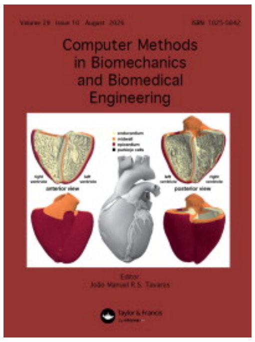

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Understanding the effect of lumbar lordosis angle on vertebral load distribution during walking

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ABSTRACT
Atypical sagittal spinopelvic alignment is correlated with exacerbating lower back pain (LBP). This study investigated the effects of simulated sagittal spinopelvic alignment *via* altered lumbar lordosis (LL) on lumbar vertebral contact forces during walking. A full-body OpenSim model with custom lumbar joints was developed to estimate lumbar vertebral loads for self-selected speed walking gaits of 18 healthy participants. Limited LL during walking augmented the resultant vertebral compressive and shear forces, and vertebral body compression. Excessive LL increased resultant vertebral shear forces, compression at facet joints and L5/S1 vertebral body, potentially progressing to different types of LBP.

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KEYWORDS
Lumbar lordosis; spinopelvic alignment; vertebral load; low back pain; walking; OpenSim

1. Introduction

Lower back pain (LBP) is a serious global health problem and constitutes a tremendous economic burden worldwide (Ma et al. 2014; Spencer et al. 2018). LBP has a multifactorial pathogenesis, with biomechanics playing a crucial role among the various risk factors (Stokes 1997; Lotz 1999). Altered lumbar lordosis (LL) has been found to increase the risk of developing low back pain (Chun et al. 2017; Sadler et al. 2017; Sugavanam et al. 2025).

Balanced sagittal spinopelvic alignment is a vital component of spinal function, which is necessary to maintain an energy-efficient posture (Mehta et al. 2012). Deviation of this alignment (e.g. abnormal LL (Mehta et al. 2012; Presciutti et al. 2018), atypical pelvic incidence (PI) (Mehta et al. 2012), or PI-LL mismatch (Presciutti et al. 2018)) has been reported for many spinal pathologies, including degenerative scoliosis, isthmic spondylolysis, degenerative spondylolisthesis, spinal deformity, and impacts outcomes after spinal fusion surgery. As an important anatomical parameter, spinopelvic alignment is speculated to play a potential role in the pathogenesis of LBP; however, associations between spinopelvic alignment and LBP have been inconsistent, especially between LL and LBP (Roncarati and McMullen 1988; Schroeder et al. 2013; Mirzashahi et al. 2023). In some studies, a reduced LL angle was a risk factor for LBP (Adams et al. 1999; Mirzashahi et al. 2023), and was usually reported in chronic LBP patients (Jackson and McManus 1994; Chaléat-Valayer et al. 2011; Król et al. 2017). Existing systematic reviews supported this relationship between LBP and decreased LL angle, while this association was not statistically significant in most of them (Sadler et al. 2017; Sugavanam et al. 2025), and included studies were highly heterogeneous (Chun et al. 2017; Sadler et al. 2017; Sugavanam et al. 2025). On the contrary, other studies showed exaggerated LL (Roncarati and McMullen 1988; Christie et al. 1995; Norton et al. 2004; Sorensen et al. 2015), or excessive anterior pelvic tilt (Roncarati and McMullen 1988; Tatsumi et al. 2019; Sugavanam et al. 2025), to be related to LBP. Many other investigations have failed to confirm any

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correlation between LL and LBP (Gautier et al. 1999; Murrie et al. 2003; Korovessis et al. 2004; Schroeder et al. 2013). Therefore, no firm conclusions could be made (Sugavanam et al. 2025); further exploration of the influence of sagittal spinopelvic alignment on LBP is necessary.

Atypical sagittal spinopelvic alignment (abnormal single indicator such as LL or PI, or a mismatch of these indicators) may result in abnormal lumbar load distribution or compensatory patterns (Mehta et al. 2012; Presciutti et al. 2018), which play a significant mechanical role in the LBP etiology (Adams and Hutton 1982). Musculoskeletal simulation offers a valuable tool to explore lumbar joint loads across different sagittal spinopelvic alignments, providing a biomechanical perspective independent of patient-specific factors during different activities. Several studies have investigated the effect of LL on lumbar joint loads during static upright standing or forward bending movements (Senteler et al. 2014; Naserkhaki et al. 2016; Müller et al. 2021), but its effect during walking remains unexplored. Walking is the most common repetitive daily task; therefore, it is important to understand how different LL angles affect lumbar joint loads during such a common activity.

This study aimed to investigate the effects of simulated sagittal spinopelvic alignments (LL angle) on lumbar vertebral joint contact forces and load distribution during walking in a cohort of adult participants without musculoskeletal health issues. To achieve this aim we: (1) developed a lumbar joint model that accounted for computing loads between the vertebral body and facet joint during walking; (2) evaluated the generated gait kinematics and lumbar loads using previous studies; (3) evaluated the effects of varying sagittal spinopelvic alignment on the computed lumbar vertebrae joint contact forces.

2. Methods

2.1. Participants information

Eighteen healthy adult participants (9 F and 9 M, 28 ± 5 years old, 1.70 ± 0.08 m, 66 ± 10 kg) participated in this study. Ethical approval was received from the University of Auckland Human Participant Ethics Committee (reference number 019911). Informed written consent was obtained from all participants. No participants had any known neuromusculoskeletal disorders or musculoskeletal pain (Christophy et al. 2012; Raabe and Chaudhari 2016). Inclusion criteria were adults aged 18 years and older, and exclusion criteria included previous lower back surgery, chronic specific/nonspecific LBP, and acute pain or any injury to the lower back in the past six months prior to data collection.

2.2. Data collection and post-processing

Twenty-seven reflective markers were placed on participants using hypoallergenic double-sided tape (Figure 1). The trajectories of markers were recorded at 200 Hz using a 12-camera optical motion capture system (Vicon Motion Systems Ltd., UK). Participants walked over three ground embedded force plates (Bertec, Columbus, Ohio, USA) to measure ground reaction forces at 1000 Hz. Participants performed one static and six walking trials at their self-selected speed. One gait cycle per trial in which the foot was in contact with a force plate was selected for inverse dynamics analysis. A gait cycle was defined as the period from initial contact of one foot to the subsequent initial contact of the same foot.

All captured data were synchronized, and marker trajectories were reconstructed using Vicon Nexus software (Version 2.12). The MOtoNMS toolbox (MATLAB Motion Data Elaboration Toolbox for Neuromusculoskeletal Applications) was employed to filter 3D marker position and ground reaction force data using a 4th-order Butterworth low-pass filter at 10 Hz (Mantoan et al. 2015). Data were rotated according to the OpenSim coordinate system, where the X-axis was perpendicular to the frontal plane pointing forward, the Y-axis was perpendicular to the transverse plane pointing upward, and the Z-axis was perpendicular to the sagittal plane pointing to the right.

2.3. Model development

To understand the load distribution between each lumbar vertebrae, the estimation of intervertebral joint loads at the facet and vertebral body were analyzed separately based on Raabe's full-body lumbar spine model (Raabe and Chaudhari 2016), we developed an OpenSim model and added new joint definitions between each lumbar vertebrae (from L1 to sacrum).

To resolve distinct anterior (vertebral body) and posterior (facet joint) lumbar vertebral loads, additional body frames were introduced. Firstly, the lumbar vertebrae joint from Christophy et al. (Christophy et al. 2012) defined the three rotational degrees of freedom (flexion-extension, lateral bending, and axial rotation) of lumbar vertebral joints between the

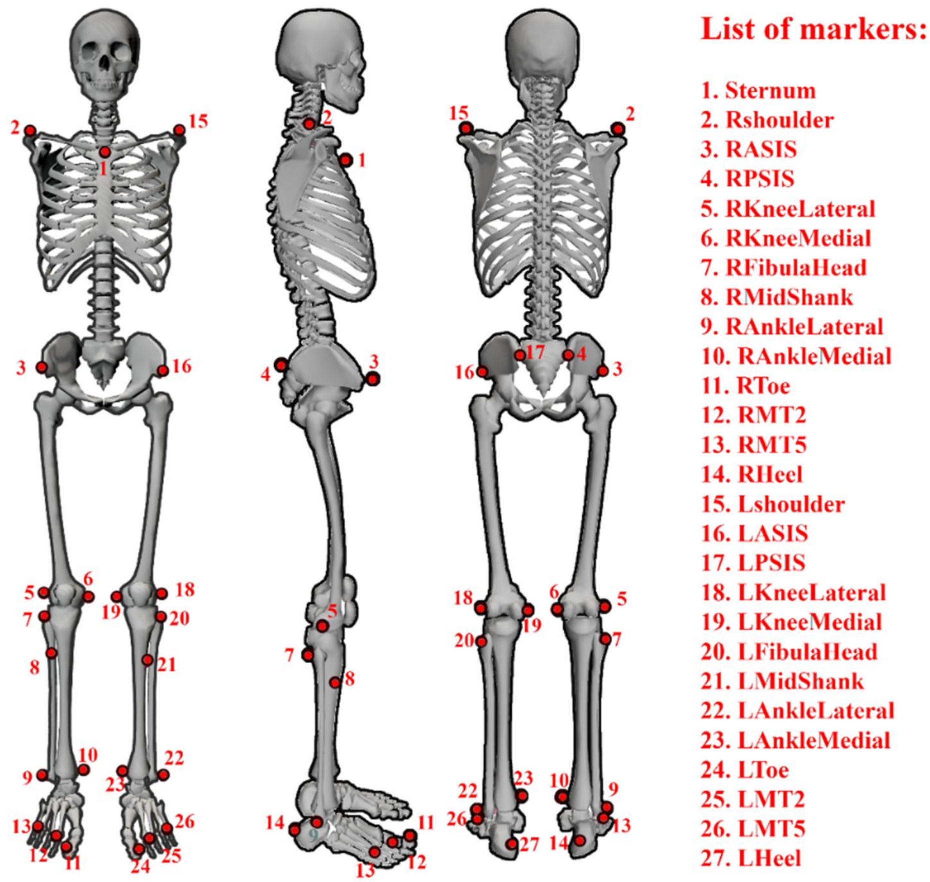

Figure 1. Marker placement locations.

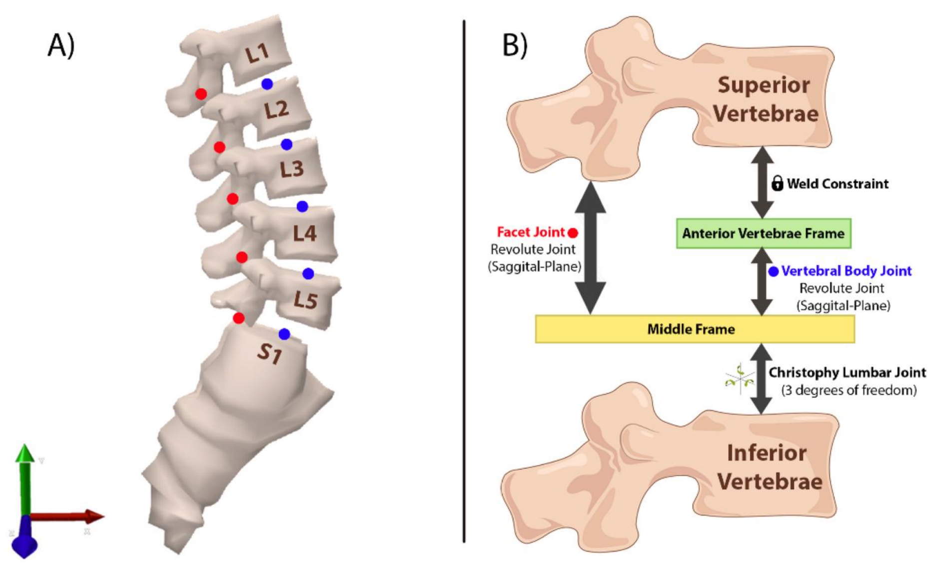

Figure 2. Graphical (A) and schematic (B) depictions of lumbar vertebral body/facet joint (anterior/posterior) structures in the OpenSim model. Both graphic and schematic are shown in the sagittal plane.

inferior vertebrae and the middle frame (Figure 2B, yellow box). Secondly, two hinge joints, with axes perpendicular to the sagittal plane, connected the middle frame to: 1) the anterior vertebrae frame through the vertebral body joint (Figure 2B, green box and Figures 2A,B blue dots); and 2) the posterior part of the superior vertebrae through the facet joint (Figures 2A,B, red dots). These two joints allowed the OpenSim model to compute the anterior and posterior loads of each lumbar vertebrae. Lastly, the anterior lumbar vertebrae frame was fixed to the superior vertebrae by a weld constraint (Figure 2B).

In the lumbar vertebral joint mechanism, while the anterior and posterior revolute joints individually could not resist sagittal-plane moments, they acted in parallel to share all loads transmitted between inferior and superior vertebrae. This resolved them as anterior and posterior contact forces, balancing net reaction forces and sagittal-plane moments across the lumbar vertebral joint (Abouhossein et al. 2011). Correspondingly, the vertebral body and facet joints were restricted to probe the joint loads without affecting mobility of the original lumbar joints defined by Christophy et al. (Christophy et al. 2012).

2.4. Lumbar lordosis angle variation

The developed full-body OpenSim model with custom lumbar intervertebral joints was used to compute the lumbar vertebral joint contact forces during gait. The model was scaled according to marker placement using the Scaling Tool in OpenSim. Filtered 3D marker trajectories and ground reaction forces during each gait cycle were used as input for the inverse kinematics and inverse dynamics steps in OpenSim. Since radiographic imaging, which is currently considered the gold standard for measuring lumbar curvature, is challenging to implement during dynamic activities, it is difficult to obtain accurate LL angles during walking from prior literature. Therefore, although the lumbar spine undergoes slight flexion during gait (Castillo and Lieberman 2018), we adopted the LL angle measured in the standing posture as a reference. Prior studies observed asymptomatic LL angles in a neutral upright position ranging from 28 to 66° with average LL angles varying from 43 to 49° (Bernhardt and Bridwell 1989; Wood et al. 1996; Kuntz et al. 2007). To explore the impact of different LL angles, the LL angles in our model were set to simulate hyperlordosis (85°, 75°, 65°), neutral lordosis (55°, 45°, 35°), and hypolordosis (25°, 15°, and 5°) (Müller et al. 2021). Therefore, before running the inverse kinematics pipeline, the lumbar flexion/extension range of motion was locked from 5° to 85° to vary the LL angle for each participant in the OpenSim model (Müller et al. 2021). The sagittal LL angle was set by measuring the angle between the extension line of the upper endplate of L1 and the upper endplate of the sacrum (S1) (Figure 3A) (Park et al. 2023).

To maintain a reasonable relationship between LL and other spinopelvic parameters during gait, we reduced the calculation weights of markers around the pelvis (RASIS, LASIS, RPSIS, LPSIS) (Figure 1) for each inverse kinematics process. Simultaneously, we increased marker weights on the shoulders and upper body (Sternum, Rshoulder, Lshoulder) (Figure 1) to make sure the center of gravity of the trunk aligned with the center of gravity of the pelvis in the sagittal plane during walking. Each generated gait kinematics dataset received approval from a physiotherapist, who ensured the kinematics were within a realistic physiological range. After computing joint kinematics with different LL angles, the Residual Reduction Algorithm was used to generate a smoothed set of kinematics that reduced dynamic inconsistency between the measured kinematic and kinetic data. Muscle activation and force results were calculated through the Static Optimization tool in OpenSim. The Joint Reaction Analysis tool was used to estimate the lumbar vertebral compression and resultant shear forces. All vertebral joint forces were expressed relative to the local coordinates of the inferior vertebra (e.g. the L5/S1 vertebral joint forces were expressed relative to the local coordinates of S1).

The OpenSim model output for each LL angle variation was: (1) average of peak adjacent vertebral resultant compressive forces (Figure 2, Christophy lumbar joint), (2) average of peak adjacent vertebral resultant shear forces (Figure 2, Christophy lumbar joint), (3) average of peak vertebral body compressive forces (Figure 2, vertebral body joint), and (4) average of peak facet compressive forces (Figure 2, facet joint) at the L1/L2 joint, L3/L4 joint and L5/S1 joint. All average data were calculated from three gait cycles for each participant.

2.5. Evaluation

Given the strong relationship between the LL and the other sagittal spinopelvic parameters, especially sacral slope (SS) (Figure 3A) (Barrey et al. 2007; Mehta et al. 2012; Król et al. 2017), the evaluation of our modified spinopelvic alignment was performed by comparing the relationship between outputted SS and LL to those from previous studies (Roussouly et al. 2005; Naserkhaki et al. 2016; Müller et al. 2021). The regression line representing the relationship between SS and LL was calculated in a least squares sense.

Evaluation for simulated resultant vertebral loads was performed in the case of the neutral LL of 45°, by comparing compressive forces and shear forces derived from the Christophy lumbar joint (output 1 and 2 above) between adjacent vertebrae to previous studies, including L4-L5 loads from Schäfer et al.

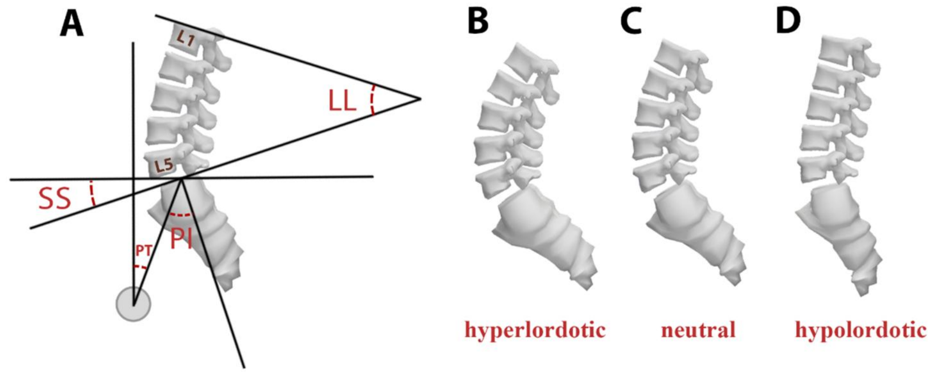

Figure 3. (A) Schematic depiction of sagittal lumbar lordosis (LL), sacral slope (SS) pelvic incidence (PI) and pelvic tilt (PT) and examples of LL curvature (B) hyperlordotic, (C) neutral, and (D) hypolordotic. LL: Normal inward curvature of the lumbar spine from the rostral L1 endplate to the rostral S1 endplate; SS: Angle between S1 endplate and a horizontal reference line; pelvic tilt (PT): angle between a vertical reference line and a line from the S1 endplate midpoint to the femoral head center axis; pelvic incidence (PI): a line subtended from the femoral head to a reference line perpendicular to the rostral S1 endplate midpoint (Park et al. 2023).

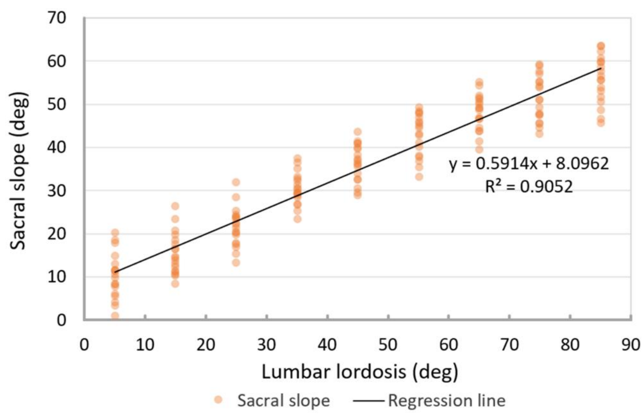

Figure 4. Sacral slope (SS) plotted against lumbar lordosis (LL) (orange points). The regression line is displayed in black and its equation and coefficient of determination (R^2) are stated in the annotations.

(2023) and L3-L4 loads from Cappozzo (1984). Simulated vertebral body and facet joint loads (outputs 3 and 4 above) were evaluated by comparison with former in-vivo experiments, specifically T12-L2 vertebral body loads from Rohlmann et al. (2014), and L2–L4 facet joint loads from Rohlmann et al. (1997).

3. Results

3.1. Model evaluation

The relationship between LL and SS showed a linear correlation between the two measurements for all OpenSim models created with varying degrees of LL ($SS = 0.59 \times LL + 8.10$, $R^2 = 0.91$) (Figure 4).

To evaluate our model, simulated lumbar loads for the neutral LL of 45° are presented in Table 1. The simulated maximum resultant compressive force at L4-L5 during walking was 878 N (±177), equivalent to 1.33 times body weight (BW). The resultant compressive force at L3-L4 was 856 N (±175), which equaled 1.3 BW. Regarding distribution between the lumbar vertebrae, the model predicted an average peak force of 898 N at the L1-L2 vertebral body during walking at a self-selected speed. Additionally, the average peak force at the L3-L4 facet joint was 148 N.

3.2. Effects of varying lumbar lordosis

The maximum intervertebral resultant compressive load consistently decreased with increasing LL angle (Figure 5A) independently of the functional tissue unit. Peak compressive forces were lower at the L1/L2 junction than L3/L4 and L5/S1. Resultant maximum shear load tended to decrease at the L1/L2 joint with increasing LL angle (Figure 5B). High shear loads were found at the L5/S1 level at the extreme LL angles (hyper and hypo lordosis) reducing to less than 100 N on average

Table 1. Average peak lumbar resultant compressive, sagittal shear force, and distributed compression between the vertebral body and facet joints (mean ± SD) against lumbar lordosis (LL) of 45°.

		L1–L2 peak force	L2–L3 peak force	L3–L4 peak force	L4–L5 peak force	L5–S1 peak force
Resultant force	Compression (N)	689 ± 145	813 ± 169	856 ± 175	878 ± 177	896 ± 177
	Shear (N)	141 ± 41	102 ± 35	62 ± 30	67 ± 27	75 ± 36
Distributed compression	Vertebral body (N)	898 ± 94	866 ± 121	842 ± 142	968 ± 128	1010 ± 126
	Facet (N)	233 ± 45	184 ± 55	148 ± 39	240 ± 82	350 ± 97

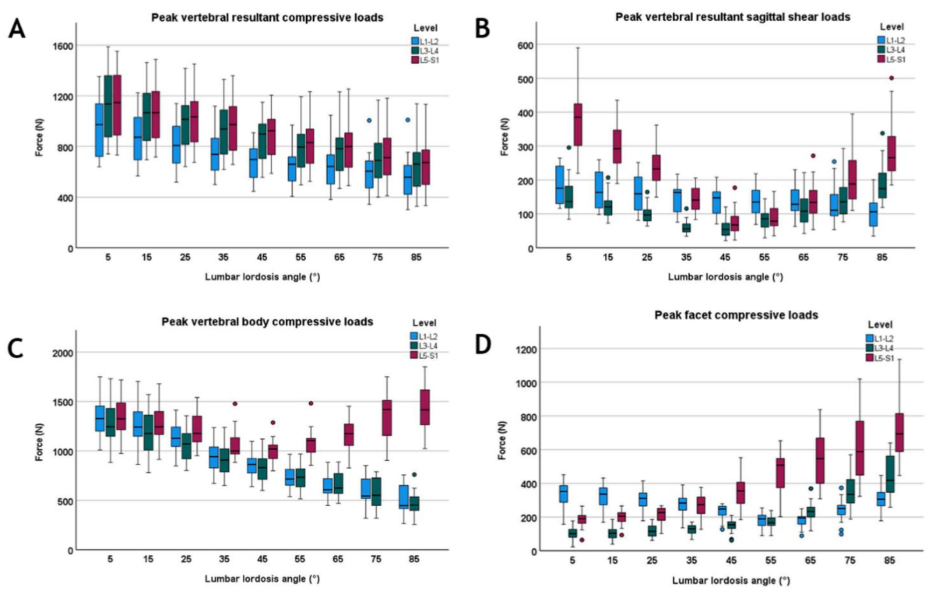

Figure 5. Average peak loads (N) for (A) vertebral resultant compressive force, (B) vertebral resultant sagittal shear force, (C) vertebral body compressive force, and (D) facet compressive force for each lumbar lordosis angle increment (from 5° to 85°). forces were computed at the L1-L2 level (blue), L3-L4 level (green) and L5-S1 level (red).

for a LL angle of 45° and 55° which was considered to be normal LL angle (Figure 5B). Peak vertebral body compressive force at the L1-L2 and L3-L4 joints consistently decreased with increased LL angle (Figure 5C). On the other hand, the L5-S1 vertebrae vertebral compressive force first decreased (from hypo to normal lordosis) and then increased (from normal to hyper lordosis) with the lowest load found at 35° LL angle (Figure 5C). Peak facet compressive loads increased with higher values of LL angles for L3/L4 and L5/S1 joints (Figure 5D). Peak facet loads were higher in the L1/L2 joints for the hypo lordosis angle and decreased until reaching 55° of lordosis and started to increase again to the hyperlordosis angles. Example loads data from a single participant during a single gait cycle is presented in Figure 6. The peak lumbar loads mostly occur shortly after heel strike within the gait cycle.

4. Discussion

While biomechanical consequences of abnormal spinopelvic alignment are known (Chun et al. 2017; Sadler et al. 2017; Sugavanam et al. 2025), limited research has explored the specific relationship between lumbar lordosis (LL) angles and changes in compressive and shear loads in the lumbar spine during walking. In the present study, we investigated the effects of simulated sagittal spinopelvic alignments (changes in LL angle) on lumbar vertebral joint contact forces and load distribution during walking in a cohort of adult participants without musculoskeletal health issues. The simulation results showed that limited LL led to increased resultant compressive and shear forces between the lumbar vertebrae, and higher vertebral body compressive loads. In contrast, excessive LL caused an increase in lumbar sagittal shear forces, but a decrease in resultant compression, along with greater facet joint compression and elevated compressive forces at the L5-S1 vertebral body.

4.1. Model evaluation

The linear relationship observed between simulated LL and SS ($SS = 0.59 \times LL + 8.10$, $R^2 = 0.91$) aligned well with prior published work presented in Roussouly et al. (2005), Müller et al. (2021), and

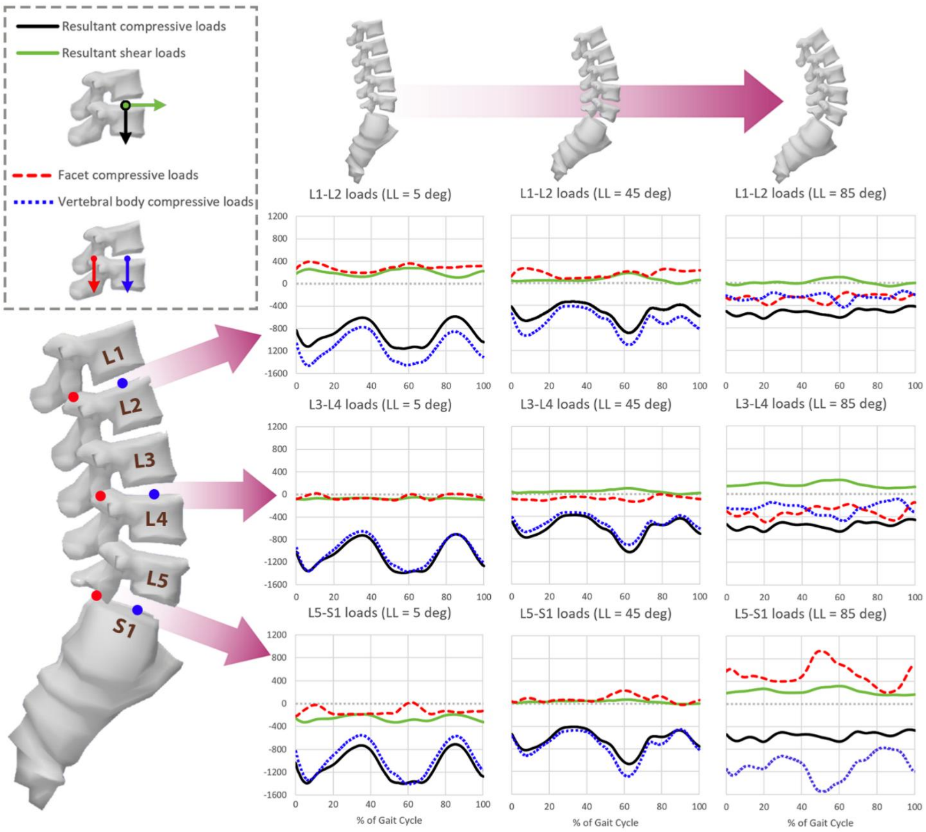

Figure 6. Example data from a participant showing sagittal plane lumbar spinal loads (a. vertebral resultant compressive force (black, solid), b. vertebral resultant sagittal shear force (green, solid), c. vertebral body compressive force (blue, dashed), d. facet compressive force (red, dashed)) at the L1-L2 level, L3-L4 level and L5-S1 level during a single gait cycle with different lumbar lordosis angles (5°, 45° and 85°).

Naserkhaki et al. (2016), which supported the accuracy of the models in representing physiological lumbopelvic alignment (Figure 4).

Previous research concentrated on computing spinal loads in participants with an average LL. To evaluate our model, we concentrated on comparing simulated lumbar loads for the neutral LL of 45° to align with previous studies. The simulated maximum resultant compressive force at L4-L5 during walking, 878 N (1.33 BW) (Table 1), was within the range of a previous inverse-dynamic simulation, which reported 966 N (1.4 BW) (Schäfer et al. 2023). Our OpenSim model computed an average of 856 N (±175) (Table 1), which was equivalent to 1.3 BW at L3-L4. A previous biomechanical trunk model showed a predicted axial load of 1.5BW on the L3-L4 during 1.38 m/s walking (Cappozzo 1984). The differences observed could be attributed to variability in experimental conditions, such as differences in subject characteristics or walking speed.

In terms of load distribution between the lumbar vertebrae, an in-vivo study recorded a peak force of 760 N at the T12-L2 vertebral body during walking at 4 km/h, which aligned with the results we found: 898 N average peak force at the L1-L2 vertebral body at self-selected speed (Rohlmann et al. 2014). Another study looked at the peak axial force on internal fixators of 125 N spanning the L3 at the pedicles during level walking, which was comparable to our results showing an average peak force of 148 N at the L3-L4 facet joint (Rohlmann et al. 1997).

The consistency between our simulated outcomes and previous experimental and modeling studies provided evidence that the OpenSim model in this study can reliably predict lumbar spine loads, even potentially across a range of LL configurations, not just for an LL of 45°.

4.2. Lumbar loads in hypolordosis

Starting from neutral LL (45°), both resultant compressions and sagittal shear forces between lumbar vertebrae increased as the LL decreased from 45 to 5° in the walking simulation (Figure 5A,B). The same trend was found in vertebral body compression (Figure 5C). This variation tendency aligned with forward simulation results in standing posture (Bruno et al. 2017; Müller et al. 2021). With a lower LL, the center of mass was anteriorly displaced, deviating from its ideal position (Whitcome et al. 2007), potentially leading to higher compressive forces at the vertebral body, which subsequently transmitted to the intervertebral disc. The low compressive forces at the facet joints during hypolordosis in the L3/L4 and L5/S1 levels also supported this hypothesis (Figure 5D). Regarding the vertebral resultant sagittal shear forces, prior biomechanical and morphological analyses indicated that the lumbar shear loads increased with decreasing LL (Keller et al. 2005), consistent with the present study findings. Umehara et al. also observed increased shear stresses because of LL reduction using cadaveric lumbar spines tested under axial loading (Umehara et al. 2000).

Interestingly, at the L5/S1 vertebral level, the decrease in LL resulted in a four-times increased rate of sagittal shear force compared to compressive force (x1.5). From a neutral (45°) to a flattened (5°) LL, shear forces increased by 411%, while compressive forces only increased by 32%. This was concerning as shear forces acting on the intervertebral joint may play a more significant role than compressive force in degenerative mechanisms of the intervertebral disc. Some in-vivo studies in rats and rabbits have shown that sustained shear forces induced intervertebral disc degeneration, with shear forces potentially being more harmful than compression loads (Kim et al. 2012; Xia et al. 2015). The shear force required to induce degenerative changes in the intervertebral discs was significantly lower compared to the compression force reported in similar studies, leading to the conclusion that shear force may exert a more harmful effect on intervertebral discs than compression. Participants with reduced LL may experience increased shear loads, triggering the sensitization of nociceptive neurons and eventual disc tissue degradation (Goupille et al. 1998; Weiler et al. 2002). Previous authors have recommended shear force limits of 500–700 N for repetitive actions (100–1000 loads/day) (Gallagher and Marras 2012). The average peak loads at extreme flatten LL values were already close to this limit in the present simulation, and several participants had exceeded this limit. Given this study was simulating low-level activity such as walking, other strenuous movements (e.g. golfing) may further increase the loads and exceed the recommended threshold (Bae et al. 2014), which may cause potential discopathies that could further decrease LL.

In comparison to pain-free participants, patients with LBP, both with and without lower lumbar degenerative discs, have exhibited a statistically significant reduction in LL (Adams et al. 1999; Mirzashahi et al. 2023). LL is known to decrease with age, primarily due to degenerative diseases such as decreased intervertebral disc space height or osteoporosis-related vertebral flattening (Gelb et al. 1995), all of which appear to be associated with LBP.

4.3. Lumbar loads in hyperlordosis

In comparison to neutral LL (45°), excessive LL also leads to a rapid increase in lumbar vertebral sagittal shear force at the L3-L4 and L5-S1 levels during walking, approaching the recommended limit (Gallagher and Marras 2012). However, sagittal shear forces observed at the L1-L2 level and resultant compressive forces of all levels decreased. Some other forward dynamic simulations have similarly observed a decrease in compressive forces with increasing LL during upright standing (Bruno et al. 2017; Müller et al. 2021). This phenomenon may be attributed to the forward tilt of certain vertebrae, particularly those caudal in position, causing a shift of gravity from compressive to shear forces. This gravitational shift potentially explains the decrease in lumbar vertebral compression. As depicted in Figure 5B, the more caudal the level of lumbar vertebrae, the faster the load changed (the greater the absolute value of the slope), indicating a greater forward tilt of the vertebrae and a more significant transfer of gravity from compressive to shear forces.

The decrease of vertebral resultant compressive forces did not imply a reduction in compression at the vertebral body or facet joints. Given the model's simplicity (neglecting passive spinal structures) and the complex nature of intervertebral discs and facet joints, directly exploring the accurate loads between the intervertebral discs or articular processes of each vertebra was not feasible (Meszaros-Beller et al. 2023). Instead, we assessed the loading conditions of these joints by simulating the anterior and posterior vertebral loads. Results showed that compressive load on the vertebral body at the L5-S1 level increased progressively, and facet loads of the more caudal level (L3-L4, L5-S1) increased exponentially with increasing LL during walking, especially after 45° (Figure 5C,D). Other studies also found the same changing pattern during upright standing (Berlemann et al. 1999; Müller et al. 2021). With a greater LL, contrary to hypolordosis, the gravity line shifted posteriorly, concentrating on structures such as the facet joints and spinous processes in the posterior spinal region. This observation may explain why some studies have reported an association between LBP and excessive LL

(Christie et al. 1995; Adams et al. 1999; Norton et al. 2004; Sorensen et al. 2015; Tatsumi et al. 2019), potentially stemming from a different etiology. A prior investigation found a significant correlation between high LL values and the degree of lumbar facet joint degeneration which would most likely be caused by an increase in facet joints loading (Sahin et al. 2015). At the same time, the load on the L5/S1 vertebral bodies also increased with increasing LL, potentially making this functional spinal unit more prone to disorder. The trend observed in L5-S1 vertebral body compressive forces differed with L1-L2 and L3-L4 since the increase in LL inevitably caused the caudal L5-S1 to shift backward, which meant that most of the body weight above it still rested on the vertebral body in front. What we found suggested that future studies can investigate LL in patients with different specific types of LBP (e.g. distinguishing LBP related to disc protrusion and facet joint degeneration) and potentially be able to draw a more definitive association.

This study showed that abnormal LL impacted lumbar loads during walking, with potential variations in other activities. For instance, a similar simulation study has indicated that abnormal LL may result in greater loads during the phase of a relatively high risk of injury in golfing (Bae et al. 2014). Therefore, it can be inferred that in some intense activities, the impact of LL on lumbar loads may be more pronounced, potentially hastening the development of LBP. Future research can explore the effects of different LL values on lumbar loads during various types of activities.

4.4. Limitations

To avoid multifactorial consideration of patient-specific factors (e.g. age, gender, weight, mass distribution, walking speed), gait kinematics of different lumbar alignments were generated by manually modifying the kinematics from a group of healthy individuals in the present study. Hence, the study results do not reflect the lumbar loads of individuals with actual abnormal sagittal spinopelvic alignment.

It is important to note that the vertebral body or facet compressive forces in the study may be higher than true intervertebral disc or facet joint loads. This is because predefined, fixed LL angles were applied throughout the gait simulations, and passive spinal structures, such as the intervertebral discs, were not included in the model. Both LL variations and intervertebral disc function as shock absorbers under vertical loading (Hultman et al. 1992; Castillo and Lieberman 2018). It is also worth noting that the 'neutral' condition of the LL angle in this study was based on the average lumbar curvature measured from standing posture, rather than during walking.

Additionally, as LL increases, thoracic kyphosis also typically increases (Jackson et al. 1998), but the study was limited to spinal alignment in the lumbopelvic region, neglecting the potential effects of thoracic kyphosis or even cervical curvature.

5. Conclusions

The effect of sagittal spinopelvic alignment (primarily lumbar lordosis) on lumbar vertebral joint contact forces and load distribution was investigated using a developed full-body OpenSim model with custom lumbar joints. The model will be available at www.simtk.org/home/facet/. The negative consequences of limited LL during walking included increased resultant compressive and shear forces between lumbar vertebrae. The consequences of excessive lordosis included increased lumbar sagittal shear force, facet joint compression and compression at vertebral bodies in caudal levels (L5-S1), which may progress to different types of LBP. Future studies should investigate LL in patients with different specific types of LBP to draw more conclusive associations between LL and LBP. The current study findings shed light on the intricate relationship between LL and lumbar loads during walking, suggesting potential implications for spinal biomechanics and the development of interventions aimed at optimizing spinopelvic alignment.

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