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Three-dimensional biomechanical properties of the human cervical spine in vitro

I. Analysis of normal motion

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Propriétés biomécaniques tridimensionnelles du rachis cervical humain in vitro. I. Analyse du mouvement normal

Résumé. L'objectif de notre étude est de déterminer le comportement mécanique du rachis cervical humain soumis á des charges physiologiques statiques. Les déplacements tridimensionnels dus à trois moments de couple purs (flexion-extension, inflexion latérale gauche-droite et torsion axiale gauche-droite), sont mesurés sur 56 unités fonctionnelles rachidiennes intactes (UF) de C2 à C7 prélevées sur 29 sujets. Les courbes effort-déplacement sont tracées pour chaque sollicitation. Nous calculons ensuite la zone neutre (ZN), la mobilité maximale (MM), le rapport de ZN à MM, le rapport du déplacement couplé au déplacement principal (RDC), le moment limite et la rigidité sécante. L'influence de la dégénérescence du disque intervertébral et du niveau d'UF sont aussi étudiées avec une analyse de variance (ANOVA). Nos résultats montrent bien la non linéarité des courbes effort-déplacement et la ZN du rachis cervical dans les trois plans de l'espace. Nous trouvons des différences significatives de rigidité entre trois sollicitations appliquées. Lorsque nous sollicitons en inflexion latérale nous observons des différences significatives de rigidité d'un niveau vertébral à l'autre. Mais la différence de rigidité concernant différents états de dégénérescence de disque n'est significative qu'en inclinaison latérale droite. Le RDC sous inflexion latérale et torsion axiale est significativement différent entre différents niveaux d'UF. L'influence du cycle d'effort et la réponse mécanique de C1–C2 en déplacement principal sont aussi présentées.

Mots-clés: Biomécanique – Rachis cervical – Courbe effort-déplacement – Zone neutre – Rigidité.

Summary. Our aim was to determine the biomechanical properties of the normal human cervical spine under physiological static loads. The three-dimensional displacements under three pure moments: flexion-extension, left-right lateral bending and left-right axial torsion – were measured in 56 intact functional spinal units (FSUs) taken from between C2 and C7 in 29 human cadavers. For each mode of loading, load-displacement curves were plotted. Then we calculated each neutral zone, range of motion, neutral zone ratio, ratio of coupled motion, limit moment and secant stiffness. The effects of intervertebral disc degeneration and the disc level were also taken into account by the analysis of variance. Our results adequately demonstrated both the non-linearity of load-displacement curves and the neutral zone of the cervical spine in three-dimensional space. At the same time, we found statistically that the stiffness in the three planes are significantly different, as are the stiffnesses in lateral bending of successive different FSUs. However, significant differences of stiffness in different states of disc degeneration were only found in right lateral bending. There were significant differences between levels in ratio of coupled motion under both lateral bending and axial torsion. The loading cycle conditions and the biomechanical responses of principal motion of C1–2 are also reported.

Key words: Biomechanics – Cervical spine – Load-displacement curves – Neutral zone – Stiffness

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The cervical spine is the most mobile region of the spine. It bears the weight of the head and protects the spinal cord. It is often the site of arthrosis [5], trauma [32] and surgery [7]. Biomechanical studies on the normal cervical spine have already been carried out by many authors. Lysell [17], Penning and Wilminck [30, 31] and others [1, 4, 6, 12, 21] have studied the three-dimensional kinematic patterns of the cervical spine without reference to loading conditions. Panjabi et al. [28] presented load-displacement curves for six types of force and the neutral zone in cervical functional spinal units (FSUs). The neutral zone (NZ) is that part of the motion range from the neu-

Table 1. Data on spinal segment specimens

Spine	Age (year)	Weight (kg)	Height (cm)	Level(s) and degeneration grade				
				C2–3	C3–4	C4–5	C5–6	C6–7
1	55	75	176		4		3	
2	67	90	176			4		4
3	51	72	170			3		
4	65	68	165	3		3		2
5	70	72	165		3		4	
6	58	64	168		4		4	
7	73	76	175					3
8	63	57	165	4		4		4
9	64	72	170		3		3	
10	58	–	180	3		3		3
11	61	–	175	2				
12	87	–	165	3		4		4
13	85	–	165	2				
14	62	78	175	2		2		2
15	68	60	164		4		2	
16	68	108	168		2			
17	44	45	162				2	
18	72	70	165		2		3	
19	67	68	170	2		2		
20	64	62	178	3		2		
21	61	63	155		2			2
22	56	66	170	2		2		2
23	75	60	155	2		2		3
24	83	75	169			3		
25	61	–	–			3		4
26	73	–	–		3			3
27	80	–	–			3		
28	63	–	–			2		2
29	68	–	–				4	
Average	66 (10)	70 (13)	169 (6.5)	2.5	3.0	2.8	3.1	2.9

Standard deviation in parentheses

tral position up to the point at which some resistance starts to be offered by the joint. Generally application of a small load causes a large deformation at the beginning of the articular movement [36]. Moroney et al. [23] worked on the load-displacement behaviour, in force as well as in moment, of 35 FSUs, some intact and some with posterior elements removed. The viscoelastic responses of the cervical spine have been reported by McElhaney et al. [18]. The non-destructive biomechanical responses of cadaveric cervical spine as well as their strength and the pattern of failure for quasi-static loads have been measured by Coffee et al. [2] and by Shea et al. [34]. However, there are few statistical reports of the mechanical properties of the human cervical spine.

In our study, experiments were performed under three pure static moments – flexion-extension, lateral bending and axial torsion – in 56 FSUs. The three-dimensional motions of the upper vertebrae corresponding to these loads were measured, and corresponding stiffness coefficients for principal motion were calculated. The NZ is an important characteristic which differentiates the cervical spine from the lumbar region. The effects of load cycles on the measurement of the mechanical properties, the influence of the degenerative state of the intervertebral disc and the differences of mechanical response in FSU from different levels were investigated with multiple parameters.

Materials and methods

Cadaver segments

Twenty-nine cervical spinal segments were taken between C2 and C7 from fresh human cadavers, radio-sterilised with 25 kGy β-radiation and stored in an air-tight bag at −24°C until utilisation. Data regarding cadaver age, sex, height, and weight were recorded for these specimens (Table 1). Each segment was visually examined and radiographed (anteroposterior, lateral) to rule out preexisting tumours, traumatic destruction or severe arthrosis. The basic anatomic unit for most biomechanical studies of the spine was the functional spinal unit [28]. In preparation for the tests, the muscular tissue was removed from each spinal segment before division into FSUs, care being taken to preserve all discoligamentous structures. Of the 56 FSUs tested, 11 were C2–3, 9 C3–4, 15 C4–5, 8 C5–6, and 13 C6–7 (Table 1). After testing was completed, the intervertebral discs were horizontally transected and visually assessed as to the degree of degeneration, using Nachemson's method [25].

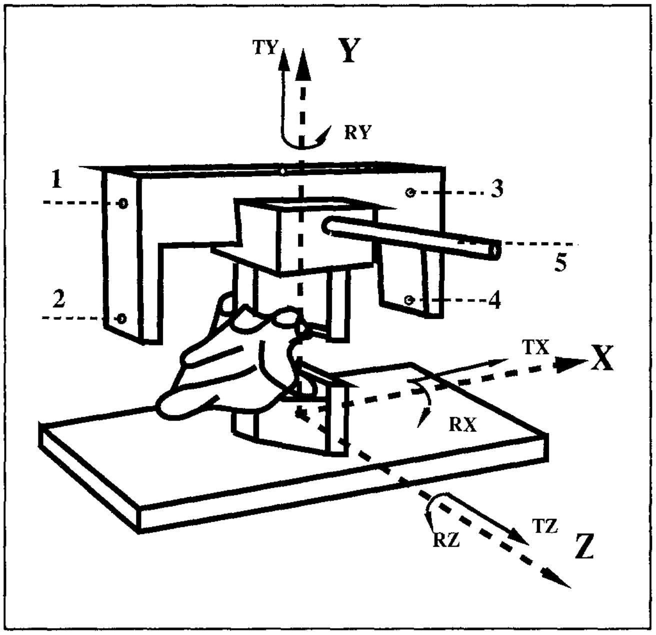

Fig. 1. The three-dimensional coordinate systems and experimental set-up. *1–4*, Attachment points for cables for application of a pair of forces. *5*, Connecting rod for two micrometric heads. *TX*, *TY*, *TZ*, Linear translation; *RX*, *RY*, *RZ*, angular rotation

Fixation of vertebrae

We designed two special devices in aluminium for fixation of the body of the upper and lower vertebrae of an FSU. Two metal screws were set into the upper and lower vertebral bodies. The vertebral bodies were embedded in a synthetic plaster cast (Microdice) with a high Brinell strength (1080 kg/cm^2) and 0.15 setting expansion. We could detach the vertebrae from the plaster and from the fixation device, not only for geometric measurement of the vertebrae but also for further use of the device.

Application of load

A U-shaped plate was rigidly attached to the fixation device of the upper vertebra. This permitted the application of three pure moments, flexion-extension (−RZ, +RZ), left and right lateral bending (−RX, +RX), left and right axial torsion (+RY, −RY). Each moment was applied by choosing an appropriate pair of cables attached to opposite loading arms (Fig. 1). No special preload was applied, but the weight of the upper fixation device (930 g) may be considered as a small axial compression preload. We tested each FSU under three loading moments. The increments and the maxima of loading were in the ranges 0.1–0.3 Nm and 1.4–4.5 Nm respectively. Each application of a moment consisted of 14 increments of continual loading in one direction about the corresponding axis and 14 decrements of continual unloading, then a repetition of the same loading-unloading pattern in the opposite direction of the same axis. There were, in total, 57 points of measurement (see Figs. 2 and 3). We applied the moments within a physiological range by relating the instantaneous results to the kinematic results reported by some authors. The same load-unload cycle was applied three times for each moment. Only the third application was recorded.

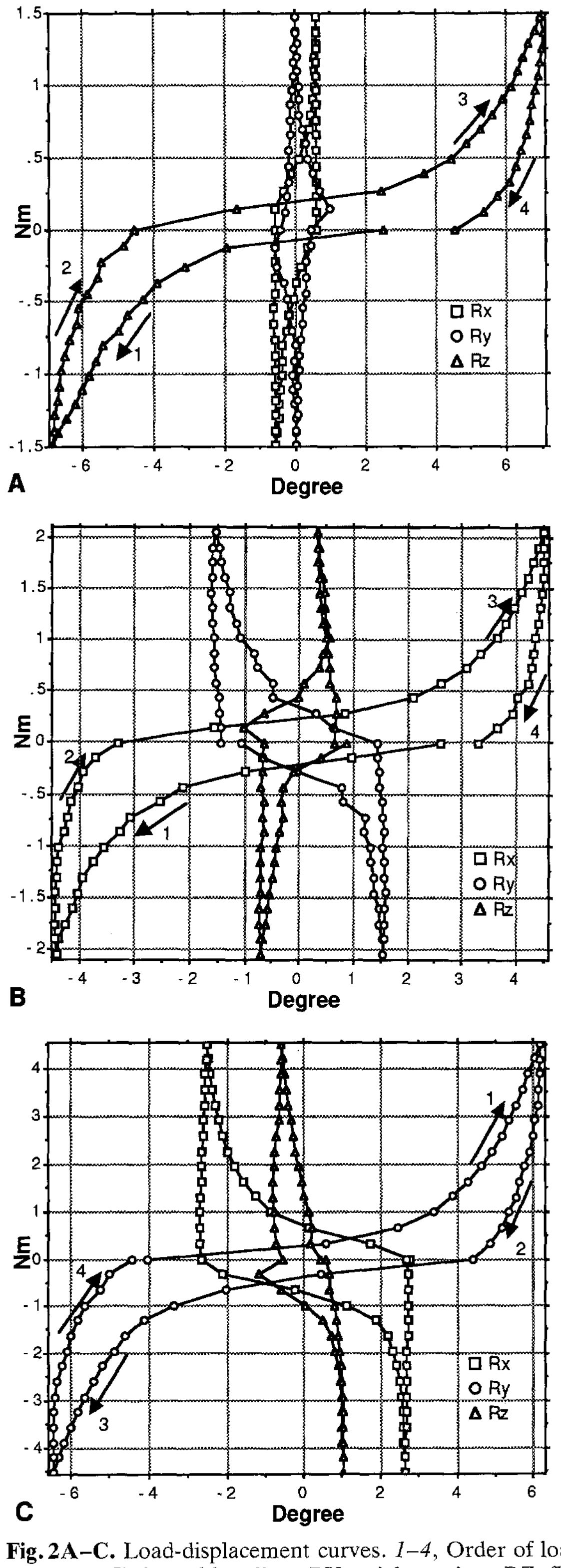

Fig. 2A–C. Load-displacement curves. *1–4*, Order of loading-unloading; *RX*, lateral bending; *RY*, axial rotation; *RZ*, flexion-extension. **A** Under flexion-extension moment (C4–5, spine 24). **B** Under lateral bending moment (C4–5, spine 24). **C** Under axial torsion (C3–4, spine 26). Both principal and coupled motions are plotted on the abscissa, while the load is shown on the ordinate

Measuring system

The set-up and the three-dimensional coordinate systems are shown in Fig. 1. The right-handed Cartesian orthogonal coordinate system was used for the measurement of displacements. The recording of data was performed at 15-s intervals after application of the load. The three-dimensional displacements of the upper vertebra in our study were measured using a system called "two micrometrical heads" [16]. This measuring system is joined to the upper vertebra through a connecting rod. One extremity of this rod is at-

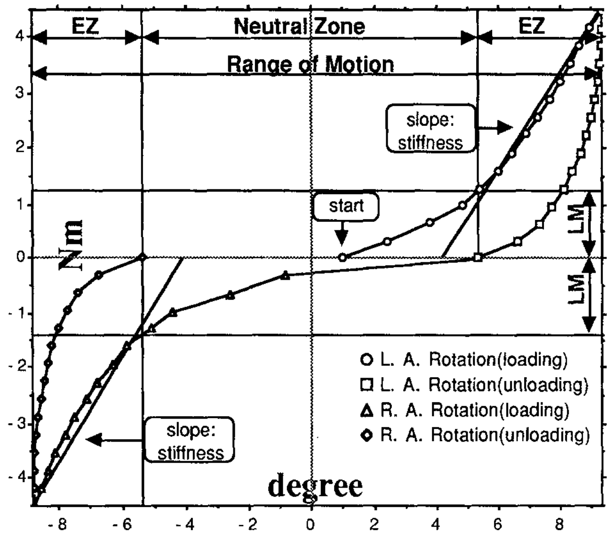

Fig. 3. Schema for calculating different parameters (under axial torsion). *LM*, Limit moment; *EZ*, (elastic zone)

tached to an articular joint which is fixed in one micrometrical head, made up of three electrical linear displacement transducers. The other extremity is attached to a universal joint fixed in a second micrometrical head, made up of three linear transducers and a rotatory transducer. The upper vertrebra and these two micrometrical heads have six degrees of freedom. The analogue values were converted into numeric data by an Orion converter (Schlumberger Solartron, 3530B) and then recorded and processed by a personal computer. Through the recordings of seven displacement transducers we calculated three rotatory displacements and three linear displacements of the geometric centre of the upper vertebral body with respect to the lower vertebra at each measuring point (Fig. 2).

Treatment of data

Neutral zone. The NZ is that zone within the range of motion in which the spine can be displaced with the application of a very small force or moment [28]. In our study, the NZ is the interval in degrees for which the couple is zero after the first and second unloadings (Fig. 3).

Neutral position. The normal neutral position of the neck is not well known. The initial position of the specimen at the beginning of experimentation is very important for its mechanical response and for explaining the results. We loaded FSUs from any spontaneous starting position. The neutral position of a FSU is defined as the centre of the NZ. The displacement curves thus obtained were recentred around the midpoint of the NZ. These recentred curves were then utilised to calculate other parameters.

Secant stiffness. The secant stiffness is the slope of the straight line calculated by a simple linear regression of the loading part of a load-displacement curve in the elastic zone between the first point after the greatest variation in the curve gradient and the point with maximum loading (Fig. 3).

Limit moment (LM). The important NZ can also be represented by another parameter: the limit moment. This is equal to the projection of the load-displacement curve between the origin and the limit of the NZ along the ordinate (Fig. 3). In other words, this LM is enough to turn a FSU to the same displacement as the NZ. The LM is the minimal resistance offered by discoligamentous structures.

Measuring of displacements between C1 and C2

The five additional segments were taken between the atlas and the axis. All ligaments between the occiput and the atlas were sectioned. Two threaded rods were screwed to the lateral mass of the atlas to apply the loads. The same measuring procedure as that for the middle-to-lower cervical specimen was performed.

Results

The three-dimensional motions under three respective moments were measured for 56 FSUs of the middle-to-lower cervical spine. The mean disc state is 2.9, using Nachemson's classification. Figure 2 shows the typical load-displacement curves in three dimensions of any FSU of the cervical spine. The NZ is a representative parameter, because it correlates well with the range of motion (Fig. 4). The ratios of NZ to ROM (NZR) were calculated in order to indicate the large amplitude of the NZ.

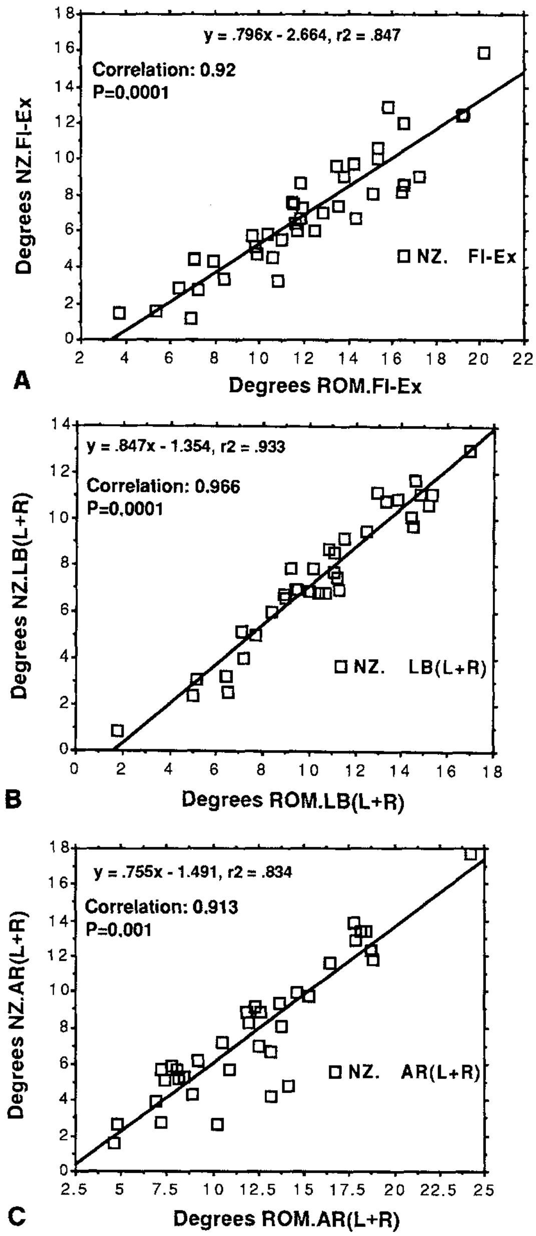

Fig. 4A–C. Correlation between neutral zone (NZ) and range of motion (ROM). **A** Flexion-extension, **B** lateral bending, **C** axial torsion

Table 2. Neutral zone and range of motion (degrees)

Levels	+RZ−(−RZ)		+RX−(−RX)		+RY−(−RY)	
	NZ (FE)	ROM (FE)	NZ (LB)	ROM (LB)	NZ (AT)	ROM (AT)
C6–7	6.6 (4.9)	11.6 (5.0)	6.7 (2.8)	9.8 (3.7)	6.2 (3.8)	9.8 (4.3)
C5–6	7.6 (3.1)	11.9 (4.4)	7.0 (2.6)	10.0 (3.3)	6.6 (2.8)	11.0 (3.6)
C4–5	7.6 (2.8)	13.3 (3.5)	7.5 (3.9)	10.5 (4.1)	9.6 (4.4)	15.5 (4.7)
C3–4	6.7 (3.2)	12.0 (4.0)	7.8 (3.2)	10.8 (3.4)	8.4 (3.5)	12.2 (4.4)
C2–3	6.8 (2.7)	11.1 (3.0)	9.0 (1.7)	11.6 (2.2)	6.7 (2.9)	11.1 (2.9)
C1–2	14.7 (9.7)	23.8 (4.9)	6.4 (3.2)	8.3 (2.2)	53.8 (21.0)	75.9 (8.3)

NZ, Neutral zone; ROM, range of motion; +RZ, extension; −RZ, flexion; FE, flexion + extension; −RX, left lateral bending; +RX, right lateral bending; LB, left + right lateral bending; +RY, left axial torsion; −RY, right axial torsion; AT, left + right axial torsion
Each NZ and ROM is the mean of values from all functional spinal units
Standard deviation in parentheses

Table 3. Neutral zone ratio and limit moment (Nm)

Levels	NZR (FE)	NZR (LB)	NZR (AT)	LM (FE)	LM (LB)	LM (AT)
C6–7	0.51 (0.20)	0.66 (0.09)	0.59 (0.13)	0.59 (0.18)	0.89 (0.15)	1.27 (0.39)
C5–6	0.62 (0.11)	0.69 (0.08)	0.60 (0.15)	0.63 (0.17)	0.93 (0.25)	1.42 (0.36)
C4–5	0.56 (0.08)	0.66 (0.14)	0.60 (0.15)	0.51 (0.12)	0.93 (0.11)	1.19 (0.54)
C3–4	0.50 (0.11)	0.70 (0.14)	0.68 (0.09)	0.52 (0.10)	0.85 (0.21)	1.20 (0.27)
C2–3	0.56 (0.10)	0.78 (0.06)	0.63 (0.21)	0.55 (0.10)	0.99 (0.12)	1.20 (0.11)
C1–2	0.62 (0.44)	0.67 (0.27)	0.70 (0.23)	–	–	–

NZR, Neutral zone ratio; LM, limit moment
In the LM analysis, spines 1–9 were excluded. The LM is the mean of values from positive and negative directions

Table 4. Secant stiffness (Nm/degree)

Levels	−RZ	+RZ	−RX	+RX	−RY	+RY
C6–7	0.474 (0.27)	0.452 (0.21)	0.919 (0.50)	0.765 (0.41)	1.125 (0.60)	1.273 (0.80)
C5–6	0.451 (0.25)	0.489 (0.23)	0.802 (0.20)	0.709 (0.20)	0.992 (0.26)	1.033 (0.16)
C4–5	0.338 (0.14)	0.446 (0.20)	0.760 (0.23)	0.706 (0.27)	0.935 (0.38)	0.933 (0.34)
C3–4	0.265 (0.10)	0.472 (0.23)	0.659 (0.43)	0.728 (0.30)	1.008 (0.31)	0.941 (0.41)
C2–3	0.401 (0.07)	0.415 (0.10)	0.728 (0.13)	0.632 (0.17)	0.758 (0.17)	0.815 (0.32)
C1–2	0.541 (0.11)	0.906 (0.41)	1.172 (0.65)	2.918 (3.90)	0.587 (0.35)	0.514 (0.08)

For the same purpose we first calculated the LM (Fig. 3). The NZ and the ROM are shown in Table 2, the NZR and LM in Table 3, and the secant stiffness in Table 4.

The kinematic type of motion followed the direction of the loading moments. The main motion was always in the direction of loading. The coupled motion was often small. However, lateral bending (LB) was coupled with considerable axial rotation (AR) and vice versa. The mean RCM (ratio of coupled motion) for AR/LB was 0.32 when the specimen was lateroflexed while the mean RCM for LB/AR was 0.49 when the specimen was rotated (Fig. 5). The lateral bending provoked by axial torsion was greater than the axial rotation provoked by lateral bending. These two RCMs, in frontal and horizontal planes, are significantly different at different spinal levels.

In spite of the variation, all FSUs have similar types of load-displacement curve. A curve averaged from serveral FSUs retains this behaviour (Fig. 6). Our experimental results show that the load-displacement curve of principal motion is non-linear in the normal state, at all spinal levels. However, this curve becomes quasi-linear when the load goes beyond an LM. We also find that the main displacement with the greatest variation in the curve gradient coincides approximately with the limit of the NZ (Fig. 3). A principal load-displacement curve also appears to obey a power law viscoelasticity relationship given by the following equation:

$$Y = aX - b + bk^{(aX/b)}$$

where X is the angular displacement in degrees, Y is the moment in Nm, a is the slope or secant stiffness in Nm * degree^{-1}, b is the intersection of the regression line at the ordinate (Nm), and k is the coefficient.

The maximum displacement did not exactly correspond to the point of maximum loading, but was delayed by approximately 15 s. This phenomenon may be explained by creep.

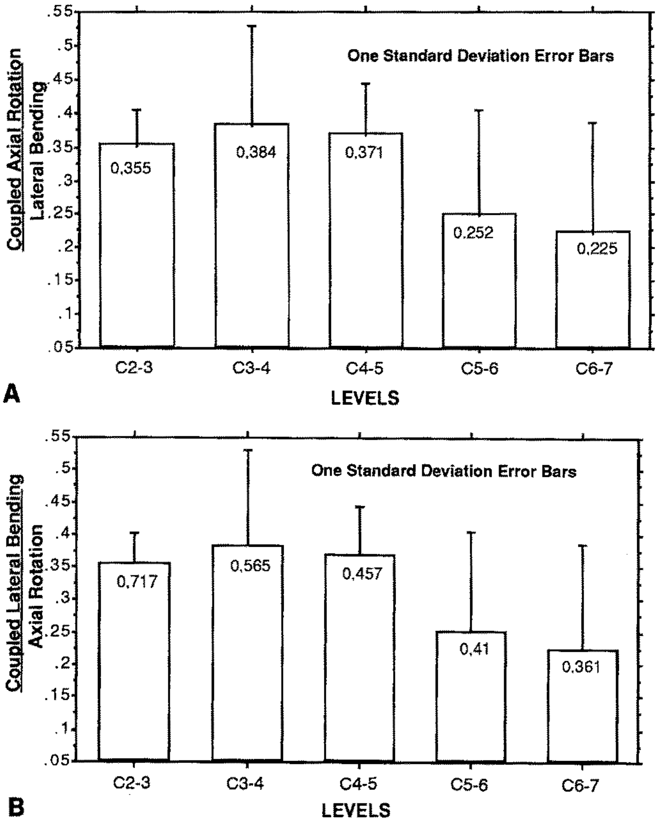

Fig. 5A, B. Important coupled motion. **A** Coupled axial rotation under lateral bending, **B** coupled lateral bending under axial torsion

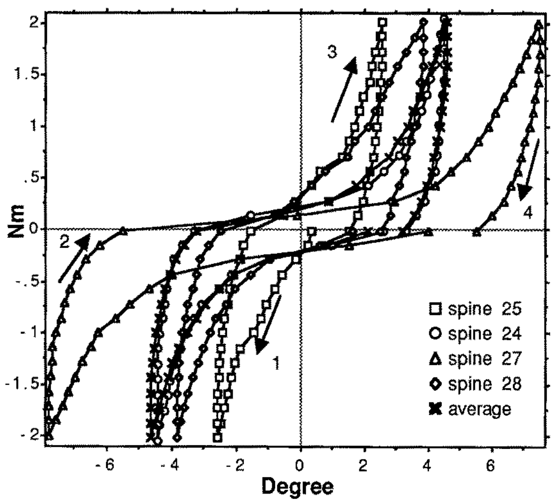

Fig. 6. Distribution of results among different specimens. Principal displacements under lateral bending in four FSUs (C4–5)

Through the analysis of variance (ANOVA) with one repetitive factor (Scheffe *F*-test), we found that the stiffness in the sagittal plane (RZ) was significantly inferior to the stiffness in other planes and the stiffness under axial torsion in the horizontal plane (RY) was superior to the stiffness in other planes. However, there were no significant differences in stiffness between two directions in the same plane (+R, −R).

Through ANOVA with two factors, level of FSU and degenerative state of intervertebral disc, we found that the stiffness at different levels varies under lateral bending, and that the stiffness of different disc states varies under right lateral bending. No other significant differences were found in the results obtained (NZ, ROM and secant stiffness) in regard to these two factors.

We analysed the influences of initial cycle and of duration of loading on the variation of results in 8 FSUs (spines 10–13). We found that the global load-displacement curve, NZ, ROM, LM, RCM and secant stiffness stabilised after the third loading cycle. With our experimental protocol, the displacement remained constant after 15 s loading in any mode and with any magnitude of loading.

Discussion

Range of motion

Many studies related to ROM have been carried out in vitro or in vivo by different authors. Tables 5–7 show a comparison of some representative results. Although differences were found between different authors, our results are within the range of these differences. Our results confirm Lysell's work in flexion-extension and lateral bending [17], and the results of Penning and Wilminck [31] and Dvorak et al. [4] in axial torsion. We found the ROM in vitro under flexion-extension to be inferior to that in vivo between C3 and C7. Several of the authors measured single-sided rotation, but this may not be exactly half of the total range because of the arbitrary position of the point of departure within a large NZ.

Neutral zone

Unlike the case for ROM, few reports on the NZ can be found. Our values for the NZ are similar to the results of Panjabi et al. [28], with only 6%–25% of difference in the middle-to-lower cervical spine. The NZ of the cervical spine is very great compared to the NZ of the lumbar spine as reported by Yamamoto et al. [37]. The mean values of the NZR in both cervical FSU and lumbar FSU are given in Table 8. These evident differences in three dimensions substantiate the idea that the NZ is a particular mechanical property of the cervical spine. In our earlier studies we separated the application of positive and negative moments, for example extension and flexion, into two isolated experiments, but we could not find a stable initial position for both flexion and extension experimentation. In any cervical FSU, the initial position changes under a very small force, such as the weight of the upper fixation device. We think that no one neutral position exists, but that there is a neutral zone in the cervical spine. These NZs were widened in the experimental specimens after loading deformation. The small force required to change the initial position was later quantified as between zero Nm and limit moment (LM).

Coupled motions

Table 9 shows the findings of different authors on the ratio of coupled motion in frontal and horizontal planes.

Table 5. ROM (degrees) under flexion-extension in some other studies

Authors	Year	Methods	C1–2	C2–3	C3–4	C4–5	C5–6	C6–7
Wen et al. (present study)	1992	Vitro, FSU	23.8	11.1	12.0	13.3	11.9	11.6
Schulte et al. [33]	1989	Vitro, segment	–	–	–	6.5	4.5	4.5
Panjabi et al. [29]	1988	Vitro, segment	22.4	–	–	–	–	–
Goel et al. [11]	1988	Vitro, 0.3 Nm, selspots	–	–	6.4	6.2	5.2	5.7
Goel et al. [10]	1984	Vitro, 0.3 Nm, sonic digitizer	–	–	–	7.3	10.1	–
Moroney et al. [22]	1984	Vitro, FSU	–	4.3	6.9	6.4	–	7.9
White and Panjabi [36]	1978	Vitro	10.0	8.0	13.0	12.0	17.0	16.0
Lysell [17]	1969	Vitro, radiography	–	4.9	10.2	13.0	14.5	13.5
Ball and Meijers [1]	1964	Vitro	–	9.5	15.1	18.1	20.2	17.6
Dvorak et al. [4]	1988	Vivo, active	12.0	10.0	15.0	19.0	20.0	19.0
Dvorak et al. [4]	1988	Vivo, passive	15.0	12.0	17.0	21.0	23.0	21.0
Gonnot et al. [12]	1985	Vivo, radiography	5.5	4.0	6.0	8.0	9.0	8.0
Penning [30]	1978	Vivo, active	30.0	12.0	18.0	20.0	20.0	15.0
Mestdagh [19, 20]	1969	Vivo, vitro, radiography	–	11.0	14.5	18.0	19.5	16.0
Fielding [7]	1957	Cineroentgenography	15.0	–	–	–	–	–

Table 6. ROM under lateral bending in some other studies

Authors	Year	Methods	C1–2	C2–3	C3–4	C4–5	C5–6	C6–7
Wen et al. (present study)	1992	Vitro, FSU	8.3	11.6	10.8	10.5	10.0	9.8
Schulte et al. [33]	1989	Vitro, segment	–	–	–	6.0	2.4	2.8
Panjabi et al. [29]	1988	Vitro, whole spine	13.4	–	–	–	–	–
Goel et al. [11]	1988	Vitro, 0.3 Nm, selspots	–	–	6.4	7.2	4.8	3.6
Goel et al. [10]	1984	Vitro, 0.3 Nm, sonic digitizer	–	–	–	5.4	4.6	–
Moroney et al. [22]	1984	Vitro, FSU	–	6.8	3.2	3.6	–	5.6
White and Panjabi [36]	1978	Vitro	0.0	20.0	22.0	22.0	16.0	14.0
Lysell [17]	1969	Vitro, radiography	–	7.9	9.8	9.1	9.0	8.4
Mestdagh [19, 20]	1969	Vivo, vitro, radiography	–	7.0	4.0	4.5	5.5	4.5
Penning [30]	1978	Vivo, active	–	14.0	14.0	14.0	14.0	14.0

Table 7. ROM under axial torsion in some other studies

Authors	Year	Methods	C1–2	C2–3	C3–4	C4–5	C5–6	C6–7
Wen et al. (present study)	1992	Vitro, FSU	75.9	11.1	12.2	15.5	11.0	9.8
Schulte et al. [33]	1989	Vitro, segment	–	–	–	5.0	2.6	2.2
Panjabi et al. [29]	1988	Vitro, whole spine	77.8	–	–	–	–	–
Goel et al. [11]	1988	Vitro, 0.3 Nm, selspots	–	–	4.6	4.6	4.6	3.4
Goel et al. [10]	1984	Vitro, 0.3 Nm, sonic digitizer	–	–	–	3.6	2.8	–
Moroney et al. [22]	1984	Vitro, FSU	–	2.4	2.2	1.2	–	2.0
White and Panjabi [36]	1978	Vitro	94.0	18.0	22.0	12.0	20.0	18.0
Lysell [17]	1969	Vitro, radiography	–	6.0	9.8	10.3	8.0	5.7
Mestdagh [20]	1969	Vivo, vitro, radiography	–	8.0	6.0	3.5	2.5	–
Mimura et al. [21]	1989	Vivo, radiography	75.2	7.4	5.8	4.2	5.4	6.4
Penning and Wilminck [31]	1987	Vivo	81.0	6.0	13.0	13.6	13.8	10.8
Dvorak et al. [4]	1987	Vivo, passive	83.0	6.0	13.0	13.4	14.0	10.8
Penning [30]	1978	Vivo, active	70.0	14.0	14.0	14.0	14.0	14.0
Mestdagh [19]	1976	Vivo, radiography	–	23.5	17.5	12.0	5.5	–
Fielding [7]	1957	Cineroentgenography	90.0	–	–	–	–	–

Our values fell within the range of the findings of Moroney et al. [23] and Lysell [17].

Non-linearity and stiffness

The viscoelastic behaviour of a cervical human spine has been demonstrated by many authors. The stiffness coefficient is different at different load magnitudes. The load-displacement curves in our experiments were similar to those of Shea et al. [34]. They and Coffee et al. [2] found that the sagittal load-displacement curves for the cervical segment were non-linear for even small loads, but became approximately linear at higher loads. We found this phenomenon in three planes. Their "non-linear part" corresponds to our NZ and their "linear part" at higher loads to our elastic zone (EZ), in which we calculated the secant stiffness by linear regression (Fig. 3). These two parameters, NZ and EZ, have been described [29, 37]. In fact, the mechanical response showed a power law load-displacement curve or biphasic behaviour, more like that of ligaments as described by Myklebust et al. [24] and Fung et al. [9]. This particular phenomenon represents a cartilaginous articulation characteristic in the absence of neck muscle, because the beginnings of the curve or NZ show a large displacement at small loads. As the loads increase, so does the resistance at an increasing rate, but the end of the curve is stable when the articulation is restrained by stress ligaments.

We wanted to compare our secant stiffness to the stiffness found by other authors (Table 10), but the main difference remained the choice of calculation methods. Panjabi et al. [28] took the secant flexibility from the slope of straight lines between the first and the third measurement point. Moroney et al. [23] took average

Table 8. Comparison of NZR, NZ and ROM in cervical and lumbar FSU (NZ and ROM in degrees)

	Cervical FSU (C2–7) (present study) NZR (NZ/ROM)	Lumbar FSU (L1–S1) [37] NZR (NZ/ROM)
Flexion-extension	0.59 (7.06/11.98)	0.27 (3.52/12.92)
Bil. lateral bending	0.72 (7.60/10.54)	0.28 (3.28/11.56)
Bil. axial torsion	0.63 (7.50/11.92)	0.29 (1.28/4.44)

Table 9. Ratio of coupled motion

Authors	C2–3	C3–4	C4–5	C5–6	C6–7	C2–7
AR/LB under lateral bending						
Wen et al. (present study)	0.35	0.38	0.37	0.25	0.23	0.32
Goel et al. [11]	–	0.53	0.39	0.50	0.39	–
Moroney et al. [23]	–	–	–	–	–	0.32
Lysell [17]	0.68	0.54	0.57	0.38	0.37	–
LB/AR under axial torsion						
Wen et al. (present study)	0.72	0.56	0.46	0.41	0.36	0.49
Mimura et al. [21]	0.22	1.07	1.48	0.74	0.42	–
Moroney et al. [23]	–	–	–	–	–	0.51
Goel et al. [11]	–	1.00	0.74	0.65	0.41	–
Penning and Wilminck [31]	–	–	–	–	–	0.74
Lysell [17]	0.88	0.63	0.48	0.46	0.52	–

AR, Axial rotation; LB, lateral bending

Table 10. Stiffness of the FSU of middle-to-lower cervical spine as found by different authors (Nm/degree)

Authors	Year	Methods	−RZ	+RZ	−RX	+RX	−RY	+RY
Wen et al. (present study)	1992	C2–C7, FSU	0.39	0.45	0.78	0.71	0.97	1.00
Shea et al. [34]	1991	5 Nm	2.26	3.76	–	–	–	–
		C2–C5[a]	2.88	4.58	–	–	–	–
		C5–T1[a]	1.66	2.38	–	–	–	–
		C2–T1[a]	2.26	3.48	–	–	–	–
Moroney et al. [23]	1988	FSU, 1.8 Nm	0.43	0.73	0.68	0.68	1.16	1.16
Coffee et al. [34[b]]	1988	Mid	2.88	4.58	–	–	–	–
		Low	1.66	3.74	–	–	–	–
Raynor et al. [32]	1987	C3–4, C6–7	0.21	0.31	0.25	0.32	0.34	0.32
Liu [34[b]]	1982	Mid, FSU	0.51	1.24	–	–	–	–
		Low, FSU	2.95	2.58	–	–	–	–

[a] Flexion 5 Nm, extension 3.5 Nm
[b] Reference noted by Shea [34]

The stiffness values of Shea and Coffee et al. were measured across three-vertebrae segments, so the values have be doubled for direct comparison. Raynor et al. gave the matrices of flexibility. The stiffness values were the simple inverses of the flexibility stiffness from the slope of the straight line calculated by linear regression of all displacement points between the origin and 1.8 Nm, and the tangential stiffness. Shea et al. [34] took the tangential stiffness from the maximum loading point. Theoretically speaking, the stiffness obtained from our calculations is greater than the average stiffness used by Moroney et al. and is less than the tangential stiffness. Shea et al. found that the mid-cervical region was stiffer in extension than the low-cervical region. However, there were no significant differences in secant stiffness between middle and lower cervical FSUs in our experiments. The bending stiffness of the cervical spine was significantly influenced by the direction of the bending moment and the previous deformation history. In living soft tissue the flexibility is not, however, the inverse of the rigidity, because of the viscosity and the NZ. In our case, the NZ was excluded from the stiffness calculation.

Degenerative disc state

Ball and Meijers [1] found that disc degeneration was associated with progressive restriction of movement at the affected level. However, Lysell [17] did not find a correlation between reducing range of motion and increasing degeneration of cervical vertebrae. Moroney et al. [23] found no statistically significant differences of the stiffness between degenerated cervical discs and normal discs under bending or axial torsions movement. Panjabi and Goel [26] quantified the relationship between chronic instability and disc degeneration by measuring the NZ of lumbar FSUs [26]. No study has been made of this NZ in the cervical spine. In contrast to Panjabi et al., we did not find a significant dependence of NZ on the degree of disc degeneration for any mode of loading at any level in the cervical region.

The classification of disc degeneration by Nachemson [25] analyses only the intervertebral disc. It does not permit analysis of arthrosis of articular facets. Johnson et al. [14] utilised another classification (four stages) of disc degeneration which is similar to that of Nachemson. Ehni recommended a classification of cervical spine degeneration which took account of the disc as well as the articular facet [5]. However, all these methods are subjective analyses and do not take into account the inherent variability in the classification.

Variation of results

A review of available published data revealed wide variation. This was probably due to biological variations between specimens (Fig. 6), but also to differences in experimental conditions and design. The four secondary variables determined by Tencer and Ahmed [35] (load axis position, nucleus pressure, number of initial cycles and load duration) were shown to significantly affect the displacement between lumbar vertebrae for various types of loading. To obtain some rotation, Panjabi et al. [28] applied a shear force to the upper vertebra of one FSU. McElhaney et al. [18] exerted an eccentric moment on the upper vertebra of a long cervical segment. Like Moroney et al. [23], we utilised pure moments to avoid combined forces.

The increment in moment and the maximum load applied were different in different authors' work. These factors play an important role in the biomechanical properties of the spine. Panjabi et al. [28], Moroney et al. [23] and Goel et al. [10, 11] utilised the same fixed increment and final load for all specimens. The maximal load was 2.2 Nm for Moroney et al., 50 N for Panjabi et al., 3.4 Nm for Raynor et al. [32], 3.5–5 Nm for Shea et al., and 0.3 Nm for Goel et al. Panjabi and Moroney et al. applied 49–50 N preload; no preload report was indicated by the others. Janevic et al. [13] found that a large compressive preload decreased motion segment flexibility [13]. We did not prescribe the maximum load for all spinal segments, but determined this parameter from the instantaneous results of principal displacement during the experimentation. We believe that this method fits in better with the individual variation than fixed final loading, and does not damage the structures for later tests. We found that the heavier the load which we applied, the more rigid the FSU became, but there was no significant difference. The loading cycle and the order of loading direction were also important. McElhaney et al. [18] investigated viscoelastic responses of the human cervical spine in depth. They indicated that the mechanically stabilised state was achieved after about 30 cycles. We found, in a separate experiment, that 90% of this stability was reached after three load-unload cycles.

Panjabi et al. [27] indicated that freezing temperatures have been shown not to affect the biomechanical properties of bone and spinal specimens. Keller et al. [15] demonstrated that the biomechanical response of lumbar segments was significantly altered by the death of the animal. Flynn et al. [8] further indicated that deep-freezing was superior to freeze-drying for spinal preservation. We wondered whether our radio-sterilisation affected the biomechanical behaviour of the spinal specimens.

There are, as yet, no reports on the stiffness of the human cervical spine in in vivo testing. The interpretation of in vitro results is restricted because the paraspinous and trunk muscles probably produce additional stiffness, stabilising the cervical spine.

In future experimentation these cadaveric spine conditions, change after death, differences in experimental design and calculation, viscosity of spine, and the role of muscles in vivo, should be determined with sufficient specimens to perform a statistical analysis.

References

1. Ball J, Meijers KAE (1964) On cervical mobility. Ann Rheum Dis 23:429–438
2. Coffee MS, Edwards WT, Hayes WC, White III AA (1988) Biomechanical properties and strength of the human cervical spine. Orthop Trans 12:476
3. Dvorak J, Hayer J, Zehnder R (1987) CT-functional diagnostics of the rotatory instability of the upper cervical spine. 2. An evaluation on healthy adults and patients with suspected instability. Spine 12:726–731

4. Dvorak J, Froehlich D, Penning L, Baumgartner H, Panjabi MM (1988) Functional radiographic diagnosis of the cervical spine: flexion/extension. Spine 13:748–755
5. Ehni G (1984) Cervical arthrosis, diseases of cervical motion segments. Year Book, Chicago, pp 44–47
6. Fielding JW (1957) Cineroentgenography of the normal cervical spine. J Bone Joint Surg [Am] 39:1280–1288
7. Fielding JW (1985) Cervical spine surgery. Past, present, and future potential. Clin Orthop 200:284–290
8. Flynn J, Rudert MJ, Olson E, Baratz M, Hanley E (1990) The effects of freezing or freeze-drying on the biomechanical properties of the canine intervertebral disc. Spine 15:567–580
9. Fung YC, Perrone N, Anliker M (1972) Properties of tendon and skin. Biomechanics. Its foundations and objectives. Prentice-Hall, New Jersey, pp 141–179
10. Goel VK, Clark CR, McGowan D, Goyal S (1984) An in-vitro study of the kinematics of the normal, injured and stabilized cervical spine. J Biomech 17:363–376
11. Goel VK, Clark CR, Harris KG, Schulte KR (1988) Kinematics of the cervical spine: effects of multiple total laminectomy and facet wiring. J Orthop Res 6:611–619
12. Gonnot GP, Mestdagh H, Deschamps G, Dimnet J, Fischer LP (1985) Etude cinématique de la colonne cervicale dans le mouvement de flexion-extension. Bull Assoc Anat (Nancy) 67:433–443
13. Janevic J, Miller JAA, Schultz AB (1991) Large compressive preloads decrease lumbar motion segment flexibility. J Orthop Res 9:228–236
14. Johnson RM, Crelin ES, White III AA, Panjabi MM, Southwick (1975) Some new observations on the functional anatomy of the lower cervical spine. Clin Orthop 111:192–200
15. Keller TS, Holm SH, Hansson TH, Spengler DM (1990) The dependence of intervertebral disc mechanical properties on physiologic conditions. Spine 15:751–761
16. Lavaste F, Asselineau A, Diop A, Grandjean JL, Laurain JM, Skalli W, Roy-Camille R (1990) Experimental procedure for mechanical evaluation of dorso-lumbar segments and osteosynthesis devices. Rachis 2:435–446
17. Lysell E (1969) Motion in the cervical spine. Acta Orthop Scand [Suppl] 123:1–61
18. McElhaney JH, Doherty BJ, Paver JG, Myers BS, Grey L (1990) Combined bending and axial loading responses of the human cervical spine. Biomed Eng Dept, Duke University, Durham, NC 27706, USA
19. Mestdagh H (1969) Anatomie fonctionelle du rachis cervical inférieur. Medical thesis, Lille
20. Mestdagh H (1976) Morphological aspects and biomechanical properties of the vertebroaxial joint (C2–C3). Acta Morphol Neerl-Scand 14:19–30
21. Mimura M, Moriya H, Watanabe T, Takahashi K, Yamagata M, Tamaki T (1989) Three-dimensional motion analysis of the cervical spine with special reference to the axial rotation. Spine 14:1135–1139
22. Moroney SP (1984) Mechanical properties and muscle force analyses of the lower cervical spine. Thesis, Chicago
23. Moroney SP, Schultz AB, Miller JAA, Andersson GBJ (1988) Load-displacement properties of lower cervical spine motion segments. J Biomech 21:769–779
24. Myklebust JB, Pintar F, Yoganandan N, Cusick JF, Maiman D, Myers T, Sances A (1988) Tensile strength of spinal ligaments. Spine 13:526–531
25. Nachemson A (1960) Lumbar intradiscal pressure. Acta Orthop Scand [Suppl] 43:9–97
26. Panjabi M, Goel V (1983) Relationship between chronic instability and disc degeneration. Orthop Trans 7:96
27. Panjabi MM, Krag M, Summers D, Videman T (1985) Biomechanical time-tolerance of fresh cadaveric human spine specimens. J Orthop Res 3:292–300
28. Panjabi MM, Summers DJ, Pelker RR, Videman T, Friedlaender GE, Southwick WO (1986) Three-dimensional load-displacement curves due to forces on the cervical spine. J Orthop Res 4:152–161
29. Panjabi MM, Dvorak J, Duranceau J, Yamamoto I, Gerber M, Rauschning W, Bueff HU (1988) Three-dimensional movements of the upper cervical spine. Spine 13:726–730
30. Penning L (1978) Normal movements of the cervical spine. Am J Roentgenol 130:317–326
31. Penning L, Wilmink JT (1987) Rotation of the cervical spine. A CT study in normal subjects. Spine 12:732–738
32. Raynor RB, Moskovich R, Zidel P, Pugh J (1987) Alterations in primary and coupled neck motions after facetectomy. Neurosurgery 21:681–687
33. Schulte K, Clark CR, Goel VK (1989) Kinematics of the cervical spine following discectomy and stabilization. Spine 14:1116–1121
34. Shea M, Edwards, White AA, Hayes WC (1991) Variations of stiffness and strength along the human cervical spine. J Biomech 24:95–107
35. Tencer AF, Ahmed AM (1981) The role of secondary variables in the measurement of the mechanical properties of the lumbar intervertebral joint. J Biomech Eng 103:103–129
36. White III AA, Panjabi MM (1990) Clinical biomechanics of the spine. Lippincott, Philadelphia, pp 110, 667
37. Yamamoto I, Panjabi MM, Cristo T, Oxland T (1989) Three-dimensional movements of the whole lumbar spine and lumbosacral joint. Spine 14:1256–1260