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Three-dimensional kinematics of the lumbar spine during treadmill walking at different speeds

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Abstract The lumbar spine is of primary importance in gait and its development is influenced by the upright posture adopted in human locomotion. However, little is known about the kinematic behavior of the lumbar spine during walking. The aim of this study was to examine (1) lumbar spine kinematics during walking, (2) the effect of walking velocity on lumbar motion patterns and (3) the coupling characteristics of rotation and bending. In 22 volunteers aged 15–57 years, the three-dimensional displacements of T12 to the sacrum were sampled during elementary movements of the trunk and during walking on a treadmill at four walking velocities. A three-dimensional electrogoniometer (CA 6000 Spine Motion Analyzer) sampling at 100 Hz was used. We analyzed maximal primary and coupled motion

ranges (ROM) and velocities in each plane. Lumbar ROM during walking did not exceed 40% of maximal active ROM. Transverse plane ROM and frontal and transverse velocities increased with walking velocity. Coupling of rotation and bending during walking was individually variable and dependent on walking velocity. Moreover, the smoothness of the bending-rotation path varied with walking velocity. A simplified envelope of lumbar coupling characteristics during walking is presented, and the existence of an individually variable walking speed that is characterized by a more harmonic lumbar contribution is hypothesized.

Key words Lumbar spine · Kinematics · Gait · Three-dimensional · Electrogoniometry · CA 6000

Introduction

Walking is for humans one of the most natural activities. However, although it is well known that the lumbar spine is of primary importance in gait [11] and that its development is influenced by the upright posture adopted in human locomotion [21], little is known about the kinematic behavior of the lumbar spine during this daily activity.

Several biomechanical aspects of the lumbar spine during walking have been studied previously. For instance, the resultant forces on lumbar discs and facet joints during walking have been shown to reach 2.5 times the body

weight [4], although lower and higher peak loads have been described [7, 9, 13, 14]. A recent *in vivo* study by Wilke et al. [32], for example, showed much lower intradiscal pressures during gait, which were only moderately larger than those registered during relaxed standing or sitting. It is, however, not known whether these pressures are altered in patients with low back disorders. Lumbar loads during gait have been shown to increase with increasing walking velocity [7], although this finding remains controversial [32]. Peak erector spinae contraction forces during walking (140 N) are larger than those of the remaining trunk muscles (about 15 N) [9]. A study by Vink and Karssemeijer [28] showed bilateral activity of

intrinsic lumbar back muscles during double support. After heel strike, the homolateral muscles displayed a larger activity. This study revealed no clear relationships between pelvic sagittal and frontal rotations and muscle activation. The bilateral activity of the erector spinae was shown to occur at two points in the cycle [5]. Its total duration was 15–26% of the stride cycle. Thorstensson et al. [25] demonstrated that sagittal and frontal movements of the trunk have different paths and vary with walking velocity. The bilateral co-contraction of multifidus and longissimus just before heel strike is posited to have the effect of restricting trunk movements in the frontal plane during gait.

Clinical applications in patients suffering from low back pain showed an increase of lumbar muscle activity during the swing phase [1].

Vogt et al. [29] analyzed the kinematics and electromyographic (EMG) activation patterns of several muscles during walking. This study showed, during slope walking (10%), a significant increase in lumbar frontal and transverse motion ranges during walking as compared to level walking. Syczewska et al. [24] showed that sagittal angular changes of various spine segments during walking displayed quite similar patterns in healthy volunteers, whereas frontal plane angular changes varied from segment to segment. In the lumbar spine, bending occurred towards the swinging leg. Whittle and Levine [31] obtained average lumbar motion ranges during gait at free velocity of 4.0° ,

7.6° and 8.3° in the sagittal, frontal and transverse plane respectively. Lumbar lordosis displayed large individual variations [30,31]. Lumbar lateral bending occurred towards the supporting leg; its range was equal to the pelvic tilt. Lumbar axial rotation was slightly less than pelvic rotation, and a phase lag was observed between pelvic and lumbar rotation.

The aims of the present study were to determine the three-dimensional patterns of lumbar kinematics during gait and the influence of increasing walking velocity on these parameters. In particular, the coupling between axial rotation and lateral bending was analyzed. Although still a matter of controversy, coupling of bending and rotation during elementary trunk motion has been studied by several authors [3, 8, 16, 17, 18,22]. However, little is known about this aspect during gait, despite the fact that such information might be useful in fundamental and clinical conditions for several disorders or deformities, such as idiopathic scoliosis and chronic low back pain. Three-dimensional motion of T12 to the sacrum was sampled using a three-dimensional electrogoniometer in 22 subjects walking on a treadmill at different walking speeds (Fig. 1).

Materials and methods

Twenty-two asymptomatic volunteers participated in this study. Their mean age was 34 years (range: 15–57 years). Nine were female and 13 male.

The height, weight, trunk height (measured in seated position), and length of the lower extremities (anterior superior iliac spine to medial malleolus distance) were registered for each subject.

Three-dimensional motion between the sacrum and T12 was tracked using a six-degree-of-freedom instrumented spatial linkage (CA 6000 Spine Motion Analyzer, OSI, USA). The reliability and reproducibility of this instrument was validated in a previous study [10]. The linkage was mounted on the subject using a pelvic and a thoracic harness (Fig. 1). The sampling rate was set to 100 Hz. Global lumbar spine motion was sampled during elementary trunk movements (flexion-extension, lateral bending and axial rotation). These were carried out actively at a free pace in standing position. From neutral position, the subject was instructed to first reach maximal flexion or right bending or rotation and then continuously move to the maximal opposite motion before regaining the original neutral position. Each of these movements was repeated four times. Lumbar kinematics was then sampled during walking on a motorized treadmill (Tunturi 502 Electronic Jogging Machine, Tunturi, Finland) at four walking velocities (0.8, 1.1, 1.4 and 1.7 m/s). The subject was allowed to practice on the treadmill for 5–10 min. Eight to 12 gait cycles were sampled at each walking velocity.

The parameters considered were

1. Average motion ranges (ROM) in each plane (averaged over all cycles)
2. Coupling of lateral bending and rotation
3. Peak motion velocities in each plane

The statistical analysis aimed at evaluating

1. The effect of increasing walking velocity on lumbar kinematics
2. The relation of lumbar kinematics during gait to kinematics during elementary movements and to anthropometric parameters

Fig. 1 Electrogoniometer mounted on a subject using thoracic and pelvic harnesses

Table 1 Average (SD) lumbar motion ranges and maximal velocities during walking at different velocities and during elementary trunk movements

Walking velocity	Sagittal		Frontal		Transverse	
	ROM (°)	Velocity (°/s)	ROM (°)	Velocity (°/s)	ROM (°)	Velocity (°/s)
0.8 m/s	5 (1)	55 (15)	9 (2)	51 (11)	9 (3)	46 (13)
1.1 m/s	5 (1)	61 (19)	10 (2)	64 (15)*	11 (4)*	60 (16)*
1.4 m/s	6 (2)	71 (20)	11 (3)	71 (18)*	13 (5)*	76 (23)*
1.7 m/s	7 (2)	80 (22)*	12 (3)*	87 (23)*	16 (5)*	95 (29)*
Elementary	96 (17)	79 (24)	69 (13)	23 (10)	43 (9)	25 (12)

* $P < 8.93 \times 10^{-4}$; significant increase as compared to the next slowest velocity (Student's *t*-test for paired samples)

The tests used to compare parameters at different walking velocities were Student's *t*-test for independent samples (after verification of normal distribution using Kolmogorov-Smirnov's test). Bonferroni's correction for multiple comparisons was applied [2], for which the significance level α (classically 0.05) is divided by the number of means implied in the comparisons. In the present analysis, this correction yielded a level below which a difference could be regarded as significant of $P < 8.93 \times 10^{-4}$. This value will be regarded as the significance threshold. Pearson's linear correlation coefficients were used to analyze the relationships between variables sampled during elementary motion and during walking. The effect of walking velocity on lumbar motion parameters was studied using ANOVA.

Results

No significant differences were found either between female ($N=9$) and male ($N=13$) subjects, or between subjects aged 30 years or less ($N=11$) and those aged 40 years or more ($N=11$). Thus only results concerning the entire population will be presented.

Average sagittal ROM (Table 1) during walking was 6% (SD 2%) of maximal active flexion-extension ROM. A trend to an increase with increasing walking velocity occurred, starting at the speed of 1.1 m/s ($P=1 \times 10^{-2}$). Frontal plane ROM during walking tended to increase with walking velocity, although this was not significant after Bonferroni's correction ($3 \times 10^{-2} > P > 3 \times 10^{-3}$). Its range amounted to 13–18% (SD 4–5%) of maximal active lateral bending, according to the velocity. Transverse plane ROM during walking equalled 21–37% (SD 6–11%) of maximal active axial rotation ROM, and also significantly increased with walking velocity ($7 \times 10^{-8} > P > 1 \times 10^{-8}$). None of these motion ranges was correlated to any of the anthropometrical parameters.

Maximal motion velocity (Table 1) was approximately similar in each plane, and increased with walking speed in all three planes. For frontal and transverse plane motion, this increase was significant for all velocities after Bonferroni's correction ($6 \times 10^{-4} > P > 4 \times 10^{-7}$). However, for sagittal plane motion, a significant increase in velocity only occurred between the two higher walking speeds ($P=2 \times 10^{-4}$). There was a significant direct linear correlation between ROM and maximal velocity in each plane ($R=0.59$ – 0.89). Among correlations to anthropometrical parameters, the only significant coefficients concerned the relation between maximal transverse plane motion veloc-

ities at 1.1 and 1.4 m/s and the length of both lower extremities ($R=0.42$ – 0.46); velocities increasing with decreasing extremity length.

The average maximal envelope of frontal and transverse plane ROM during walking presented in Fig. 2 showed the increase of ROM with walking velocity. Horizontal rotation range tended to display a larger increase with increasing walking velocity than lateral bending.

Coupling of rotation and bending was not constant over the population. Homolateral and heterolateral coupling patterns at the maxima of lateral bending and axial rotation were observed (Fig. 3, Fig. 4). In some subjects, modifications of the shape of the coupling envelope were observed with increasing walking velocity (Fig. 3d, for instance). Average ranges of lateral bending at the maxima of axial rotation and vice-versa (Table 2) were close to zero (ranging from -12° to $+11^\circ$), while standard deviations ranged from 3° to 5° . These findings confirm the phase lag between lumbar axial rotation and lateral bending during gait [24, 29, 31]. The coupling characteristics

Fig. 2 Average coupling characteristics of bending (y-axis) and rotation (x-axis) during walking at different speeds. At each maximum of left and right rotation and bending, the range of the other motion component was determined. The four resulting points were linked to create a geometric figure, constituting the envelope of lumbar rotation and bending coupling during gait. Positive axis directions indicate right bending and rotation. Clear markers indicate the center of gravity of the bending versus rotation envelope at different speeds

Fig. 3a–d Bending versus rotation envelopes of four subjects at each velocity. Positive axis directions indicate right bending and rotation. **a** Homolateral rotation at maximal bending, rotation range larger than bending range, envelope center offset towards left rotation and left bending, shape of a parallelogram. **b** Heterolateral rotation at maximal bending, rotation range larger than bending range, envelope center offset towards left rotation and right bending, shape of a parallelogram. **c** Minimal rotation at maximum of bending and vice-versa, similar rotation and bending range, envelope center offset towards left rotation and left bending, shape of a rhombus. **d** Velocity dependent coupling characteristics: no envelope center offset, envelope tending towards a triangular shape

Fig. 4 Proportions of positive (homolateral bending and rotation) and negative (heterolateral bending and rotation) coupling ratios of lumbar lateral bending and axial rotation during walking at different velocities. Ratios were computed at peak rotation and lateral bending ranges (maximal ranges achieved during walking). Percentages of homolateral and heterolateral coupling ratios are less than 100% due to absence or weak range of one component in certain subjects (ratios lower than 0.05 or higher than 20 were omitted)

could not be correlated to any of the parameters sampled during gait nor during maximal active movements. They were not correlated to any anthropometric parameter either, except for the ratio of lateral bending to maximal ro-

tation at 1.4 m/s, which was significantly correlated to most of these parameters. Moreover, walking velocity did not significantly modify the ratios of bending to rotation (paired Student's *t*-test, $P > 5 \times 10^{-2}$).

Table 2 Average (SD) coupling characteristics of lumbar bending and rotation ranges (in degrees) during walking at different velocities (*bold type* indicates maximal motion ranges, *normal type* indicates motion range in other plane)

Velocity	Component	Max R bending	Max L bending	Max R rotation	Max L rotation
0.8 m/s	Bending	7 (4)	-5 (3)	1 (4)	1 (5)
	Rotation	-1 (3)	-2 (3)	5 (3)	-7 (3)
1.1 m/s	Bending	8 (4)	-5 (3)	1 (4)	1 (4)
	Rotation	-1 (3)	-2 (4)	6 (4)*	-8 (3)
1.4 m/s	Bending	8 (4)	-6 (3)*	1 (4)	2 (4)
	Rotation	-1 (3)	-1 (4)	7 (4)*	-9 (3)
1.7 m/s	Bending	9 (4)	-7 (4)	0 (5)	1 (4)
	Rotation	-1 (4)	-1 (5)	9 (4)*	-10 (3)*

* $P < 8.93 \times 10^{-4}$; significant increase as compared to next slowest velocity (Student's *t*-test for paired samples)

Observation of the individual bending versus rotation curves at different walking velocities revealed that for most subjects (80%) the curves were more harmonic at one particular walking velocity (Fig. 5). This velocity, which was characterized by a more regular lumbar spine contribution, was in most cases (16 subjects) 1.4 or 1.7 m/s. Interestingly, there was no correlation between this specific

walking velocity and anthropometric parameters (Spearman's correlation coefficients ranging from -0.07 to 0.22). The only significant Spearman's coefficients between this specific velocity and kinematic parameters (maximal active ROM, lumbar ROM and velocities during walking) were those concerning transverse plane ROM (at 1.1 and 1.7 m/s). A decrease of lumbar axial rotation range during walking at these speeds occurred in subjects with a larger "harmonic" velocity.

Discussion

Several studies have analyzed the behavior of the lumbar spine during walking. Most of these studies have focused on loads supported by the lumbar disks and facet joints [4, 7, 9, 13, 14, 28] and on muscular activation patterns [1, 5, 25, 28] during this activity. However, only a few studies have assessed the three-dimensional motion characteristics of the lumbar spine during gait [24, 29, 30, 31].

The electrogoniometric technique used in this study to sample lumbar motion during gait has several inherent limitations. As is the case for most surface-mounted instruments, displacements between device and skin are difficult to avoid, limiting the reliability of measurements as

Fig. 5 Bending versus rotation graphs at four walking velocities in one subject: **a** at 0.8 m/s, **b** at 1.1 m/s, **c** at 1.4 m/s, and **d** at 1.7 m/s. Positive axis directions indicate right bending and rotation. Note changes in regularity with varying walking velocity

compared to techniques based on implanted markers or medical imaging. Similarly, differences may exist between surface displacement and real joint kinematics. It has, however, to be noted that several studies have shown that, at least when considering the spine, kinematics sampling methods based on surface-mounted tools provide a satisfactory approximation of vertebral displacements [12,26], or results that are in close relationship to vertebral displacements [19,27], although several studies do not support these observations [20,23]. To minimize displacements of the electrogoniometer extremities to the skin, the original thoracic harness has been modified (extendable shoulder straps, more secure buckle). Moreover, the technique used here does not provide data on segmental lumbar kinematics during gait, although such data are of considerable interest in clinical and fundamental research. However, techniques based on medical imaging are difficult to apply during gait, and the use of implanted devices may be subject to ethical criticisms, compared to the non-invasive characteristics of surface-mounted instruments.

The ranges of lumbar motion obtained during gait in this study were comparable to those reported by other authors [29, 30,31]. The results of this study showed that global lumbar spine ROM during walking at low velocities does not exceed one-fourth of maximal lumbar ROM during maximal active trunk movements. Transverse plane ROM and frontal and transverse plane velocity significantly increase with increasing walking velocity – a finding that was not obtained for sagittal plane parameters. Vogt et al. [29] obtained similar results for the displacements of T12 during walking on an ascending slope as compared to level walking. In our study, maximal lumbar ROM did not exceed 40% of maximal active motion ranges.

The direction and magnitude of rotation and lateral bending coupling displayed individual variations during walking. This result confirms the findings of several studies analyzing elementary motion of the lumbar spine [3, 16, 18,22], although some other authors have described coupling as constant (while differing from segment to segment) and well established [15,17]. Several studies have shown that lateral bending is accompanied by rotation to the same side, but that coupling is variable during rotation [16,22]. Coupling characteristics are probably influenced by several parameters, of which lumbar lordosis and posture, muscular control and intrinsic mechanical properties are most frequently put forth [8, 16,17]. However, a correlation of these coupling characteristics to lumbar flexion-extension during walking, which would support the role of lumbar lordosis or posture in this respect, could not be established in the present study. Simi-

larly, a significant correlation of these coupling characteristics to height, weight or lower extremity length was not supported by our results.

Observations suggesting the existence of an individually varying walking velocity, at which lumbar motion patterns are more harmonic and regular, complete our findings. For most subjects, the path of bending versus rotation was very reproducible between cycles at one particular velocity, whereas it was more irregular at other walking speeds. It can be hypothesized that lumbar motion coordination is more precisely organized at a specific walking velocity. Interestingly, this specific velocity was in most cases one of the two higher walking speeds investigated. It could have been hypothesized that subjects displaying a larger stature or longer lower extremities could be characterized by either less or more harmonic involvement of the lumbar spine during walking at higher speeds. However, the results of this work do not provide evidence for a relationship between anthropometric parameters and these harmony variations of lumbar kinematics during gait. The influence of upper body inertia at higher walking velocities could also be advanced to explain this finding, inducing a more regular pendulum effect. It has, however, to be noted that Cavagna et al. [6] showed that, at a walking velocity of approximately 1.1 m/s, the work done to lift the center of mass equals the work done to increase its forward speed, as in a pendulum, and the total mechanical energy is minimal. This velocity, at which the similarity with the pendulum model was largest in the study of Cavagna et al. [6], may be slightly higher in the volunteers involved in our study.

Further studies on trunk muscle activation during walking should be able to throw more light on the kinematic results of this study, and provide a basis for the application of gait analysis in patients suffering from lumbar spine pathologies for the evaluation and follow-up of pathology and treatment.

Conclusions

In this study we showed that lumbar spine rotation ranges and bending and rotation velocities increase with walking velocity. However, sagittal plane motion ranges did not display significant changes with walking velocity. The range and direction of coupled motions were dependent on subject and walking velocity. The hypothesis of the existence of an individually variable walking speed producing a more harmonic lumbar contribution was advanced.

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