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ORIGINAL ARTICLE

Regarding loads after spinal fusion, every level should be seen separately: a musculoskeletal analysis

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Abstract

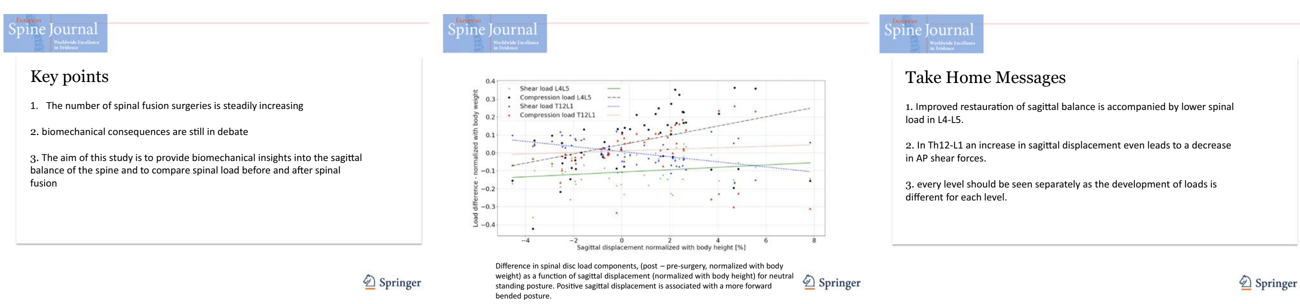
Introduction The number of spinal fusion surgeries is steadily increasing and biomechanical consequences are still in debate. The aim of this study is to provide biomechanical insights into the sagittal balance of the spine and to compare spinal load before and after spinal fusion.

Method The joint reaction forces of 52 patients were analyzed in proximo-distal and antero-posterior direction from the levels T12–L1 to L5–S1 using musculoskeletal simulations.

Results In 104 simulations, pre-surgical forces were equal to post-surgical. The levels L4–L5 and T12–L1, however, showed increased spinal forces compression forces with higher sagittal displacement. Improved restauration of sagittal balance was accompanied by lower spinal load. AP shear stress, interestingly decreased with sagittal imbalance.

Conclusion Imbalanced spines have a risk of increased compression forces at Th12–L1. L4–L5 always has increased spinal loads.

Graphical abstract These slides can be retrieved under Electronic Supplementary Material.



Keywords Sagittal balance · Spinal fusion · AnyBody Modeling System · Spine biomechanics · Musculoskeletal analysis

Introduction

One of the most important functions of the spine is to keep the balance in any body position. Achieving this balance as efficiently as possible not only requires good interaction between muscles, ligaments, and the spine but also the correct position of the pelvis and the legs [1].

From a sagittal point of view, balance is generally defined by a perpendicular line beginning at the center of the seventh cervical vertebra that intersects the posterior edge of the S1 end plate located behind the hip joint axis. To maintain

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this position with as little force as possible, the position of the pelvis is important because of its alignment with the spinal column. One of the basis of spinal balance is the cone of economy. In this theory, "balance is defined as the ability of the human body to maintain its center of mass within the base of support with minimal postural sway" [2]. Recently, Haddas et al. could quantify the cone of economy, which "will enable spine care practitioners to objectively evaluate their patients in an effort to determine the most appropriate treatment options, and in objectively documenting the effectiveness of their intervention" [2]. Although spinopelvic balance has initially little clinical significance [3, 4], it is very important in the surgical treatment of various vertebral deformities, particularly in lumbar fusion and in managing vertebral fractures [5, 6]. Sacral slope (SS) refers to the angle between the sacral end plate and a horizontal reference line and pelvic tilt (PT) to the angle between the line connecting the midpoint sacral end plate with the center of the hip joint and a vertical reference line. PI is the sum of SS and PT.

After lumbar fusion, loss of lumbar lordosis (LL) with compensatory mechanisms may occur. SS and thoracic kyphosis (TK) may be decreased, and PT may be increased, which is correlated with postoperative back pain after lumbar fusion. A decrease in SS and/or an abnormal sagittal vertical axis (SVA) are often associated with adjacent segment degeneration. The risk of sagittal imbalance after spine fusion is increased by high pelvic incidence (PI). Good clinical outcome is often associated with the restoration of normal PT [7, 8].

Therefore, knowledge about the biomechanics of the spine including sagittal balance and associated forces is very important. When investigating the effect of lumbar vertebral geometry, Putzer et al. found that vertebral body height has the largest influence on lumbar load [9, 10]. The comparison of the influence of different lumbar spinal rhythms on spinal load showed that spinal rhythm mostly influences intervertebral shear forces [11–13]. Although many influences on spinal loading have been investigated, the biomechanical effects of sagittal balance and other spino-pelvic parameters have not yet been examined, particularly in combination with regard to the effects of spinal fusion.

Aim of the study

The aim of this study was to get biomechanical insights into the sagittal balance of the spine and to compare spinal load before and after spinal fusion.

Table 1 Number of fusion levels

Fusion level	Number (*n*)
L3–S1	18
L4–L5	11
L4–S1	28
L5–S1	13

For fusions from L3–S1, L4–L5, L4–S1, and L5–S1

Table 2 Mean and range of pre- and postoperative lordosis angles and pelvic parameters

Parameter	Pre-surgical		Post-surgical	
	mean (°) ± SD	Range (°)	Mean (°) ± SD	Range (°)
L1–S1	52 ± 16	14 to 86	51 ± 14.5	20 to 90
L2–S1	46 ± 25.5	11 to 81	43 ± 12.5	20 to 61
L3–S1	36 ± 13.1	10 to 49	39 ± 10.5	19 to 64
L4–S1	29.8 ± 13.4	− 4 to 65	29 ± 10.8	10 to 50
L5–S1	11.6 ± 14.2	− 22 to 22	18 ± 3.4	11 to 24
Pelvic incidence	59.2 ± 14.6	32 to 98	59.5 ± 14.8	30 to 94
Sacral slope	37.9 ± 12.2	9 to 70	37.8 ± 11.3	15 to 66
Pelvic tilt	21.4 ± 9.1	3.5 to 44.5	21.7 ± 9.1	− 1 to 51

Methods

Test subjects and data

The spino-pelvic parameters of 51 patients (29 female, 2 male) with different spinal fusion treatments were available. The mean (± SD) patient age was 61 (± 13) years, ranging from 17 to 81 years. The average height was 1.68 ± 0.09 m, ranging from 1.52 to 1.89 m. The average weight was 83.2 ± 19.3 kg, ranging from 50 to 140 kg. All fusions were in dorsal TLIF technique in at least the segment L4–L5 or L5–S1. Table 1 displays the incidence of different fusion levels. Data were obtained from X-ray images taken before (pre-surgery) and shortly after surgery (post-surgery). The analyzed parameters were: vertebral body height, lumbar lordosis, sacral slope and pelvic tilt.

All X-rays were taken in standing position, and measurements were independently done twice by two of the co-authors. The mean value of the four measurements was used for simulation.

Pelvic parameters can be seen in Table 2.

The musculoskeletal model

Patient specific musculoskeletal models have been created utilizing the standing human body model of the AnyBody Modeling System (AMS, Version 6.0.5.4379). The AnyBody Managed Model Repository (AMMR Version 1.4.1) provided different basic body parts, such as the skull, arms, spine, pelvis, and legs. The cervical and thoracic spine as well as the ribcage were one lumped part. Intervertebral joints in the lumbar spine were spherical joints with three degrees of freedom (DOF). The lumbar spine model was described in detail by de Zee et al. [14] and has been extended and validated by Putzer et al. [9]. The lumbar spine anatomy was defined for each patient with the accordingly measured data on lumbar lordosis (LL) from L1 end plate to S1 end plate, sacral slope, and pelvic tilt. Furthermore, body height and body mass were used to scale the overall human body model. To solve the inverse dynamics, motion has to be predefined. For this, integral spinal motion was distributed on the lumbar discs according to the spinal rhythm described by Wong et al. [11] and the spino-pelvic ratio described by Lee et al. [15]. To maintain balance, the human body model compensated the forward movement of the trunk with counter wise motion in the ankles. Spinal fusion has been simulated by changing the spinal rhythm components. Values of the spinal rhythm matrix, which distribute the integral thorax motion onto the individual disc level have been set to zero for the fused levels and increased for the adjacent-to-fusion levels to spread the “lost” motion of the fused segment equally to upper and lower discs. In cases of no lower or upper disc, L5–S1 and T12–L1 level, all “lost” motion has been assigned to the corresponding adjacent levels L4–L5 and L1–L2, respectively. Therefore, motion was prevented in the fused motion segments and increased in the adjacent levels. Full force transfer between the fused vertebral bodies has been activated which simulates a rigid connection between the vertebral bodies. Thus, all force and moment components could be transferred through the fusion (Fig. 1).

Fig. 1 Musculoskeletal full body model

Outcome variables

The initial sagittal balance of the musculoskeletal model because of measured components was computed. Sagittal displacement was determined by the distance from the posterior superior corner of the sacrum to a vertical plumb line from C7 center. Joint reaction forces, compression and anterior–posterior shear (AP shear) were analyzed for intervertebral joints L4–L5 and T12–L1 for static, neutral standing posture (according to X-ray images) and dynamic forward flexion from neutral to 50°. The forces were analyzed for pre- and post-surgery data. Load differences between pre- and post-surgery have been determined for neutral standing. Linear regression analysis was used to determine the slope of spinal disc forces as a function of forward bending. Statistics have been computed using SciPy.

Results

104 patient specific models were analysed. Neutral standing posture changed in most patients due to spinal fusion surgery. Figure 2 shows spinal disc forces changes in relation to changes in sagittal displacement due to spinal fusion for neutral standing. Forces were normalized by body mass and sagittal displacement by body height to increase comparability of the data sets.

Relationships between differences in load components for L4–L5 and T12–L1 for shear and compression force components have been investigated.

Weak linear correlations are found for T12–L1 shear and L4–L5 compression forces. Slope for significant components were significantly different from zero for both, T12–L1 ($p < 0.002$) and L4–L5 ($p < 0.024$). No correlations could be observed for the other components. L4–L5 compression forces increase with increased sagittal displacement whereas T12–L1 shear showed an inverse

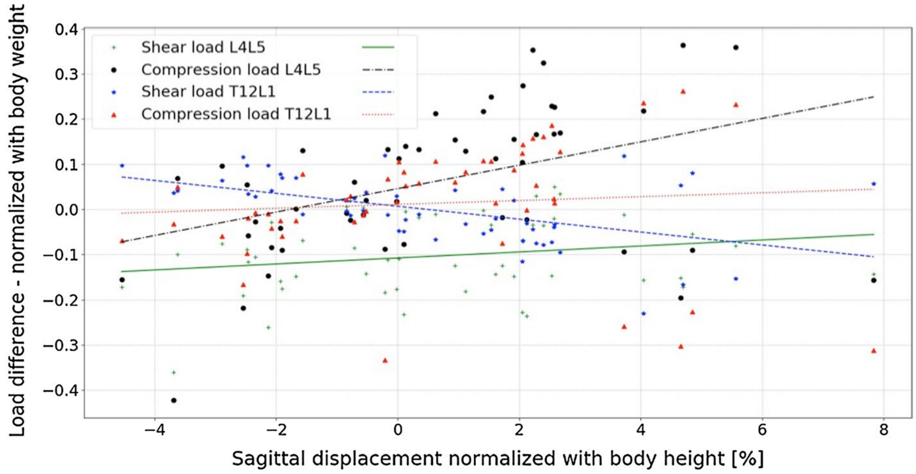

Fig. 2 Difference in spinal disc load components, (post–pre-surgery, normalized with body weight) as a function of sagittal displacement (normalized with body height) for neutral standing posture. Positive sagittal displacement is associated with a more forward bended posture

Table 3 Mean force values and standard deviation for different spinal force components, pre- and post- surgery

	L4–L5		T12–L1	
	Compression	AP shear	Compression	AP shear
Pre surgery	545 N	52 N	501 N	178 N
Post surgery	606 N	31 N	521 N	185 N

trend. Maximal force increase in compression is almost 0.4 BW (body weight) (Table 3).

The influence of sagittal displacement on dynamic forces has been investigated. Figure 3 displays results of slope analysis of dynamic musculoskeletal simulations. There was no significant relationship between slope of dynamic forces and sagittal displacement.

Discussion

The aim of this study was to get biomechanical insights into sagittal balance of the spine and to compare spinal load before and after spinal fusion.

Over the past years, the number of literature reports on spinopelvic measurements in patients undergoing spinal fusion has been increasing. Spinal fusion is the gold standard for lumbar degenerative disc disease (LDDD) but often results in loss of lumbar lordosis (flat back) and compensatory mechanisms causing increased pelvic tilt and sacral slope [1, 16]. Postoperative pain is highly correlated with increased PT after fusion [17–19]. Patients with an abnormal C7 plumb line have a higher rate of adjacent-segment degeneration after spinal fusion [20].

In this study, the initial sagittal balance of patients significantly influences the spinal load on L4–L5 level in compression and T12–L1 level in anterior–posterior shear forces.

Arshad et al. [13] simulated upper-body flexion with a very similar musculoskeletal model based on the patient of

Fig. 3 Differences in slopes of spinal disc force progression of dynamic trails as a function of sagittal displacement

Wilke et al. [21] At L4–L5, the proximo-distal force was slightly higher (639 N) in the upright standing model than in the vivo measurement (617 N). The models in our study had an average compression force of 545 N pre-surgery and 606 N post-surgery in the level L4–L5. The patients described by Wilke et al. and Arjmand et al. were healthy adults, aged 45 and 52 years, who had no degenerative disease of the spine. In contrast, our patients were elderly with a mean age of 62 years, who had a history of spinal diseases that often cause extreme spino-pelvic geometry and thus a higher spinal load than that in healthy subjects.

The missing decrease in spinal load of joints within the fusion zone after surgery maybe partly caused by the lumbar intervertebral joints that are blocked in motion by the driver in the model options and subject to full force transmission, such as a completely fused motion segment, but braces are neglected. The braces will bear parts of the load going through the specific joints, subsequently lowering the forces in the affected intervertebral joints. But because the investigation of the fusion was a secondary aim of the study, braces and cages that may influence the data obtained for evaluating sagittal balance were not implemented in the musculoskeletal model. Nevertheless, the important spino-pelvic parameters, such as lumbar lordosis, pelvic tilt, and sacral slope, were included in the study as well as the height and weight of the specific length-mass scale including the fat percentage. Therefore, the model positioning itself gave an accurate but not exact portrayal of the patient. The sagittal balance of the musculoskeletal model is mostly governed by spino-pelvic parameters. As already mentioned, the model provided an accurate image of the patient because lumbar lordosis, pelvic tilt, and sacral slope were given as patient parameters.

Sagittal balance was computed for all models in the same manner, equalizing such deviations. Sagittal balance as defined in the literature does not consider the parameter ‘patient height’. Therefore, sagittal balance had to be normalized with patient height to obtain a universal parameter. In our study, sagittal balance or its displacement is given as a percentage of body height. A higher percentage of sagittal displacement leads to higher AP shear (T12–L1) and compression forces (L4–L5) in the intervertebral joints. However, no statistically significant relationship could be found for compression in T12–L1 and AP shear in L4–L5 indicating a less pronounced dependency on sagittal displacement. Interestingly, AP shear decreased with increased sagittal displacement in this model. This does not support the clinical routine, as especially in the thoracic-lumbar junction adjacent segment diseases are often observed. Thus, this should be investigated in further studies.

Furthermore, the length-mass-fat scaling law leads to a linear distribution of weight along the body, while patients may have a non-linear distribution of weight. The inclusion of vertebral geometry data showed, as Putzer et al. already stated, the influence of vertebral geometry on spinal load [10]. But geometry alone does not improve model positioning. A non-linear scaling law or modeling according to a body-surface grid could result in more accurate spinal geometries and intervertebral joint forces. Nonetheless, the given data clearly show the trend of higher deviations from the balance causing higher spinal load in certain parts of the spine. Transferring our results to clinical routine shows on the one hand, that not regarding the sagittal balance increases the compression forces in L4–L5. That confirms the clinical routine, that in the lower lumbar spine is important to restore the sagittal balance. This effect can be explained as 70% of the global lumbar lordosis is out of L4–S1 [7].

On the other hand, in contrast to the clinical routine, in the critical level of the thoraco-lumbar junction Th12–L1, our results show that the higher the imbalance of the spine, the less the developing AP shear forces. As it is contrary to the observed developing of adjacent segment diseases, this area should be investigated in further studies.

The study has some limitations. Data on the intervertebral joints were not available, which may lead to deviation in sagittal balance from the real model. A strength is the usage of a validated and standardized musculoskeletal model.

Despite these limitations, the data obtained by means of these simulations provide good insights into spinal load and its influencing factors. Locking vertebral joint motion alone did not give a perfect portrayal of spinal fusion; therefore, further simulations with specific braces and cages and a more patient-specific modeling through non-linear scaling laws may give a more detailed musculoskeletal model for more accurate predictions of spinal load after fusion. Overall, musculoskeletal models may only be suitable for evaluating the effect of spinal fusion, if the model contains segments depicting the surgical braces and cages applied to the human spine to absorb the developing forces.

Conclusion

Improved restauration of sagittal balance is accompanied by the lower spinal load in L4–L5, but in Th12–L1 an increase in sagittal displacement even leads to a decrease in AP shear forces. From a biomechanical point of view, sagittal balance is the most important spino-pelvic parameter for spinal fusion, but every level should be seen separately as they development of loads is different for each level.

Compliance with ethical standards

Conflict of interest The authors declare that they have no competing interests.

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