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Article

Effects of Wheelchair Seat Sagging on Seat Interface Pressure and Shear, and Its Relationship with Changes in Sitting Posture

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Abstract: Objectives: Wheelchair seat sagging is hypothesized to increase pressure and shear forces, potentially leading to pressure injuries. The objective of this study was to assess the impact of correcting wheelchair seat sagging on ischial pressure, shear force, and posture in a population of healthy adults. **Methods**: A total of twenty-two participants who met the study requirements were included in the study. Participants were evaluated under two conditions: with seat base correction (With Correction) and without it (No Correction). Correction was achieved using insert panels. Ischial pressure was measured using a pressure-mapping system (CONFORMat), shear force with a specialized sensor (iShear), and posture with accelerometers (TSND151). The primary analysis compared peak pressure index (PPI), shear force, slide, and postural changes between conditions. The subgroup analysis was conducted as an exploratory approach to assess potential variation among participants with elevated shear forces. **Results**: There was no statistically significant difference in ischial pressure between the No Correction and With Correction conditions ($p = 0.37$). However, shear force and slide were significantly reduced when seat sagging was corrected ($p < 0.05$). Accelerometer data showed no significant difference in postural changes between conditions ($p \geq 0.05$), although the With Correction condition displayed a slight trend toward greater positional variability over time. **Conclusions**: These findings indicate that correcting seat sagging can reduce shear force and slide, potentially lowering the risk of pressure injuries. However, because this study targeted healthy adults, further research involving older or at-risk populations is necessary. Addressing seat sagging could be an important component of comprehensive pressure injury prevention strategies.

Keywords: pressure ulcer; wheelchairs; seating; posture; biomechanics

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1. Introduction

Older adults frequently experience restricted mobility due to the aging process and the progression of medical conditions, leading to prolonged bed rest [1]. Because extended periods out of bed are strongly associated with improved activities of daily living, wheelchairs are frequently used for prolonged daytime seating, thereby helping individuals maintain these beneficial effects [1,2]. In Japan, wheelchairs are provided as facility equipment in hospitals and eldercare institutions and are also accessible at home through long-term care insurance services for assistive devices [3–5].

Nevertheless, although wheelchairs are convenient for living away from bed, prolonged wheelchair use has been associated with musculoskeletal pain and discomfort [6,7]. Additionally, many wheelchair seats and backrests feature sling-type constructions composed of fabric or belts, which often result in seat sagging and compromised seated posture [8,9]. Prolonged seating under these conditions increases the risk of pressure injury development in the seated area. Consequently, guidelines advise using wheelchair cushions with pressure redistribution capabilities to prevent pressure injuries [10]. However, certain types of wheelchair cushions placed on sling seats may have diminished effectiveness due to seat sagging [11]. Pressure injuries are induced not only by vertical pressure but also by friction and shear forces, which can cause deep tissue deformation and ischemia, especially under prolonged loading conditions [12]. Occupying a sagging seat encourages posterior or lateral pelvic tilt, thereby resulting in uneven loading on the ischial tuberosities [9]. As a result, the postural control system of wheelchair users may attempt to minimize instability by tilting the pelvis posteriorly and leaning against the backrest. This compensatory strategy increases loading on the backrest and may lead to increased forward sliding of the buttocks. Such a posture compromises postural stability, alters pressure distribution [13,14], increases anterior shear forces [8], and ultimately elevates the risk of pressure injury development.

Regarding sling seats and their associated sagging, Harms et al. [9] demonstrated that sitting on a sling seat without a wheelchair cushion induces lumbar kyphosis and posterior pelvic tilt, thereby compromising posture. As a countermeasure, Kamegaya et al. [15] placed a wooden insert panel on a sling seat to correct sagging, resulting in marked decreases in pelvic tilt and difficulty with forward reaching; however, peak pressure in the ischial region tended to increase. Conversely, Shin et al. [16] reported reduced pressure when using a urethane pad designed to offload the ischial area, and Yoshikawa et al. [17] observed similar results. Recent research has further highlighted the critical impact of wheelchair seat structure on interface pressure and shear. Shirogane et al. [18] quantitatively measured pressure and shear stress using a flexible sheet-type sensor, highlighting how variations in seat inclination influence shear forces. Barks et al. [19] observed significant associations between wheelchair seat fit and seated posture, focusing on health outcomes such as pain and risk of pressure injuries in older wheelchair users. Additionally, Koda et al. [20] examined the effects of tilt-in-space and reclining angles of wheelchairs on normal and shear forces in the gluteal region, confirming substantial influences of seat adjustments on pressure and shear distributions.

While a few studies have investigated how sling seats affect posture and seating pressure, few have thoroughly explored the specific impact of seat sagging on shear forces. Therefore, the aim of this study was to investigate the effects of wheelchair seat sagging on pressure, shear force (specifically, anterior–posterior shear force, which contributes to posterior pelvic tilt), and posture in order to inform the development of effective strategies for the prevention of pressure injuries. Our overarching objective is to support pressure injury prevention for adults dependent on wheelchairs. However, due to difficulties controlling variables such as body size, posture, and physical conditions (e.g., contractures or deformities) in older populations, healthy adults were recruited for this study. This approach allows us to clearly ascertain how correcting seat sagging influences pressure, shear, and posture and elucidate underlying biomechanical mechanisms.

2. Methods

2.1. Research Design

The research design was a crossover comparison study of healthy subjects.

2.2. Ethical Considerations

This study received approval from the Ethics Committee of the Graduate School of Nara Gakuen University (approval number: 5-002). Written informed consent was obtained from all participants. All data were managed to preserve anonymity, and participant privacy was rigorously protected.

2.3. Participants

The experimental procedures of this study are presented in Figure 1. This study included 24 healthy adults aged 18 years or older. After measuring each participant's height and weight, the following body measurements (with the corresponding wheelchair dimensions in parentheses) were taken to adjust the wheelchair to the user's body: sitting hip width (seat width), sitting buttock–popliteal length (seat depth), sitting seat to elbow height (arm support height), sitting seat to axillary height (back support height), and lower leg length (footrest height). To minimize discrepancies related to wheelchair fit, only individuals whose sitting hip width ranged from 34 to 40 cm and whose buttock–popliteal length ranged from 41 to 49 cm were included. Exclusion criteria encompassed a sitting hip width below 33 cm or above 41 cm, a buttock–popliteal length below 40 cm or above 50 cm, or any history of orthopedic disorders affecting the lumbar region or lower limbs. The sample size was determined based on a previous study [21], which reported mean and standard deviation values for seat interface pressure (experimental group: 25.9 $\pm$ 4.7 mmHg, control group: 23.4 $\pm$ 5.4 mmHg), yielding an effect size of roughly 0.5. The effect size of 0.5 was calculated using Cohen's d formula, where the mean difference between groups ($25.9 - 23.4 = 2.5$ mmHg) was divided by the pooled standard deviation [$\sqrt{((4.7^2 + 5.4^2)/2)}$, which is approximately 5.07], resulting in d of approximately 0.49, which was rounded to 0.5 for power analysis. Using G*Power (version 3.1.9.7) for a *t*-test with an effect size of 0.5, an alpha of 0.05, and a power of 0.8, we derived a sample size of 21. To accommodate potential attrition, 24 participants were ultimately recruited.

2.4. Equipment

An adjustable "Revo" wheelchair (Rack Healthcare, Osaka, Japan) was used for data collection. The backrest height was 40 cm, the seat depth was 40 cm, the seat width was 40 cm, and the seat angle was 3.85°. Seat sagging was defined by stretching a straight reference line between the two seat supports, then measuring the vertical displacement (sag) occurring when the central areas of the sling seat at both the front and rear were pressed downward. In the wheelchair employed, a deflection of 5 cm was noted at both the front and rear (Figure 2). A flat urethane Moderato cushion (Rack Healthcare, Osaka, Japan), measuring 40 $\times$ 40 $\times$ 6 cm, served as the wheelchair cushion. A commercially available Kiso seat base (Tatsuno Cork Industry, Tatsuno, Japan), measuring 40 $\times$ 40 $\times$ 3.5 cm, was used to correct seat deflection (Figure 3b).

Ischial pressure was measured using CONFORMat (Tekscan Inc., Boston, MA, USA). The sensor sheet's specifications included dimensions of 471 mm (length) $\times$ 471 mm (width), 1024 sensors (32 rows $\times$ 32 columns), a thickness of 1.8 mm, and a resolution of 14.7 mm.

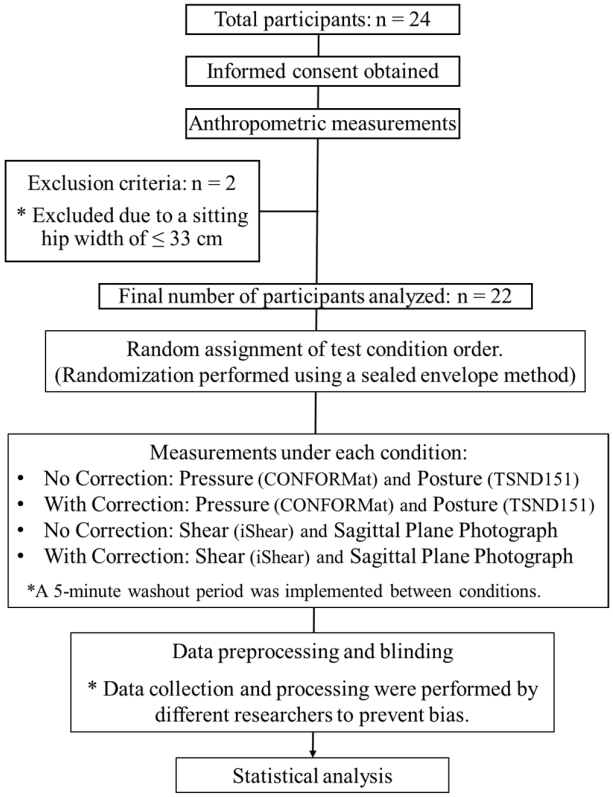


Figure 1. Study procedures.



(**a**) Seat sagging (Front) (**b**) Seat sagging (Rear)

Figure 2. Measurement of wheelchair seat sagging. Panel (**a**) measures the seat sagging at the front of the seat, and panel (**b**) measures the seat sagging at the rear of the seat. The wheelchair seat exhibited 5 cm of seat sagging at both the front and the rear.



(**a**) No Correction (**b**) Seat base (**c**) With Correction

Figure 3. Correcting sling seat sag using the seat base and not using it. Panel (**a**) is the state where the seat sagging of the seat is not corrected (No Correction), panel (**b**) is the seat base used to correct the seat sagging, and panel (**c**) is the state where the seat sagging is corrected using the seat base (With Correction).

Shear force was recorded using the iShear system (Invacare Inc., Gland, Switzerland). The sensor measures 27 × 690 × 615 mm, weighs 1.7 kg, and accommodates users weighing 45–120 kg. It consists of a lightweight aluminum sensor bar attached to a PU-coated tricot/nylon fabric mat, a construction that minimizes interference with seat immersion and posture (Supplementary Figure S1, Supplementary Table S1).

Wheelchair-seated posture was evaluated using TSND151 accelerometers (ATR-Promotions, Kyoto, Japan) sampled at 5 Hz.

Furthermore, the three measurement devices used in this study (CONFORMat, iShear, and TSND151) have been utilized in prior studies, and their measurement methodologies are well-established [22–25].

2.5. Measurement Procedure and Analysis

In this study, the measurers and evaluators were different individuals. Measurements were conducted by an occupational therapist with 30 years of experience, a physical therapist with 18 years of experience (who holds a master's degree), and a physical therapist with 13 years of experience. Under each measurement condition, the height of the arm support was adjusted to match each subject's sitting axillary height. This positioning ensured that the shoulders remained relaxed, allowing the forearms to comfortably support the weight of the upper body. Additionally, the height of the foot support was adjusted so that the thighs were horizontal relative to each subject's sitting lower leg length, thereby supporting both the buttocks and thighs. Because the required support heights differed depending on whether correction was applied, they were adjusted each time. During measurements, a plain urethane wheelchair cushion (without contour) was placed on the seat. When sling seat sag was corrected using the seat base, the condition was labeled "With Correction," and when no seat base was used, it was labeled "No Correction." Measurements of ischial pressure using CONFORMat and shear force using iShear were each conducted separately for 10 min. This approach was adopted because iShear is made of a hard material, which could potentially affect the pressure measurements and deflection corrections.

CONFORMat was placed on the wheelchair cushion to measure ischial pressure. Calibration was performed before measurement, and the sampling frequency was 20 Hz. When deflection correction was performed, the wheelchair cushion was first placed on the insert panel, and then CONFORMat was placed on top of it. After participants were seated in the wheelchair and their posture stabilized (approximately 1 min), pressure recording began. Pressure was continuously monitored for 10 min. Before the measurements began, both ischial areas to be analyzed were identified. This was performed by locating the position of the ischial bones through palpation. The values recorded after 10 min were then used for analysis, specifically focusing on the peak pressure within the identified area and the total of the three adjacent cells (the four cells with the highest average value) [26,27]. This was referred to as the peak pressure index (PPI) [27].

iShear was placed on the wheelchair cushion in the same orientation used for CONFORMat measurements, with its rear edge extending slightly beyond the bottom of the backrest to prevent contact with the back support and to maintain the same posture as during the ischial-pressure test. For the deflection-corrected condition, the cushion was positioned on the insert panel with iShear on top. To establish a baseline, each participant first sat on the seat plate without the cushion, and shear force was sampled at 4 Hz. The cushion (and insert panel, when applicable) was then installed, the participant reseated, and after a 1 min stabilization period, shear force was continuously recorded for 10 min. The highest shear-force value (N) observed during this interval was normalized to the baseline (N/N). Although pressure and shear forces were sampled at different frequencies

(20 Hz vs. 4 Hz), each outcome was expressed as the maximum value observed within the 10 min window; therefore, differences in sampling frequency do not compromise the validity of the analysis.

Additionally, to evaluate body displacement, photographs were taken from the sagittal plane at the start and end of each measurement to assess how much the body slid along this plane (Figure 4). These photographs were always captured from the same position, with a camera mounted on a tripod located 5 m from the wheelchair and at a height of 1.5 m. Each image included markers placed on the wheelchair axle and on the participant's patella. The photographs were then transferred to ImageJ version 1.54p, where distances were scaled according to the known diameter of the wheelchair wheel. In ImageJ, the distance between the wheelchair axle and the participant's patella was measured (Figure 4). The amount of slip was calculated by subtracting the distance measured at the start of the session from the distance measured at the end. To minimize observer bias, data collection and image analysis were carried out by different individuals. All photographs were captured by the measurer, whereas slide distance was quantified in ImageJ by a separate evaluator. The evaluator received only de-identified image files renamed with anonymized codes and was therefore blinded to whether each image corresponded to the With Correction or No Correction condition; the code key was opened only after all measurements had been completed.



Figure 4. Method for measuring slide. All photographs included markers at the wheelchair axle and the patella. The yellow line represents the straight-line distance between the wheelchair axle and the patella. Images were imported into ImageJ, and distances were adjusted according to the wheel's inch size. For the analysis, the distance between the wheelchair axle and the patella was measured using ImageJ.

Moreover, wheelchair-seated posture was quantified using TSND151 accelerometers. A single examiner attached accelerometers to each participant's forehead, sternum, and both iliac crests using hook-and-loop straps. We selected the manubrium and iliac crests as relatively stable bony landmarks with minimal soft tissue movement, thereby helping to maintain accurate sensor alignment. To further ensure reliable detection of postural movement, we adjusted and confirmed sensor placement before and after each measurement session. Tilt angles were defined relative to the participant's initial posture, with forward tilt designated as positive and backward tilt as negative. The angle data used for analysis were the 1 s averages at 1 min (start of posture measurement) and 10 min (end of posture measurement). Postural change was determined by subtracting the initial angle from the final angle.

The measurement order was randomized by an envelope method. Participants were instructed to relax as much as possible and to avoid intentional movement or repositioning during each 10 min measurement session.

2.6. Statistical Analysis

All statistical analyses were conducted using EZR (Easy R; version 1.53), which is a graphical user interface for R designed to add statistical functions frequently used in biostatistics [28]. The Shapiro–Wilk test was employed to assess the normality of each variable (PPI, shear force, slide, and angle data) prior to selecting the appropriate statistical test for comparing the With Correction and No Correction conditions. To compare differences in shear force, slide, and angle values between the With Correction and No Correction conditions, the paired *t*-test was applied to data that followed a normal distribution. For data that did not meet normality, the Wilcoxon signed-rank test was used instead. Subgroup analysis was performed on participants who exhibited a notably elevated shear force (5 N/N). The 5 N/N threshold was derived from the median measured value with deflection correction. For any variable demonstrating statistical significance, correlations with height, weight, and BMI were explored. The significance level was set at 5%.

3. Results

Of the 24 people who applied, 2 were excluded because their sitting hip width was 33 cm or less, leaving a final total of 22 participants (11 men and 11 women). Their mean height was 162.3 ± 6.6 cm, mean weight was 59.3 ± 13 kg, and mean BMI was 22.3 ± 3.8 (Table 1). With respect to body dimensions, the average sitting hip width was 35.7 ± 1.8 cm, buttock–popliteal length was 44.0 ± 2.8 cm, sitting lower leg length was 41.5 ± 2.3 cm, sitting elbow height was 22.9 ± 2.8 cm, and sitting axillary height was 41.7 ± 6.7 cm (Table 1). The Shapiro–Wilk test revealed that the PPI, shear force, and slide data did not significantly deviate from normality ($p \geqq 0.05$), whereas tilt angle change data did ($p < 0.05$).

Table 1. Basic participant characteristics.

	Mean ± SD [2]
Sex (male/female)	11/11
Age (years)	37.5 ± 11.3
Height (cm)	162.3 ± 6.6
Weight (kg)	59.3 ± 13.0
BMI [1]	22.3 ± 3.8
Sitting Hip Width (cm)	35.7 ± 1.8
Buttock–Popliteal Length (cm)	44.0 ± 2.8
Sitting Lower Leg Length (cm)	41.5 ± 2.3
Sitting Elbow Height (cm)	22.9 ± 2.8
Sitting Axillary Height (cm)	41.7 ± 6.7

[1] BMI: body mass index, [2] SD: standard deviation.

3.1. Ischial Pressure

The PPI over the ischial tuberosity was 65.4 ± 39.9 mmHg in the No Correction condition and 68.1 ± 38.4 mmHg in the With Correction condition, with no statistically significant difference between them (Table 2, Figure 5).

Table 2. Ischial pressure, shear force/slide, and changes in posture.

	No Correction (Mean ± SD) or (Median [IQR])	With Correction (Mean ± SD) or (Median [IQR])	*p*-Value
Ischial pressure (mmHg) ‡	65.4 ± 39.9	68.1 ± 38.4	0.37
Shear force (N/N) ‡	22.6 ± 28.3	13.0 ± 16.7	<0.05
Slide (cm) ‡	0.6 ± 0.3	0.3 ± 0.2	<0.05
Tilt angle (Head; °) †	−0.15 [−3.82–9.61]	−1.08 [−5.37–7.96]	0.23
Tilt angle (Chest; °) †	0.19 [−4.25–8.13]	−0.08 [−5.93–21.26]	0.92
Tilt angle (Left Iliac Crest; °) †	−0.16 [−1.37–1.18]	−0.74 [−5.52–5.52]	0.20
Tilt angle (Right Iliac Crest; °) †	−0.24 [−1.36–0.81]	−1.27 [−8.62–2.58]	0.08

‡ Paired *t*-test (Mean ± SD), † Wilcoxon signed-rank test (median [IQR]).

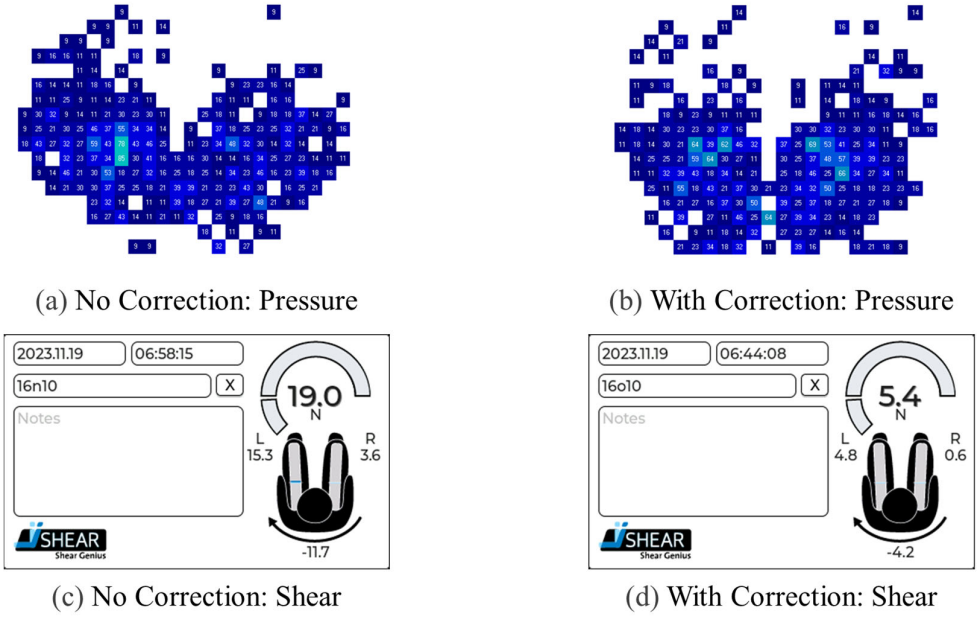


Figure 5. Representative distribution maps of pressure and shear force: (**a**) pressure distribution without correction, (**b**) pressure distribution With Correction, (**c**) shear force without correction, and (**d**) shear force With Correction. While no notable differences were observed in pressure distribution between conditions, shear force was reduced when seat sagging was corrected. The numbers shown in the mapping in (**a**,**b**) indicate pressure values in mmHg.

3.2. Shear Force and Slide

In the No Correction condition, shear force was 22.6 ± 28.3 N/N, whereas it was 13.0 ± 16.7 N/N in the With Correction condition, indicating a significant reduction in shear force with seat base correction ($p < 0.05$, Table 2 and Figure 5). Slide in the No Correction condition (0.6 ± 0.3 cm) was also significantly greater than in the With Correction condition (0.3 ± 0.2 cm) ($p < 0.05$, Table 2). An examination of shear force correlations with height, weight, and BMI showed no significant relationships in either condition (No Correction: r = 0.18, 0.06, 0.17; With Correction: r = 0.17, 0.02, 0.15) (Table 3).

Among the 22 participants, 12 exhibited a maximum shear force ≥ 5 N/N. In this subgroup, shear force in the No Correction condition was 39.6 ± 29.0 (range: 6.5–93.0) N/N, whereas it was 21.9 ± 17.7 (range: 6.0–54.0) N/N With Correction, showing a significant decrease ($p < 0.05$, Table 4). Correlations between shear force and height, weight, or BMI in this subgroup were also nonsignificant (No Correction: r = 0.25, 0.11, 0.06; With Correction: r = 0.18, 0.08, 0.05).

Table 3. Correlations with shear force (Pearson's correlation: r).

	No Correction	With Correction
Height	0.18	0.17
Weight	0.06	0.02
BMI [1]	0.17	0.15

[1] BMI: body mass index.

Table 4. Shear force among participants with elevated shear force.

	No Correction [95%CI]	With Correction [95%CI]	*p*-Value
Shear sorce (N/N) ‡	39.6 ± 29.0 [21.2–58.0]	21.9 ± 17.7 [10.7–33.1]	<0.05

‡ Paired *t*-test (Mean ± SD).

3.3. Postural Changes

Changes in the angles of the head, chest, left iliac crest, and right iliac crest (expressed as median [interquartile range]) were assessed. The head angle changes were −0.15 [−3.82–9.61]° in the No Correction condition and −1.08 [−5.37–7.96]° in the With Correction condition. The chest angle changes were 0.19 [−4.25–8.13]° under No Correction and −0.08 [−5.93–21.26]° under With Correction. For the left iliac crest, the change was −0.16 [−1.37–1.18]° in the No Correction condition and −0.74 [−5.52–5.52]° in the With Correction condition. For the right iliac crest, the change was −0.24 [−1.36–0.81]° under No Correction and −1.27 [−8.62–2.58]° under With Correction. Although there were no significant differences between the two conditions, there was a tendency toward greater postural variation under With Correction (Table 2).

4. Discussion

This study examined how correcting or not correcting wheelchair seat sagging influences ischial pressure, shear force/slide, and postural adjustments. The results revealed no significant differences in ischial pressure between the No Correction and With Correction conditions; however, shear force and slide were somewhat reduced under the With Correction condition. Postural changes in head, thoracic, or pelvic angles did not differ significantly between the two conditions, although there was a tendency for smaller postural fluctuations under No Correction.

In this study, we used a standard wheelchair with a seat width of 40 cm and a seat depth of 40 cm. However, this depth was shallower than the participants' average seated Buttock–Popliteal Length (44 cm), and the width was a few centimeters wider than their average seated Sitting Hip Width (35.7 cm). Such size mismatches may influence how the pelvis shifts anteriorly/posteriorly and laterally, as well as how the lower limbs are supported, potentially altering the distribution of pressure and shear forces. Nonetheless, in many clinical settings, standardized wheelchairs are often used as facility equipment, and they are not always optimally adjusted for each individual user. We believe our results, to some extent, reflect these real-world conditions where "standard-sized" wheelchairs must be used. Conversely, utilizing seat dimensions and cushions tailored more closely to individual needs could improve seated postural stability, potentially leading to better pressure redistribution and reduced shear forces. Going forward, it will be necessary to examine these aspects in greater detail by investigating how fine-tuned seat sizing and cushion shape adjustments affect wheelchair use in actual clinical environments.

In wheelchair seating, tissue breakdown is generally believed to occur when high pressures are applied to the seat surface [29], although no definitive cutoff value has been established [11]. The Braden Scale is a validated assessment tool widely used to identify individuals at risk of pressure injury, with lower scores indicating higher risk [30,31].

A Braden score of 18 is commonly considered a threshold for moderate risk [30]. Brienza et al. [29] examined seat interface pressure in individuals aged 65 and older who use wheelchairs in nursing care settings and have Braden scores of 18 or lower; their findings demonstrated that an ischial PPI of 70 ± 16 mmHg did not lead to pressure injury formation. In the present study, mean ischial pressure ranged from approximately 65 to 70 mmHg under both conditions, suggesting a typical value for the ischial region. Linder-Ganz et al. [32] noted that skeletal muscle cell death ensues after at least two hours of exposure to pressures exceeding 68 mmHg, and clinical guidelines [10] recommend performing weight shifts every 15 min to prevent pressure injuries in wheelchair users. Because the measurement period in this study lasted only 10 min, no weight shifting was required. Nevertheless, since pressures within this range can pose a risk over extended periods, regular weight shifts remain advisable, irrespective of the absolute interface pressure level.

In this study, seat sagging was corrected by employing insert panels composed of polyethylene and polystyrene, which did not substantially alter ischial pressures relative to the No Correction condition. Yoshikawa et al. [17] demonstrated that soft, contoured urethane inserts lowered ischial pressure when combined with an air cushion, also showing a tendency for reduced pressure with urethane or gel materials. Likewise, Shin et al. [16] reported that a urethane pad specifically designed to offload the ischial region effectively reduced pressure. In contrast, Kamegaya et al. [15] noted that a wooden insert panel heightened peak pressure. Collectively, these findings suggest that the impact of seat sagging correction on pressure may vary according to the material and contour of the insert panel, thereby warranting further investigation.

Additionally, it is acknowledged that the formation of pressure injuries is influenced not only by vertical pressure but also by friction and shear forces in the horizontal plane [10]. When an individual sits on a sling seat, lumbar kyphosis and posterior pelvic tilt may arise, thereby augmenting the load on the backrest and amplifying the forward-sliding force on the buttocks [8,9,33]. In this study, we assessed how shear force and slide vary based on whether frontal plane sag in the sling seat is corrected. Our findings revealed that both shear force and slide were substantially higher when the seat was not corrected. In a wheelchair-seated posture, the seat surface and backrest collectively support the user's body weight; however, the buttocks bear the brunt of the load, resulting in concentrated pressure and stress within this soft tissue region [34]. When frontal plane sagging of the sling seat causes the pelvis to tilt to one side, the depressed portion of the sling also shifts in that direction, thereby increasing pelvic sway in the frontal plane [8]. Although this lateral (frontal plane) tilt may introduce or exacerbate shear forces in the medial–lateral direction, in the present study we primarily measured shear forces in the anterior–posterior axis (i.e., the direction of forward sliding). The instability arising from a lack of correction presumably promotes shear; in an effort to mitigate this, the user's postural control system attempts to minimize sway by shifting the pelvis posteriorly. Consequently, the load on the backrest escalates, encouraging anterior sliding of the buttocks [9,33]. An increase in shear not only elevates the risk of deep tissue injury [35], but the coexistence of pressure and shear has also been shown to constrict blood vessels, triggering ischemic conditions [36,37]. Although no significant difference in ischial pressure was identified in this study, the heightened shear force and slide under the No Correction condition suggest that ischemic states in soft tissues could be exacerbated, thereby increasing the likelihood of pressure injuries.

Nonetheless, despite the substantial difference in shear force and slide, no discernible differences were observed in the measured postural changes, irrespective of seat correction. Earlier studies have indicated that posterior pelvic tilt (in the sagittal plane) generates shear force primarily in the anterior–posterior direction [8,33] and that seat correction can influence lumbar alignment [9]. However, seat correction addressing frontal plane sag may

also help mitigate any medial–lateral shear components. Hence, we hypothesized that seat correction would lessen postural changes and diminish shear force production. In practice, although seat correction considerably lowered shear force, it did not significantly influence postural changes. Future studies should consider assessing shear forces in both anterior–posterior and medial–lateral directions to clarify the full impact of seat correction. Although we measured tilt angles of the iliac crests, we did not directly assess vertical displacement or asymmetry between the left and right sides. Future studies should consider incorporating pelvic height measurements to better understand lateral instability and its contribution to shear forces. Without correction, from the moment they sat on the sling seat, they leaned back against the back support as previously mentioned, and their posture transitioned to one characterized by lumbar lordosis and pelvic tilt [9], and because it was fixed, it is believed that minimal postural change occurred. Although previous research has established that posterior pelvic tilt can engender shear force [8,33], our results indicate that elevated shear force at the seat does not necessarily present as a visible postural change. Moreover, in this study, even though the head and trunk posture were not considerably distorted, shear force and slide increased, suggesting that in cases where the wheelchair seat deflection is not corrected, shear stress is induced in the soft tissue surrounding the ischium, even without overt postural distortion. This observation is particularly alarming because it implies a heightened risk of developing deep tissue injury [38]. No such injuries occurred in this study—likely because the participants were healthy adults tested for only 10 min. Guidelines [10] recommend performing push-ups or pressure redistribution every 15 min. However, this is often challenging for older adults in care facilities, particularly those with dementia. Because our measurements were limited to a 10 min period, we stress the need for caution when working with older individuals who remain seated in a wheelchair for extended periods and cannot change their position, as well as those with reduced sensation in the seated area. Furthermore, this study targeted healthy participants and instructed them to maintain a comfortable sitting posture. Many older adults who use wheelchairs, however, exhibit kyphosis, scoliosis, or posterior pelvic tilt [39,40]. Such conditions can alter load distribution, as backward tilting of the pelvis may exacerbate pressure [13,14], shear force, and sliding [14], as well as increase postural changes. Consequently, the outcomes in this study may not fully capture the responses of older adults with these conditions.

An analysis of the entire cohort examining the relationships between shear force and height, weight, and BMI revealed no significant correlations in any condition, indicating that these anthropometric characteristics are not the principal determinants of shear force. Instead, a seat sagging in sling chairs likely fosters posterior pelvic tilt and exacerbates shear force.

One limitation of this study is that all participants were healthy adults, whereas older individuals and those who use wheelchairs daily often exhibit different load-bearing patterns [38], as well as differences in muscle mass and skin integrity. Older adults, whose physical functions have declined, not only experience greater instability in seated balance but are also more prone to muscle fatigue and posterior pelvic tilt during prolonged wheelchair sitting. As a result, the manner in which pressure and shear forces arise may differ substantially from that of healthy adults. Future research should therefore focus on older individuals who rely on wheelchairs for daily mobility to corroborate these findings. Additionally, the standardized methodology and posture measurements used to measure shear force accurately were not based on previously reported methods but were independently developed. It is essential to develop standardized methodologies and posture measurements to ensure accurate shear force assessment. Further investigations are also needed to identify the most suitable materials for insert panels used in correcting sling seat sagging.

5. Conclusions

Pressure injury prevention requires controlling both vertical pressure and shear force. The results of this study suggest that correcting seat sagging in a wheelchair can reduce shear force and slide at the seat interface. Consequently, in addition to using cushions that redistribute pressure, addressing seat sagging is vital for mitigating shear force during wheelchair use. Further research should be conducted with older individuals or patients with mobility impairments to better understand how these findings translate to real-world clinical use.

Supplementary Materials: The following supporting information can be downloaded at: https://www.mdpi.com/article/10.3390/biomechanics5020041/s1, Figure S1: Seat sagging under conditions with iShear, cushion, and seat base correction; Table S1: Seat sagging under conditions with iShear, cushion, and seat base correction.

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