^5	0.28
Forward bending	12.49 *	-3.60×10^3	0.83

*: $p < 0.01$.

4. Discussion

This study hypothesized that posterior tilting of the sacral (increase in $\Delta\theta_{ST}$) increases in lumbar joint stiffness. Therefore, parameter α_1 was expected to obtain a positive value. Similarly, the lumbar lordosis curvature was expected to decrease with the increase in lumbar joint stiffness, and α_2 should obtain a negative value. In the forward bending position, lumbar joint stiffness increased with the changes in alignment, as expected. Table 2 shows that the sacral posture angle mainly included changes in stiffness for lumbar bending. However, in the standing position, no significant correlation was observed, and α_1 obtained a negative value. We assume that this result is due to the backward tilting of the sacrum to the limit of motion range in the standing position. Thus, the study considered that joint stiffness increased because the tightening force applied to the pelvis and sacrum generated stress, which increased the frictional resistance of the joint surface. Nevertheless, the alignment remained the same because of the limit of the motion range of the sacroiliac joint.

Next, we discuss modeling errors due to the use of pelvic alignment information measured by X-ray images as input. The proposed method showed a significant relationship between the change in lumbar joint stiffness and the sacral tilt angle relative to the pelvis (PI angle) θ_{ST} in the forward bending posture. Figure 7 shows that most of the volunteers are within three degrees of the change in θ_{ST} , which raises concerns about the influence of measurement error on the model output. This point is discussed based on the measurement accuracy of X-ray images. Czubak-Wrzosek et al. [33] measured the PI angle (θ_{ST}) using X-ray images, and stated that the intraclass correlation coefficient (ICC), which represents intraobserver reproducibility, is at least 0.99 and the standard error of measurement (SEM) does not exceed 0.8 degrees. This means that the measurement error of $\Delta\theta_{ST}$, the amount of change in θ_{ST} due to pelvic tightening, is at most 1.62 degrees. From Table 3 in the manuscript, the coefficient α_1 on $\Delta\theta_{ST}$ is 12.49, so the estimation error of the model output, the change in lumbar joint stiffness, Δk_L is less than 20.2 Nm/rad. The distribution of stiffness value change Δk_L ranges from -117.4 Nm/rad to 105.7 Nm/rad for the volunteers in this study, and the proposed method is a model with a possibility of 9.1% estimation error for this range.

Czubak-Wrzosek et al. [33] obtained the standard error of the measurements from 93 measurements, so the standard deviation σ is 0.17. Based on this, we have verified the procedures (a) through (c) below to confirm that the measurement error does not affect the conclusions of this study:

- Input 93 randomly sampled simulated data $\Delta\tilde{\theta}_{ST}$ and $\Delta\tilde{C}_L$ instead of $\Delta\theta_{ST}$ and ΔC_L to the model and obtain $\Delta\tilde{k}_L$ as output;
- Add random noise to $\Delta\tilde{\theta}_{ST}$ within 3σ ($=0.51$) to obtain $\Delta\tilde{\theta}_{ST}^{3\sigma}$;
- Multiple regression analysis with $\Delta\tilde{\theta}_{ST}^{3\sigma}$ and $\Delta\tilde{C}_L$ and $\Delta\tilde{k}_L$.

$\Delta\tilde{\theta}_{ST}$ and $\Delta\tilde{C}_L$ are simulated data randomly sampled from within the maximum (3.76 degrees) and minimum (-2.79 degrees) values of $\Delta\theta_{ST}$ and maximum (2.5×10^{-4}) and minimum (-2.1×10^{-4}) values of ΔC_L , respectively. Even when random noise equivalent to the measurement error was added to $\Delta\theta_{ST}$, it was found that $\Delta\theta_{ST}$ had a significant correlation with stiffness change in the forward bending posture. The influence of measurement error in pelvic alignment is sufficiently small.

The data used to develop the proposed model consisted of 25 Japanese adults, including 20 males and 5 females. Male height ranged from 1.615 to 1.811 m and weight from 54.3 to 93.9 kg, while female height ranged from 1.503 to 1.649 m and weight from 53.2 to 68.6 kg. According to the 2018 National Health and Nutrition Survey [34] by the Japanese Ministry of Health, Labour and Welfare, the mean and standard deviation of height and weight of Japanese males were 1.680 ± 0.071 m and 67.3 ± 11.1 kg, respectively, while those of females were 1.544 ± 0.070 m and 55.9 ± 9.3 kg. Assuming that the distributions of height and weight follow a Gaussian distribution, 78.7% of Japanese males and 65.4% of Japanese females for height and 87.1% of males and 34.6% of females for weight are in the range where the volunteers who participated in this experiment are distributed, and are covered by the proposed model. We will need to examine the general applicability of this model to other groups in the future. However, the development of the proposed model is a challenge to explain the process by which pelvic belt tightening force changes joint stiffness from a biomechanical perspective, which has not been clarified previously. This is expected to be useful for future clinical practice and for studying the effectiveness of pelvic belts.

The practical contribution of the developed model is expected to be the preliminary diagnosis of the assisting effect of pelvic belts in clinical practice. The developed model provides outputs on the degree of change in lumbar joint stiffness when changes in pelvic alignment are input. In other words, by inputting alignment changes, the model can estimate the degree of assisting effect required through pelvic tightening. The individual differences were observed regarding the reduction of lumbar burden by pelvic tightening [2,9,14]. From the viewpoint of preventing blood-flow obstruction due to excessive tightening force [15], or muscle atrophy due to prolonged use of the belt [16], the use of pelvic belt should be avoided by those who do not obtain the lumbar burden reduction. Therefore, an advanced diagnosis of assistance effect is required. The preliminary estimation of the expected effect of wearing a pelvic belt is used to determine whether to use a pelvic belt in the treatment of lower-back pain.

In the present study, X-ray images were used to measure the pelvic alignment features. However, if such input information can be obtained by a simple measurement using a non-invasive method, then determining the applicability of pelvic tightening is possible through a preliminary examination before prescribing orthosis. The alignment features to be measured are the curvature of the lumbar lordosis and sacral posture angle relative to the pelvis, which were the input of the model. Tsuchiya [35] estimated lumbosacral alignment based on a musculoskeletal model using a bending sensor and nine-axis IMUs attached to the lumbar and upper body. Based on this method, the curvature of the lumbar lordosis can be measured from the body surface. In addition, the posture angle of the sacrum relative to the pelvis can be estimated based on the position of the barycenter of the upper body. Roussouly et al. [19] demonstrated that the upper-body posture from the seventh cervical vertebra to the fifth lumbar vertebra is dependent on the PI and SS angles. The inclination angle of the upper body appears as a change in the position of the body CoG. In our subsequent study, we will aim to estimate the sacral posture angle from the lumbar posture measured with IMU and CoG measured using a floor reaction force meter.

5. Conclusions

The study found that pelvic tightening increases the lumbar joint stiffness relative to the hip joint. The results demonstrated that the following alignment features largely contributed to the change in lumbar joint stiffness:

- Sacral tilting angle relative to the pelvis;
- Curvature of the lumbar lordosis.

Moreover, the study determined the regression coefficient of the two variables to the change in lumbar joint stiffness. A model that represents the change in lumbar joint stiffness during forward bending was developed, thus achieving the goal of this research. However, during standing, no significant relationship was observed between the changes in alignment and stiffness. These results confirm hypothesis (2), “Increased lumbar joint stiffness due to changes in alignment,” which is one of the unexplained hypotheses of the lumbar burden reduction hypothesis described in the introduction. This study found a tendency for stiffness to change under the influence of pelvic alignment, although it cannot directly prove the increase in frictional resistance of the sacroiliac joint surface due to changes in the posture of the sacrum. Future work should include FEM analysis of the stress increase and frictional resistance changes on the joint surfaces to verify our overall hypothesis.

The method proposed in this study enables quantitative estimation of the stiffness change caused by the pelvic belt. The proposed method can interpolate stiffness changes for 78.7% of male and 65.4% of female healthy Japanese adults based on the height distribution, and for 87.1% of male and 34.6% of female adults based on the weight distribution. However, its applicability to non-Japanese adults, those with lower-back pain, hip joint dysfunction, and children has not been confirmed, and further research is needed. By applying this method, the effectiveness of the pelvic belt can be determined without the long trial period that was necessary to determine the effectiveness of the pelvic belt. However, this method requires radiography, and there are obstacles to its practical use in small hospital facilities. Therefore, for the next task, we propose a method of estimating pelvic alignment by simple measurement from outside the body.

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