Biomechanical study of anterior spinal instrumentation configurations

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Abstract The biomechanical impact of the surgical instrumentation configuration for spine surgery is hard to evaluate by the surgeons in pre-operative situation. This study was performed to evaluate different configurations of the anterior instrumentation of the spine, with simulated post-operative conditions, to recommend configurations to the surgeons. Four biomechanical parameters of the anterior instrumentation with simulated post-operative conditions have been studied. They were the screw diameter (5.5–7.5 mm) and its angle (0° – 22.5°), the bone grip of the screw (mono–bi cortical) and the amount of instrumented levels (5–8). Eight configurations were tested using an experimental plan with instrumented synthetic spinal models. A follower load was applied and the models were loaded in flexion, torsion and lateral bending. At 5 Nm, average final stiffness was greater in flexion (0.92 Nm°) than in lateral bending (0.56 Nm°) and than in torsion (0.26 Nm°). The screw angle was the parameter influencing the most the final stiffness and the coupling behaviors. It has a significant effect

($p \leq 0.05$) on increasing the final stiffness for a 22.5° screw angle in flexion and for a coronal screw angle (0°) in lateral bending. The bi-cortical bone grip of the screw significantly increased the initial stiffness in flexion and lateral bending. Mathematical models representing the behavior of an instrumented spinal model have been used to identify optimal instrumentation configurations. A variation of the angle of the screw from 22.5° to 0° gave a global final stiffness diminution of 13% and a global coupling diminution of 40%. The screw angle was the most important parameter affecting the stiffness and the coupling of the instrumented spine with simulated post-operative conditions. Information about the effect of four different biomechanical parameters will be helpful in preoperative situations to guide surgeons in their clinical choices.

Keywords Anterior instrumentation · Biomechanical testing · Coupling · Follower load

Introduction

Spinal surgeries are indicated for pathologies and traumas such as tumor, fracture, osteoporosis, congenital malformation and scoliosis. Severe cases are usually treated by a surgical posterior or anterior instrumentation and a spinal fusion. Even though the posterior instrumentation is currently the golden standard, no clear consensus exists among the orthopedic community, especially for thoracic surgeries [2]. In the past, the introduction of the first anterior instrumentation by Dwyer in 1964 [8] allowed for less instrumented levels, therefore for a greater final

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mobility of the spine [2, 16]. More recently, in the nineties, the introduction of the endoscopy in spinal surgeries [23] allowed for less post-operative morbidity and faster patient recovery [2, 7]. However, even if the anterior instrumentation is more attractive than ever, orthopedic surgeons still hesitate to use it mainly because of the loss of correction in post-operative situation, which is more likely to happen than with the posterior instrumentation [2, 7]. More studies are therefore necessary to see if it is possible to optimize the configurations of the anterior instrumentation in post-operative conditions.

Many parameters are affecting the biomechanical performance of the instrumentation as shown by the variety and quantity of studies. Some have been performed to compare the stiffness of different anterior instrumentations and they showed great differences [5, 13, 25, 26]. Others have shown stiffer configurations using multi-rods [9, 18, 30] and better grip on the bone for a bi-cortical anchor (versus mono-cortical) and for longer screws using pullout tests [12]. However, no studies on screw positioning for anterior instrumentation were found, though some exist for the posterior instrumentation [1, 4, 6].

Simulations of post-operative conditions have never been incorporated into the biomechanical tests of an anterior instrumented spine such as the use of a follower load [21] or the use of synthetic graft. Finally, such as the natural coupling, which is a 3D spinal movement that occurs because of the complex geometry of the spine, non-physiological coupling could be generated on the spine by the instrumentation configuration. This phenomenon has never been taken into consideration as a parameter affecting the biomechanical performance and could depend on many factors such as the number and location of the levels instrumented and the position of the implants on the spine.

The goal of this study is to evaluate the stiffness and the coupling behaviors of different configurations of the anterior instrumentation of the spine, in post-operative conditions, and to recommend instrumentation configurations to the orthopedic surgeons.

Materials and methods

An experimental setup allowing stiffness and coupling measurements of an instrumented spinal model has been developed (Fig. 1). Stacked cylinders [1] composed of two materials simulating cortical and cancellous bone (Sawbones Pacific Research Laboratories, Vashon, WA) were used to approximate mechanical and geometrical properties of human vertebrae body

[17, 28, 29]. An outside diameter of the cylinders of 35 mm was chosen as an average size of the T4-L2 segment that is usually instrumented [20].

Eight different configurations (#1–#8 in Table 1) have been tested following a Box, Hunter and Hunter [3] experimental plan in order to evaluate four parameters: the screw diameter (5.5 or 7.5 mm) and its angle (0° or 22.5° with respect to the coronal plane), the bone grip of the screw (mono or bi cortical) and the number of instrumented levels (5 or 8). CD Horizon Eclipse screws (Medtronic Sofamor Danek, Memphis, TN) of 25 or 40 mm in length were used to simulate the mono and bi cortical grips respectively. Parameters ranges represented typical, minimal and maximal, limits used during surgeries performed by experienced local surgeons. Each level was instrumented the same way using standardized templates to insure repeatability in the preparation of the models. Synthetic graft of bone like material (Sawbones Pacific Research Laboratories, Vashon, WA) cut to a diameter of 2–4 mm was used to fill the disk spaces simulating a graft taken from the patient iliac crest bone.

End cylinders were firmly fixed in their receptacle using a low melting point metal (Metspec158; MCP Metalspecialties, Fairfield, CT). The bottom receptacle was fixed on the base of the loading apparatus. A follower load [21] simulating the loading condition of the spine in the erect position was introduced. It consisted of a cable attached on the top vertebra and going down through a serie of pairs of holes drilled in all the vertebral bodies and of a mass of 22.5 kg [14] suspended at the lower extremity of the cable.

The spinal models were loaded in flexion, torsion and lateral bending using a servo-hydraulic material testing system (Bionix 858; MTS Systems Corp., Minneapolis, MN). The loads were applied on the top vertebra using cables in order to insure a free non-restricted motion (Fig. 1). A 3D optical camera (Visualize; PTI, Phoenix, AZ) linked to a computer, to a hand-pointer and light-emitting diodes has been used to measure the real 3D motion of the top vertebra. Three diodes were fixed on this vertebra to define a local coordinate system. The axis of the vertebral body was then found and the angle of rotation computed using a fixed reference. Moments were computed using the forces applied times the calculated lever arms.

Seventy-two experimentations have been performed including the three loading cases and the eight configurations tested, repeated three times each to measure the variability. Three curves were calculated per experimentation: the rotation versus moment in the direction of the application of the force and the two coupling rotations in the orthogonal directions.

Fig. 1 Experimental setup

Table 1 Configurations tested (#1–#8)

Configurations	Screw diameter (mm)	Screw angle (°)	Screw grip on the bone	Number of instrumented levels
#1	5.5	22.5	Bi cortical	5
#2	7.5	22.5	Bi cortical	8
#3	5.5	0	Bi cortical	8
#4	7.5	0	Bi cortical	5
#5	5.5	22.5	Mono cortical	8
#6	7.5	22.5	Mono cortical	5
#7	5.5	0	Mono cortical	5
#8	7.5	0	Mono cortical	8

Fig. 2 Typical moment rotation curves (configuration #8). **a** Flexion *F*, **b** lateral bending *LB*, **c** torsion *T*

measured in order to verify the behavior of the synthetic spinal model under the follower load only. They were prepared exactly the same way as the instrumented spinal models.

The effects and interactions of the variation of the four parameters, i.e. the variation of the instrumentation configurations, on the stiffness and the coupling were computed. ANOVA statistical analysis was used (Statistica; Statsoft, Tulsa, OK) to identify the significant effects ($p \leq 0.05$) and to obtain polynomial mathematical models of the spinal model behavior of the following form.

$$Y = \beta_0 + \beta_1.X_1 + \beta_2.X_2 + \beta_3.X_3 + \dots + \beta_{12}.X_1.X_2 + \beta_{13}.X_1.X_3 + \beta_{23}.X_2.X_3 + \dots + \varepsilon$$

where Y are the experimental responses measured, i.e. stiffness (K_i and K_f) and coupling, X are the four parameters studied, β are the polynomial coefficients and ε the experimental error.

These models were then used in the search of optimal instrumentation configurations. Two criteria have been retained for the aimed configurations: (1) maximum initial and final stiffnesses (K_i and K_f) and (2) minimum coupling. Because physiological movements done by human body are never pure ones, global stiffness and coupling, using the resulting vector of the three loading cases, have been calculated for comparison of the different instrumentation configurations.

Results

Non-linear behaviors with stiffness gradually increasing were found for principal and coupling responses (Fig. 2). At 5 Nm, average final stiffness (Fig. 3) for the eight different configurations tested (#1–#8) was greater in flexion (0.92 Nm/°) than in lateral bending (0.56 Nm/°) and than in torsion (0.26 Nm/°).

Average coupling measured was higher for the flexion (0.38) loading case than for the lateral bending (0.17) and the torsion (0.12) loading cases (Fig. 4a–c). Configurations with screws oriented at 22.5° (#1, #2, #5 and #6) were showing more coupling for a lateral bending motion (T and F couplings in Fig. 4b) and for a flexion motion (LB coupling in Fig. 4a). Coupling was quite negligible for the two non-instrumented models with an average of 0.05 for the three typical loading cases.

The screw angle was the parameter increasing the most the final stiffness. In flexion a 22.5° screw angle significantly increased the final stiffness (Fig. 5). In lateral bending a 0° screw angle significantly increased the final stiffness but was 1.6 times less important. The initial stiffness in flexion and lateral bending was significantly increased by the bi-cortical bone grip of the screw. All the p values observed for the previous effects were lower than 0.000019.

Statistical analysis ANOVA showed that the screw angle was also the most important parameter increasing the coupling behaviors. It also showed that this coupling behavior was the opposite of the stiffness behavior. In lateral bending a 22.5° screw angle significantly increased the coupling. In flexion, a 0° screw angle significantly increased the coupling.

Maximum stiffness (K_i and K_f) and minimum coupling have been calculated using the polynomial models and the most important parameter found. Therefore, a variation of the screw angle from 22.5° to 0° gave: (1) a global final stiffness diminution of 13% (Fig. 6) and (2) a global coupling diminution of 40% (Fig. 7).

Three repetitions per experiments have been performed in order to evaluate the experimental error (factor ε of the previous polynomial equation). This factor was computed by Statistica and has shown a standard deviation of 5–8%, depending on the loading case studied. Also, important significant effect of the

Fig. 3 Final stiffness (Nm/°) at 5 Nm

Fig. 4 Coupling (Nm/°). **a** Flexion *F*, **b** lateral bending *LB*, **c** Torsion *T*

Fig. 5 Importance of the parameters on the final stiffness for a flexion loading case

screw angle parameter and good correlation coefficient of the polynomial models (10/14 were higher than 84%) have demonstrated the quality and the repeatability of the measures.

Discussion

This study examined different anterior spinal instrumentation configurations. These configurations were based on the variation of four parameters that are commonly pre-determined by orthopedic surgeons in pre-operative situations.

Non-physiological coupling has been considered as being an important factor that could over stress the instrumentation and potentially lead to problems such as higher non-fusion rate or longer fusion time. As an example, the natural rotation, which is produced on the spine following a lateral bending will be amplified, as shown in Fig. 4b, if screws are put in place with angles (configurations #1–2 & #5–6). Because of the non-lordosis and non-kyphosis models used, coupling was totally generated by the instrumentation configuration as expected and confirmed by the two non-instrumented models tested, which have shown near zero coupling.

The configurations tested were not typical of real spine surgeries configurations where angles, diameter and bone grips of the screws are not instrumented the same way for each level as in this study. More options are also available to surgeons such as the use of two screws in a same vertebra, which is more common in fracture treatment, or multi-rods configurations. The surgical choice of these parameters is driven by the presence or not of a spinal pathology, such as a scoliosis, its severity, the surgical access, the mineral density of the bones and the proximity of the aorta. However, having all the levels instrumented the same way helped to characterize the effect of the four parameters targeted which mainly showed that the coupling and the stiffness have an opposite behavior when varying the angle of the screws. The configurations were stiffer with screws oriented at a 22.5° angle but a better stiffness distribution and much less coupling existed with screws oriented in the coronal plane. This suggests that the anterior instrumentation should be put in place with axis of the screws aligned as close as possible with the coronal plane. However, for the two screw angles tested, important torsion coupling was always occurring in flexion (Fig. 7). Bi-cortical grip on bones should also be employed to assure a better initial stiffness while the effects of the screw diameter and the number of levels instrumented were much less important (Fig. 5).

In biomechanical experiments, it is possible to fully constrain the displacement of the spinal model [32] or not [11]. In this study, the displacement [10, 24] has not been fully constrained and was achieved by the use of cables applying the loads on the top vertebra. This

Fig. 6 Final stiffness for a variation of the angle of the screw from 22.5° (left) to 0° (right)

Fig. 7 Coupling for a variation of the angle of the screw from 22.5° (left) to 0° (right) (Flexion F, lateral bending LB, torsion-T)

method has shown important residual angles (Fig. 2) due to the fact that the follower load was the only force pulling back the model to its initial position. But results were computed only from the loaded portion of the curves, eliminating any effects from these residual angles. In the literature, follower loads have been confined to non-instrumented cadaver spine [21, 24]. Our study is the first one using a follower load on an instrumented spine to simulate the post-operative erect conditions.

Other experimental studies using human and calf cadaver spines with anterior instrumentation are also showing, as in this study, that the stiffness is greater in flexion than in lateral bending than in torsion [9, 26, 27]. This suggests that the synthetic spinal models instrumented with standardized templates were valid even if they were representing non-pathological vertebrae, without osteoporosis or other spinal pathology. This choice of model helped for the grip of the screw in the bone and helped to reduce the biological and geometrical variability that would have occurred with animal or human cadaver spines [12, 15, 25]. All these simplifications aimed for a better repeatability for the characterization of the four instrumentation parameters. However, this type of model has its own limitations such as a simplified vertebra shape and the absence of thoracic kyphosis, scoliotic curvatures, muscles, facet joints and posterior elements. Regarding these simplifications, it has been shown that the thoracic kyphosis lowers the stiffness value measured on a synthetic model [1] while in vivo, idiopathic scoliotic

curves show an apical lordosis, which destabilizes the spine especially for rotation and supports the frontal displacement observed. No scoliotic curvature was used on our synthetic spines to reflect the result of a scoliosis surgery. Regarding the absence of muscles, it is well known that they are affecting the vertebral rotations, less in flexion and torsion than in lateral bending [19, 22]. Finally, facet joints and posterior elements are allowing approximately 5° of rotation per level [31], which would have given 40° of rotation for eight levels under a torsion load. Experimental values measured were not exceeding 25°, well under this value. Globally, it is difficult to estimate the combined effect of these simplifications. It is therefore important to realize that these experiments, if performed in vivo, could give different behaviours and cannot be simply transferred to the clinical settings.

Conclusions

The effects of four parameters of the anterior spine instrumentation have been studied. This study is the first one that considers the coupling and post-operative conditions in the search of optimal spinal configurations. The screw angle is the most important parameter affecting the final stiffness and the coupling behavior. The initial stiffness is affected by the bone grip of the screw. This study showed that the anterior instrumentation should be put in place with axis of the screws aligned as close as possible with the coronal plane.

Bi-cortical grip on bones should also be employed. However, all these conclusions are based on experiments performed on synthetic spinal models.

Prospective studies on patients who underwent anterior spinal surgeries could be performed to determine the effect of the amplified coupling on the rapidity and quality of their recoveries. Experiments on cadaver spines also should be conducted to confirm these conclusions. The patients and the surgeons would then fully appreciate the benefits of the anterior spinal instrumentation and its less invasive surgical procedure.

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References

- Belmont P, Polly D, Cunningham B, Klemme W (2001) The effect of hook pattern and kyphotic angulation on mechanical strength and apical rod strain in a long-segment posterior construct using a synthetic model. *Spine* 26:627–636
- Betz R, Shufflebarger H (2001) Anterior versus posterior instrumentation for the correction of thoracic idiopathic scoliosis. *Spine* 26:1095–1100
- Box G, Hunter W, Hunter J (1978) Statistics for experimenters. Wiley, New York
- Brodke D, Bachus K, Mohr A, Nguyen BK (2001) Segmental pedicle screw fixation or cross-links in multilevel lumbar constructs: a biomechanical analysis. *Spine* 1:373–379
- Brodke D, Gologly S, Bachus K, Mohr A, Nguyen BK (2003) Anterior thoracolumbar instrumentation: stiffness and load sharing characteristics of plate and rod systems. *Spine* 1794–1801
- Choma T, Chwirut D, Polly D (2001) Biomechanics of long segment fixation: hook patterns and rod strain. *J Spinal Disord* 14:125–132
- Crawford A (2004) Anterior surgery in the thoracic and lumbar spine: Endoscopic techniques in children. *J Bone Joint Surg Am* 86:2752–2673
- Dwyer AF, Newton NC, Sherwood AA (1969) An anterior approach to scoliosis: A preliminary report. *Clin Orthop* 62:192–202
- Fricka K, Mahar A, Newton P (2002) Biomechanical analysis of anterior scoliosis instrumentation: differences between single and dual rod systems with and without structural support. *Spine* 27:702–706
- Goel V, Ilder D, Pope M (1995) Biomechanical testing of the spine: load-controlled versus displacement-controlled analysis. *Spine* 20:2354–2357
- Grassmann S, Oxland T, Gerich U, Nolte L (1998) Constrained testing conditions affect the axial rotation response of lumbar functional spinal units. *Spine* 23:1155–1162
- Hitchon P, Brenton M, Coppes J, From A, Torner J (2003) Factors affecting the pullout strength of self-drilling and self-tapping anterior cervical screws. *Spine* 28:9–13
- Hitchon P, Goel V, Rogge T, Grosland N, Torner J (1999) Biomechanical studies on two thoracolumbar implants in cadaveric spines. *Spine* 24:213–218
- Joncas J, Labelle H, Poitras B, Duhaime M, Rivard CH, Leblanc R (1996) Dorso-lumbar pain and idiopathic scoliosis in adolescence. *Ann Chir* 50:637–640
- Liljenqvist U, Hackenberg L, Link T, Halm H (2001) Pullout strength of pedicle screws versus pedicle and laminar hooks in the thoracic spine. *Acta Orthop Belg* 67:157–163
- Lowe T, Betz R, Lenke L, Clements D, Harms J, Newton P, Maher T, Merola A, Wenger D (2003) Anterior single-rod instrumentation of the thoracic and lumbar spine: saving levels. *Spine* 28:208–216
- Mullet H, Odonnell T, Felle P, O'Rourke K, Fitzpatrick D (2002) Development of a model for occipital fixation: validation of an analogue bone material. *Proc Inst Mech Eng* 216:37–42
- Oda I, Cunningham B, Lee G, Abumi K, Kaneda K, McAfee P (2000) Biomechanical properties of anterior thoracolumbar multisegmental fixation. *Spine* 25:2303–2311
- Panjabi M, Abumi K, Duranceau J, Oxland T (1989) Spinal stability and intersegmental muscles forces. A biomechanical model. *Spine* 14:194–200
- Panjabi M, Takata K, Goel V, Federico D, Oxland T, Duranceau J, Krag M (1991) Thoracic human vertebrae: quantitative three-dimensional anatomy. *Spine* 116:888–901
- Patwardhan A, Havey R, Meade K, Lee B, Dunlap B (1999) A follower load increases the load-carrying capacity of the lumbar spine in compression. *Spine* 24:1003–1012
- Quint U, Wilke H, Shirazi-adl A, Parnianpour M, Loer F, Claes L (1998) Importance of the intersegmental trunk muscles for the stability of the lumbar spine. A biomechanical study in vitro. *Spine* 23:1937–1945
- Regan J, Mack M, Picetti G (1995) A technical report on video-assisted thoracoscopy in thoracic spinal surgery: preliminary description. *Spine* 20:831–837
- Rohlmann A, Neller S, Claes L, Bergmann G, Wilke HJ (2001) Influence of a follower load on intradiscal pressure and intersegmental rotation of the lumbar spine. *Spine* 26:557–561
- Shimamoto N, Kotani Y, Shono Y, Kadoya K, Abumi K, Kaneda K, Minami A (2001) Biomechanical evaluation of anterior spinal instrumentation systems for scoliosis: in vitro fatigue simulation. *Spine* 26:2701–2708
- Shimamoto N, Kotani Y, Shono Y, Kadoya K, Abumi K, Minami A, Kaneda K (2003) Static and dynamic analysis of five anterior instrumentation systems for thoracolumbar scoliosis. *Spine* 28:1678–1685
- Spiegel D, Cunningham B, Oda I, Dormans J, McAfee P, Drummond D (2000) Anterior vertebral screw strain with and without solid interspace support. *Spine* 25:2755–2761
- Szivek JA, Thomas M, Benjamin JB (1993) Characterization of a synthetic foam as a model for human cancellous bone. *J appl Biomater* 4:269–272
- Takemura Y, Yamamoto H, Tani T (1999) Biomechanical study of the development of scoliosis, using a thoracolumbar spine model. *J Orthop Sci* 4:439–445
- Wattenbarger J, Herring J, Bronson D, Ashman R (2001) Mechanical testing of a single rod versus a double rod in a long-segment animal model. *J Spinal Disord* 14:232–236
- White A III, Panjabi M (1990) Clinical biomechanics of the spine. JB Lippincott Company, Philadelphia
- Wilke H, Antonius R, Neller S, Schulthei M, Bergmann G, Friedmar G, Claes L (2001) Is it possible to simulate physiologic loading conditions by applying pure moments? a comparison of in vivo and in vitro load components in an internal fixator. *Spine* 26:636–642