THE NORM AND PATHOLOGY IN SPINAL BIOMECHANICS: MAIN ASPECTS OF STUDIES*

A.Y. Mushkin[1], E.V. Ulrikh[2], I.V. Zuev[3]
[1]*Saint Petersburg Research Institute of Phthisiopulmonology, Ministry of Health of Russia, St. Petersburg, Russia*
[2]*Saint Petersburg State Pediatric Medical University, Ministry of Health of the Russian Federation, St. Petersburg, Russia*
[3]*A.L. Polenov Russian Research Neurosurgical Institute, St. Petersburg, Russia*

The review focuses on the main aspects of spinal biomechanics, such as the strength of anatomical structures forming the anterior and posterior spinal columns, kinematic properties of spinal motion segments and spinal regions, biomechanics of spinal deformities, remodeling of bone graft under the deformity correction, biomechanics of implants and instrumented spine.
Key Words: spine, biomechanics of spine, spinal deformities, scoliosis, kyphosis, instability, adjacent segment pathology, spine fixation, instrumental fixation.

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The spine is a complex three-dimensional system; its anatomic properties allow motion in any single plane as well as in many planes simultaneously. Most of the spine deformities are multiple ones: the torsion accompanies typical scoliosis and kyphosis [8, 13, 15, 20, 22, 25, 39, 40, 44, 76]. One-plane sagittal and frontal deformities, the so-called kyphoses, lordoses and scolioses, occur less frequently and are usually congenital (or associated with vertebral anomalies). However, they are considered to be simple in the context of biomechanics. This simplicity makes it easy to analyze them using theoretical and computational methods.

There are five major trends in the research field of spine biomechanics:

1) strength of the anatomical structures of the anterior and posterior spinal columns. The two-column model of spine structure proposed by F.W. Holdsworth [74, 75] is usually employed in these works;

2) kinematics of the isolated spinal motion segments (SMS) and the vertebral column regions;

3) biomechanics of spinal deformities.

4) remodeling of bone grafts under deformity correction.

5) biomechanics of grafts and spine with instrumental fixation.

The major aspects of these studies are of interest.

The strength parameters of anatomical structures of the anterior and posterior spinal columns. According to the definition proposed by A.I. Kazmin etc. [8], each element of a vertebra (body, arch and processes) mostly plays a single functional role: the vertebral body is a structural frame; the vertebral arch plays a protective role; and the system of vertebral processes has the kinematic function.

Vertebral bodies, as shown in the mechanistic models, are able to resist the compressive loads that are several times higher than the human body's weight. For example, the maximum load for cervical, thoracic, and lumbar vertebrae is 150–300, 200–800, and 300–1300 kg, respectively [1, 2, 8, 23, 27, 28, 42, 92–95, 106, 115, 117]. Resistance of vertebral arches to compression is much lower; however, they also can resist a load comparable to the weight of an adult human.

The vertebral arch of young children is able to resist the compressive loads measuring up to 80 % body's weight. However, the increased compression results in epiphysiolysis of the arch. It is the reason why the use of spinal implants in young children was limited for an appreciable long time [16, 17, 24, 31, 71, 102].

Compressive loads experienced by intervertebral discs distribute non-uniformly between the nucleus pulposus and annulus fibrosus. Short-term loads of 50–100 kg produce neither compression nor deformation in nucleus pulposus and annulus fibrosus. The increase in compressive load or its duration results in elastic compression of nucleus pulposus and deformation of annulus fibrosus without any visible anatomic damage. The total load on the intervertebral disc in the lumbar spine may be as high as 450–2500 kg. Strength of intervertebral discs decreases in elderly people by a factor of 1.5–2.5 [8, 23, 71, 95, 108].

Resistance of discs to traction is much lower. The traction causes the damage to the cervical spine at a force of 100 kg, while the traction force of 400 kg damages the lumbar spine [8].

Research has shown that spinal ligaments under physiological conditions are preliminary stretched at forces of 0.12–0.35 kg (supraspinal ligaments) and 0.4–1.5 kg (yellow ligaments). Under natural spinal motion, these forces increase to 6–8 kg and 13–50 kg, respectively. The increase in these stretching axial forces by a factor of 3–4 results in a damage of ligaments, the loss of stability, and emergence of vertebral hypermobility. A decrease in the buffering function of intervertebral discs and a 40 % drop of pressure inside the nucleus pulposus are simultaneously observed [27, 28, 92, 94, 95, 108, 111, 115].

The kinematic properties of the spine. Each spinal motion segment (SMS) in the sagittal plane under stable conditions is a system of balanced levers with the point of support at the level of joints (Fig. 1).

Table 1 provides the data on physiological amplitude of motions in isolated SMSs and spinal regions in the sagittal plane [3, 8, 120–122].

The original physiological rigidity of the thoracic spine is ensured by strength of the costovertebral skeleton, as well as by spatial orientation of zygapophyseal joints. These facet joints are positioned in the frontal plane (with some inclination) in the cervical spine, and at an angle of 20–40° to the frontal plane in the thoracic spine. Sagittal orientation of zygapophyseal (facet) joints in the thoracolumbar and lumbar spine ensures significant mobility of these spinal regions. Removal of articular processes of the thoracic spine increases spine mobility the frontal plane by 7–50 %, in the sagittal plane – by 28–80 %, and in the horizontal plane — by 22–60 % [99, 119–121].

The studies on biomechanics of cervical SMSs [57, 58, 62, 100, 120] consider vertebral motions in the sagittal plane as a combination of translation (linear translocation or shift) and rotation (Fig. 2).

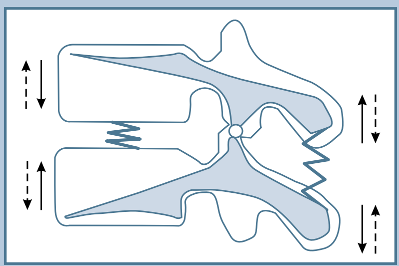

Fig. 1
Motions in spinal motion segment and functional activity of spinal ligaments: directions of the motions for bending and extending spine are indicated with the solid and dotted arrows, respectively [8]

Table 1.
Physiological amplitude (angle) of motions in isolated SMSs and spinal regions in the sagittal plane

Spinal regions	Spinal motion segment	Amplitude (angle) of motions, deg.	
		in one segment	total
Cervical	O—C1	12	64
	C1—C2	10	
	C2—C3	8	
	C3—C4	13	
	C4—C5	12	
	C5—C6	17	
	C6—C7	16	
	C7—T1	9	
Thoracic	T1—T2	4	35
	T2—T3	4	
	T3—T4	4	
	T4—T5	4	
	T5—T6	4	
	T6—T7	6	
	T7—T8	6	
	T8—T9	6	
	T9—T10	6	
	T10—T11	9	
	T11—T12	12	
	T12—L1	12	
Lumbar	L1—L2	12	65—90
	L2—L3	14	
	L3—L4	15	
	L4—L5	17	
	L5—S1	20	

With allowance for different amplitudes of natural motions in the sagittal plane and movements estimated in mechanistic models, the 'in vivo kinematics' concept has been proposed for quantitative analysis of motions produced by SMSs and the spine. "*In vivo* kinematics" is employed along with amplitude of arbitrary movements including the amplitude of motions produced due to the complete flexion and extension that are externally stimulated by an experimentalist [99].

Biomechanics of spine. All international publications concerning the treatment of spinal deformities have citations or references to the studies conducted by White and Panjabi [121–123]. M.B. Mikhailovsky was the first author who conducted a thorough analysis of these important works in Russian scientific literature.

According to the conception proposed by White and Panjabi, the spine is a stable semi-rigid cylindrical system (Fig. 3). The 'tolerance' concept reflects stability of the system, or projection of the center of gravity on the support. The model, originally developed for kyphosis, reflects factors that affect this kind of deformity. Kyphosis develops under the force of gravity Fg directed vertically and the bending force Ff directed forward or forward-downward (or downward-sideward in patients with scoliosis). The sum vector F1 is always directed at an angle and downward. The force F2 acting symmetrically on the lower point of the support also has the vertical and the horizontal components.

The system is stable if the sum vector of forces acting along the horizontal axis is zero. Force F3 maintains the system stability, acts on the apex of the kyphosis and is directed backward. If the system stability is lost (decompensation of the sum vectors of forces F1, F2 and F3), the center of gravity shifts, and the system begins to bend forward or backward. This corresponds to the decompensation of deformity in patients with scoliosis. The necessity of keeping a vertical pose results in the development of pose maintaining mechanisms, formation of compensatory spine curves (the reactive lordosis), the change in the angle of sagittal rotation of the sacral spine, lumbar-femoral angles, etc.

The roles of the axial (Fg) and horizontal (Ff) components in the development of deformity have been studied (Fig. 4).

The crosswise-directed bending force was found to play the leading role at the beginning stages of deformity; the roles of the axial and horizontal components of the force are equal at an angle of 53°, while the gravity component acquires the leading role as kyphosis progresses further.

The authors have actually distinguished between the concept 'stability',

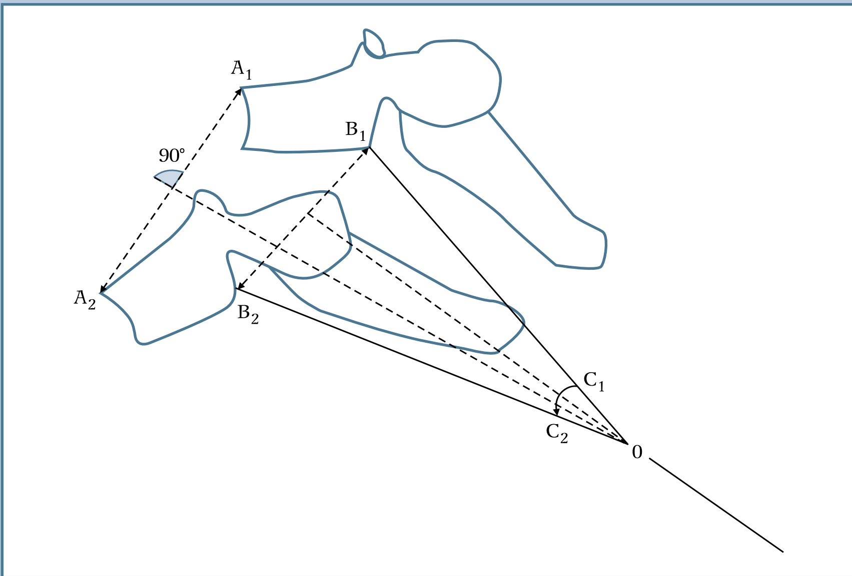

Fig. 2
Translocation of an isolated vertebra during kyphosis formation: *1 – original position of the vertebra; 2 – position of the vertebra in a patient with kyphosis; A1–A2 and B1–B2 – linear shift; C1–C2 – angular shift [100]*

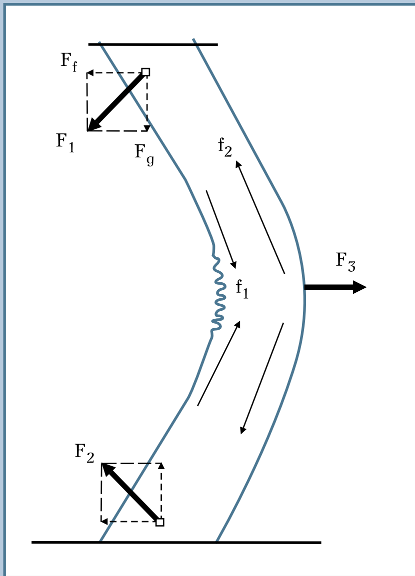

Fig. 3
Directions of the forces acting on spinal deformities [98]

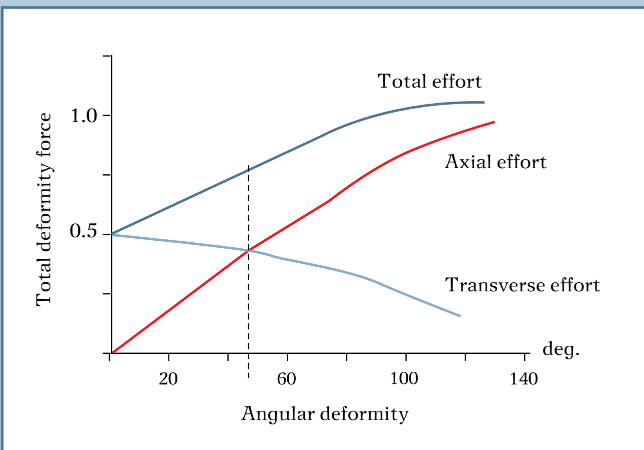

Fig. 4
Distribution of the forces acting on the spine at different deformity angles [99, 120–123]

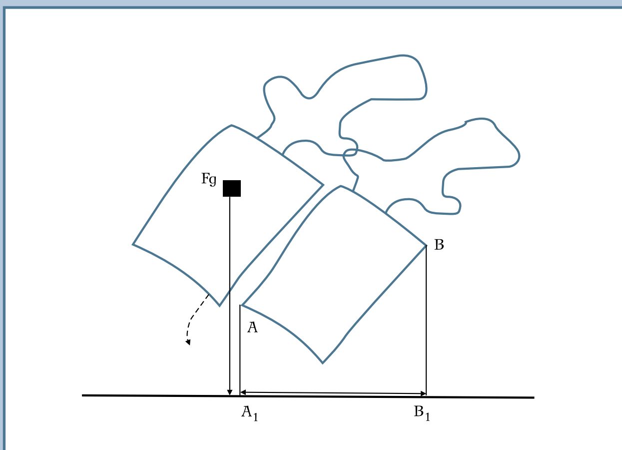

Fig. 5
Loss of support in SMS: displacement of the gravity center of the upper vertebra outside the projection of the lower vertebra onto the horizontal plane [99]

which is an ability of SMS to make normal physiologic motions under the loads, and the concept 'buttress', which is the situation whereby the gravity center of the cranial vertebra in SMS is projected onto the area of transversal section of the caudal vertebra. At high deformities, SMS stays stable but fails to provide the supporting point, since the center of gravity of the cranial vertebra is displaced forward from the area of transversal section of the caudal vertebra, resulting in a collapse of the entire upper section of kyphosis [99] (Fig. 5). It is not surprising that all orthopedists point out the collapse progression of naturally developing deformities larger than 50°, which is independent of frontal or sagittal curvature planes [8, 13, 21, 22, 26, 29, 30, 85, 86, 127].

The pathogenetic mechanisms of formation of angular sagittal deformities (humpbacks) are very similar to those described above but have some specific features. Destruction of one or several vertebral bodies causes the loss of vertebral support, brings the vertebral bodies closer together, and triggers the development of kyphosis. The tonic tension and fibrotization of prevertebral muscles, stretching of the articular capsules of zygapophyseal joints, formation of the vertebral segmental instability and the displacement of vertebral arches with respect to vertebral bodies that is typical of children, giving rise to specific deformities, may promote kyphosis progression [5, 7, 10, 11, 18, 51, 66, 67, 72, 73,106, 116, 129].

The model proposed by White and Panjabi defined the points of application and directions of action of the forces required to correct the spinal deformities; in doing so, it theoretically supports the so-called three-point correction model (Fig. 6). This model underlies using the brace wearers proposed for correction by Blount and Moe back in 1958 (the model was elaborated while relying on the empirical data) [45].

The use of the correcting brace wearers is based on combining the axial distraction and the horizontal pressure on the spine; stretching of the spine is provided by the external rods supported by the cervical and pelvic fixing rings, while the horizontal pressure on the apex of deformity is provided by the pulling belt with the use of the external pressure [109, 128]. The use of the modern hyperextension-producing wearers is based on the principle of three-point correction. It should be emphasized that the same principle is employed in manual surgical correction of kyphosis and in instrumental correction of spinal deformities with the use of multisegmental supportive fixtures [14, 18, 52, 54, 87–89].

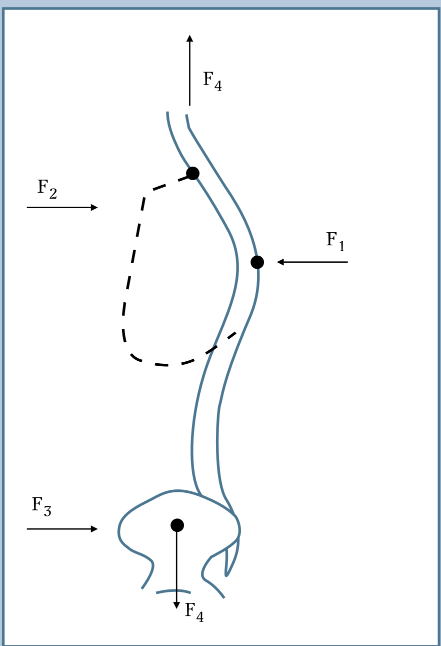

Fig. 6
Three-point model showing the orientations of the external forces applied to the spine to correct the kyphotic deformity [14, 45]

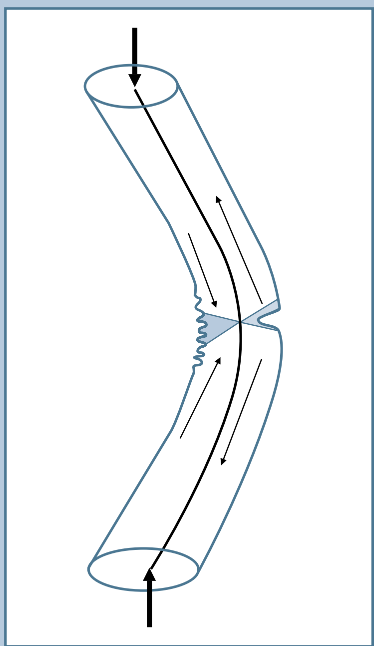

Fig. 7
Internal tension experienced by a semi-cylindrical rod [98]

Remodeling of bone grafts under the spinal deformity conditions. The spine with a deformity is a semi-rigorous system that experiences inner tensions (Fig. 7) acting as the forces of compression and strengthening on the concave and convex sides of the deformity, respectively. The indicated tensions are distributed non-uniformly over the system's cross-section. 60% of the cross-section of the concave region encounters compression, while 40% of the section on the convex side is affected by strengthening [43, 97, 98, 119, 121–123]. The conventional null line corresponding to the minimal internal tension is always displaced towards the convex side (dorsally in patients with kyphosis).

The changes developing in the spine after stabilization via osteoplastic surgery obey the two common biological rules: the Hueter-Volkmann law and the Wolff's axiom:

– in accordance with the Hueter-Volkmann law, the excessive load acting on a growing (or remodeling) bone results in dystrophy of the regions that experience the highest pressure (i.e., localize on the concave side of the arch);

– according to the Wolff's axiom, the bone remodeling in the fusion zone runs in accordance with the direction of applied forces. These forces are the gravity force and the tension forces of muscles attached to the bone. The bones under excessive load lose their strength, resulting in the formation of the areas of pathological reconstruction (pseudarthrosis) [30, 47, 48].

It has been shown experimentally that the mature isolated posterior fusion does not ensure complete immobilization of the fused region. This results in pathological graft reconstruction and pseudarthrosis formation not only in patients with severe angular deformities, but in those without any deformities as well [108, 120].

The problem of failure of fusion performed in patients with severe kyphotic deformity has been repeatedly discussed in literature [4, 9, 12, 19, 51, 126]. Studying the bone fusion formation in of the long-term period after treatment of kyphosis with the deformity over 50°, R.B. Winter with colleagues [126] revealed pseudarthrosis in 55 % of cases when the isolated posterior fusion was used and 15 % of cases with the combined anteroposterior stabilization via bone grafting. The high rate of complications in patients with severe angular kyphosis over 55° prompted surgeons to use the anterior bridge-like fusion with a spacer, employing cortical allo- and autografting with the highest possible ventral takeaway (Fig. 8) instead of using the interbody fusion with fixing the apical vertebrae. This provides the highest stability of the newly formed structure [14, 41, 46, 47, 86, 90, 97, 125, 126].

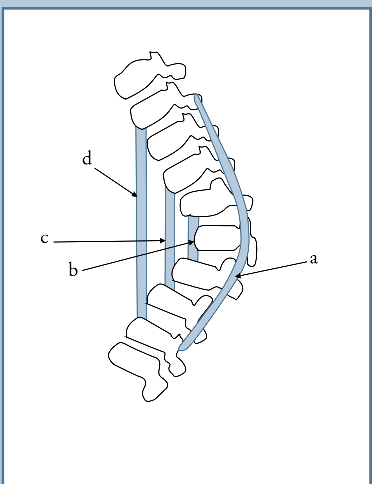

Fig. 8
Options for osteoplastic fixation of the spine: a – posterior fusion; b, c, d – various versions of anterior fusion; c, d – bridge-like fusion (spacer type) [14, 47, 86]

The range of possible ventral displacements of grafts is limited by the need to expand the operating field and by the conditions of graft vascularization: the more a graft is displaced forward and the smaller the area of contact with the recipient bed is, the worse conditions for its revascularization are. The pedicle bone grafting procedure is optimal in such cases [6, 48, 90].

Biomechanics of spinal grafts under the instrumental fixation conditions. The established directions of implant testing and studying the spine biomechanics under the instrumental fixation conditions have historically been associated with the stages of implementation of fixing and correcting devices. The studies performed within two decades, from the mid 1960s to the early 1980s were devoted to various aspects of spine biomechanics when the Harrington distraction rod was used [32, 49, 53, 56, 61, 70]. Dynamometry, tensiometry and telemetry demonstrated that the tension in the biomechanical system in patients with scoliosis reduced within 20 min after the Harrington distraction rod had been installed due to stretching of soft tissue structures of the spine. After 1 h, the tension that had originally ranged from 190 to 421 N reduced by 1/3 or 1/2. The step-by-step distraction during the surgery [32] or compensation of the rigidity loss using external brace wearers were recommended to be done to compensate for the tension loss of an internal fixator [56, 61]. The modeling of the Harrington distraction rod over the deformity contour results in a significant loss of fixation rigidity [53, 56]. The use of distraction rods with a long stepping part was fallacious, since the increase in length between the hooks resulted in the rise of the tension on the first groove (counting from the rod base) and frequent fractures of metallic parts in this spot. In this connection, the practice of using distraction rods with the stepping part being longer than 1/4 of the total length of the rod was discontinued [49].

In the late 1970–1980ss, the focus of spine biomechanics has moved from graft testing to the comparative analysis of rigidity in different spinal instrumental systems. The tension produced in the spinal biomechanical system by Harrington compression rods, Harrington distraction rods, and Weiss dynamic compression springs was found to be 350, 250, and 250 kg/cm, respectively [113]. Examination of the strength properties of the newly designed equipment Dwayer, Zielke, and especially Lugue, revealed their advantages over Harrington rods regarding the axial or torsion movements [60, 68, 69, 77, 79, 87–89, 120, 121, 130, 131].

A dramatic rise in the number of studies focused on the strength properties of the spinal fixation equipment was associated with the newly designed Cotrel – Dubousset instruments and their analogs: Isola, TSRH, Sinergy, etc. The advantages of the multi-hook systems over two-hook ones, as well as the advantages of the system with transperpendicular fixation of supportive elements over the hook-based systems were established. Moreover, the properties of individual elements of the systems (rods, plates, hooks, wires and transverse extenders) from different manufacturers were tested [36–40, 63, 80, 93, 105].

Over the past decade, significant attention has been focused on comparing the biomechanical properties of the anterior stabilization devices (plates, rods, and cages) employed in the surgeries on different regions of the spine, but more often the mobile cervical and lumbar spine [50, 55, 59, 65, 78, 91, 96, 107, 118, 124].

Abundance of the testing methods for the grafting instrumentation contributed to the development of the unified protocol for technological testing of the metal systems ASTM/ISO [34, 37, 38, 43]. This protocol is based on the unified coordinates system. In accordance with this system, the mobility of the spine under instrumental fixation is tested not only along the major orthogonal axes (vertical, sagittal and transverse) but around each of them as well. By this means the fixation rigidity is defined according to six types of movements. According to the protocol, the parameters for comparative evaluation of spinal grafts include the system's potential to fix deformities, the short-term and long-term stability of a newly developed system of the spine and the graft, and the system's impact on the surgically treated spinal segment during bone fusion.

The spinal instability after instrumental fixation, which is not associated with pseudarthrosis formation, and fracture or dislocation of metallic correcting systems, has also been discussed in medical literature. The occurrence of clinical symptoms typical of instability (pains and increasing deformities) under instrumental fixation conditions may arise from overload-produced damage of the supportive bone structures [53, 56, 61, 112, 114] and the development of pathological changes in SMS contacting with the fixation zone (conjunctional deformities). The latter fact is very important, since it is associated both with 30–60 % increase in natural overload of SMSs adjacent to the fixation (stabilization) zones [35, 64, 83, 84, 110] and with the iatrogenic collapse of the posterior soft tissue support due to symmetric positioning the supportive hooks on/under the vertebral arch [33, 81, 82, 84, 101, 103, 104].

We have not intentionally covered two other aspects of spinal biomechanics in this review: problems of balance in the body as a whole (in the sagittal and frontal planes) and the lumbosacral region in particular. Numerous publications of the past decade have been focusing on these subjects, which requires an individual comprehensive analysis.

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Corresponding author:
Mushkin Aleksandr Yur'evich
Children Surgery Clinic
St. Petersburg Research Institute of Phthisiopulmonology
Politechnicheskaya str. 32
St. Petersburg, 194064
E-mail: aymushkin@mail.ru

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A.Y. Mushkin, Doctor of Medicine, Professor, Head of the Department of Extrapulmonary Tuberculosis and the Department of Surgery of Bone Tuberculosis in Children; E.V. Ulrikh, Doctor of Medicine, Professor, Professor of the Department of Childhood Surgical Diseases; I.V. Zuev, Candidate of Medicine, Neurosurgeon of the Department of Childhood Surgical Diseases.