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Original paper

Shear force allowance in lumbar spine under follower load in neutral standing posture

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It has been shown experimentally that the load carrying capacity of the spine significantly increases when compressive loads are carried along the follower load (FL) direction. However, it is necessary to modify the current FL concept because a certain amount of shear force is produced during activities in daily life. In this study, a clinically allowable range of shear force was investigated using the modified FL concept. The shear force allowance was defined as the maximum ratio of the shear force to the follower force at each vertebral body center. Then, it was shown that the appropriate shear force allowance was within approximately 0.2 ~ 0.5 from the investigation of the follower forces, the shear forces, and the muscle force coordination. The predicted shear force allowance indicated that the resultant joint force is directed to a certain inside region between a half vertebral body and whole vertebral body.

Key words: *follower load (FL), finite element method, biomechanics, lumbar spine*

1. Introduction

It was shown that the load carrying capacity of the spine was significantly increased with little change in the shape of the spine when compressive loads were carried in a follower load (FL) direction that approximated the tangent to the curve of the lumbar spine at all vertebrae in comparison with the vertical direction [1], [2]. In addition, it was reported that the thoracolumbar spine could support compressive preloads of in vivo magnitudes and allow physiological mobility under flexion–extension moments if the preload was applied in the FL direction [3]. It is essential to demonstrate that the FL can be generated in vivo in order to prove the physiological and clinical feasibility of the FL.

Since it has been reported that the human spine can withstand considerable compressive loads and maintain its stability using the trunk muscles, the FL can be produced by the appropriate coordination of the trunk muscles. Due to the difficulty in carrying out relevant experimental measurements, computational modelling methods have been used to quantify the muscle activity pattern required to carry the applied load in the direction of the FL [4]–[7]. It was recently shown that the trunk muscle coordination pattern used to generate the FL could be predicted, and the load carrying capacity could be increased by muscle coordination using a finite element model of the lumbar spine with a large number of trunk muscles [6], [7].

However, the FL concept allowing for little shear force needs to be modified because the shear forces generated in the human lumbar spine in vivo have been estimated to be quite considerable, typically up to 800 N [8]–[10]. Thus, it is crucial to understand how great shear force could be allowed during the generation of the FL. In this study, the clinically allowable range of shear forces at each

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spinal motion segment was investigated. The modified FL concept was formulated to permit a certain amount of shear force in the three-dimensional finite element model of the lumbar spine. The follower forces, shear forces, and the trunk muscle coordination patterns were predicted in order to determine the clinically relevant shear force allowance.

2. Materials and methods

A three-dimensional finite element model of the lumbar spine from T12 to S1 in a neutral standing posture developed in [6], [7] was used in this study (figure 1). The model included 117 pairs of trunk muscles (5 longissimus pars lumborum, 4 iliocostalis pars lumborum, 12 longissimus pars thoracis, 8 iliocostalis pars thoracis, 11 psoas, 5 quadratus lumborum, 6 external oblique, 6 internal oblique, 1 rectus abdominus, 12 thoracic multifidus, 20 lumbar multifidus, 6 interspinales, 10 intertransversarii, and 11 rotatores). The anatomical data at the initial position of

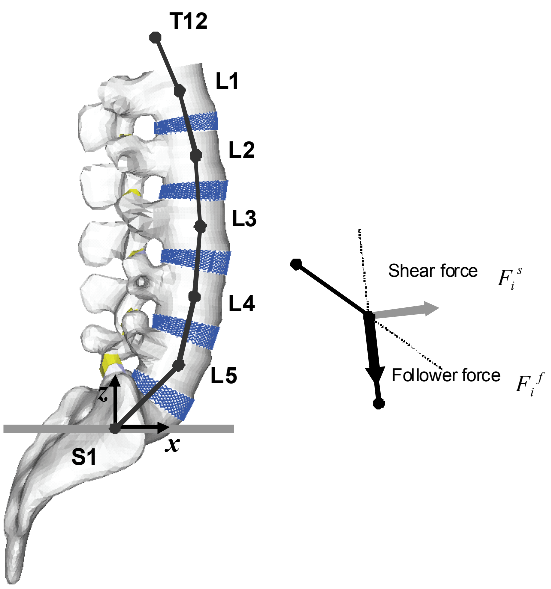


Fig. 1. A three-dimensional finite element model of the lumbar spine was developed from T12 to S1 in a neutral standing posture. Each vertebral body center was identified as a nodal point in the finite element model. A spinal motion segment consisting of a vertebra–intervertebral disc–vertebra was modelled as a linear beam element with elastic stiffness properties. The resultant joint force at the i-th node (i = 1(T12), 2(L1), 3(L2), 4(L3), 5(L4), 6(L5)) could be decomposed into a follower force component $\boldsymbol{F}_i^f$ parallel to the beam direction and a shear force component $\boldsymbol{F}_i^s$ perpendicular to the follower force

the vertebrae, muscle attachments, and physiological cross-sectional areas (PCSAs) were obtained from the literature and medical images [6], [11], [12]. The center of each vertebral body was identified as a nodal point in the finite element model (figure 1). A spinal motion segment consisting of a vertebra–intervertebral disc–vertebra was modelled as a linear beam element with elastic stiffness properties, and the global stiffness matrix was given in [11], [13]. Each muscle was considered to act statically along a straight line between the origin and the insertion given as in [6], [11], [12]. The displacement and rotation at each node, along with the muscle forces satisfying the static equilibrium equations, were estimated for a given applied external load by using an optimization technique. The quadratic sequential programming method was used to solve the optimization problem using MATLAB® (MathWorks Inc., USA).

The modified FL concept was mathematically represented as a constraint in the optimization problem. First, the resultant joint force at the i-th node (i = 1(T12), 2(L1), 3(L2), 4(L3), 5(L4), 6(L5)) could be decomposed into a follower force component $\boldsymbol{F}_i^f$ parallel to the beam direction and a shear force component $\boldsymbol{F}_i^s$ perpendicular to the follower force (figure 1). The modified FL concept was then defined as follows: a small shear force at each node was allowed such as

$$|\boldsymbol{F}_i^s| \le \alpha \cdot |\boldsymbol{F}_i^f|, \tag{1}$$

where the maximum shear force at the i-th node was proportional to the follower forces at the same node for all i. The coefficient α, which was equal to the maximum ratio of the shear force to the follower force, indicates the tangent value of the allowable angle between the FL direction and the direction of the resulting joint force. Let us call α the shear force allowance. When $\alpha = 0$ equation (1) yields the perfect FL, which does not allow for any shear force at the node.

Then, the optimization problem for the estimation of the muscle forces needed to generate the modified FL can be given as follows:

$$\text{Minimize } \boldsymbol{f} = \sum_{i=1}^{6} [\boldsymbol{w}_f (\boldsymbol{F}_i^f)^2 + \boldsymbol{w}_s (\boldsymbol{F}_i^s)^2$$

$$+ \boldsymbol{w}_m \{(\boldsymbol{M}_{i,x})^2 + (\boldsymbol{M}_{i,y})^2 + (\boldsymbol{M}_{i,z})^2\}], \tag{2}$$

$$\text{s. t. } \mathbf{F}_m - \mathbf{K} \cdot \mathbf{d} + \mathbf{F}_{\text{ext}} = \mathbf{0},$$

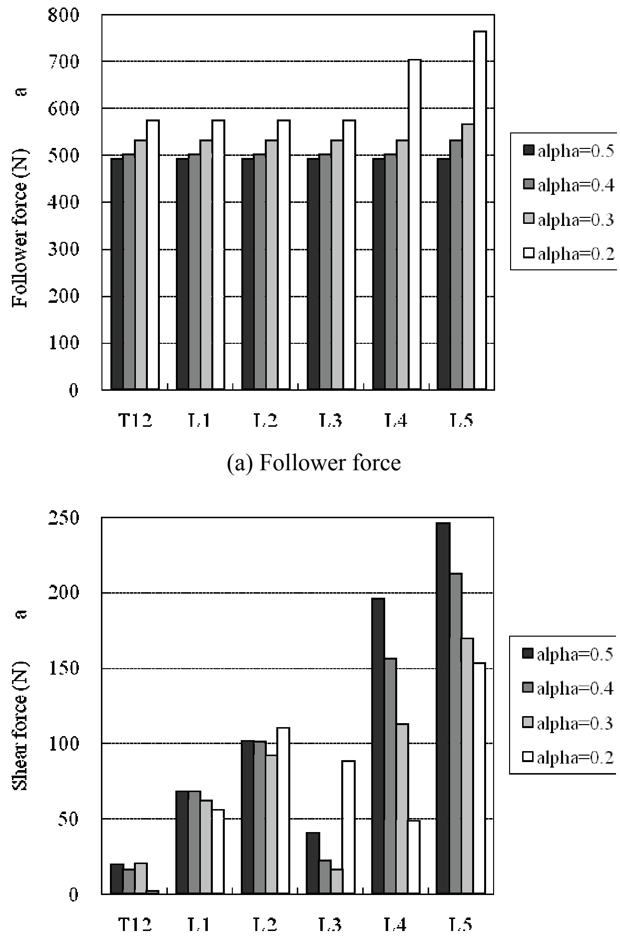
$$|\boldsymbol{F}_i^s| \le \alpha \cdot |\boldsymbol{F}_i^f|,$$

where F_i^f is the follower force, F_i^s is the shear force, $M_{i,x}$, $M_{i,y}$, and $M_{i,z}$ are the joint moments around the *x*, *y*, and *z* axes, respectively, at the *i*-th node, $\mathbf{F}_m$ is the muscle forces and the moments activated by the trunk muscles acting on each vertebrae, $\mathbf{F}_{ext}$ is the external forces and moments acting on each vertebrae, **K** is the global stiffness matrix of the motion segments in the lumbar spine, and **d** is the translations and the rotations at each node as in [6], [7], [11], [14]. In the present study, the upper bound of muscle stress was assumed to be 0.6 MPa based on the measurements reported in [15]. The objective function simultaneously minimizes the follower forces, the shear forces and the joint moment, all of which are primarily related to the spinal stability. The weight factors w_f, w_s, and $\mathbf{w}_m$ are thought to be 3, 3, and 1, respectively, since 3-N force and 1-Nmm moment are weighed equally from the presumed safe limits of intervertebral loads, which are approximately 3000 N for forces and 9000 Nmm for moments as shown in [14].

An upright neutral standing posture was considered for the loading conditions: 300 N of the upper body weight and 3000 Nmm of the resulting flexion moment were applied to T12, and a vertebral weight of 10 N was added to each lumbar vertebra from L1 to L5. The muscle forces required to generate the modified FL as well as the corresponding follower forces and the shear forces were predicted by solving the optimization problem for different shear force allowances, α = 0.0, 0.1, 0.2, 0.3, 0.4, and 0.5. The magnitude of the total muscle force produced by the erector spinae muscles (longissimus and iliocostalis) was also investigated.

3. Results

The muscle force coordination pattern required to generate the modified FL could not be found when α = 0.0 and 0.1. The follower forces increased as α decreased. The follower forces at L5 were 492.2, 530.9, 565.3, and 763.6 N for α = 0.5, 0.4, 0.3, and 0.2, respectively (figure 2a). The maximum shear force occurred at L5 regardless of α and the maximum shear force decreased as α decreased. The shear forces at L5 were 246.1, 212.3, 169.6, and 152.7 N for α = 0.5, 0.4, 0.3, and 0.2, respectively (figure 2b). Finally, the total muscle forces of the erector spinae muscles were 149.6, 189.3, 252.6, and 459.7 N for α = 0.5, 0.4, 0.3, and 0.2, respectively.



(a) Follower force

(b) Shear force

Fig. 2. The follower force (a) and the shear force (b) at each vertebral body center in the neutral standing posture according to the shear force allowance α

4. Discussion

The FL concept needs modification because a certain shear force is produced during normal activities of daily living [9], even in the neutral standing posture [8], [10]. Therefore, the shear force allowance must be considered in order to investigate the effect of trunk muscle coordination on the increase in load carrying capacity from the physiological and clinical points of view [6], [7], [8], [10]. In this paper, we mathematically defined the modified FL using equation (1) and investigated the follower forces and shear forces according to the shear force allowance α, which is the maximum ratio of the shear force to the follower force, in order to determine a feasible range of shear force. The fact that muscle force coordination could hardly be found in the physiologically feasible situation when α = 0.0 and 0.1, which are very close to the perfect FL, supports the modification of the FL concept.

The follower force was substantially great, 763.6 N when $\alpha = 0.2$, while the follower forces were below 600 N when α was larger than 0.2 (figure 2a). However, the compressive follower force between L4 and L5 was 559~650 N, which was estimated from the measured in vivo intradiscal pressure, 0.43~0.50 MPa [16], [17], and the average cross-sectional area of the disc between L4 and L5, 1300 mm^2 [18]. In addition, previous computational studies predicted that the compressive forces were 504~575 N at L5 [8], [10]. Therefore, the compressive follower force was regarded as being greater than the normal range when $\alpha = 0.2$. In addition, the total muscle force of the erector spinae was also high, 459.7 N when $\alpha = 0.2$ in comparison with the previously predicted total muscle forces of the erector spinae, 300 N in [19] and 170 N in [20], in a neutral standing posture.

On the other hand, the shear force was relatively high when $\alpha = 0.5$ in comparison with the value of 190 ~ 218 N at L5 reported in the previous studies [8], [10]. Therefore, it could be concluded that the appropriate α value needed to represent the modified FL was approximately 0.2~0.5. The findings of previous computational studies [8], [10] support our result (the table). The maximum shear force allowances were produced at L5, and their values did not exceed 0.4.

Table. The ratio of the shear force to the follower force at each vertebral body center in the neutral standing posture according to the shear force allowance α

	Present study				[8]	[10]
	$\alpha = 0.5$	$\alpha = 0.4$	$\alpha = 0.3$	$\alpha = 0.2$		
T12	0.04	0.03	0.04	0.00	0.10	0.14
L1	0.14	0.14	0.12	0.10	0.11	0.13
L2	0.21	0.20	0.17	0.19	0.14	0.14
L3	0.08	0.04	0.03	0.15	0.01	0.01
L4	0.40	0.31	0.21	0.07	0.05	0.09
L5	0.50	0.40	0.30	0.20	0.33	0.39

[8] ARJMAND and SHIRAZI-ADL, (2006).
[10] SHIRAZI-ADL, EL-RICH, POP, and PARNIANPOUR (2005).

Equation (1) geometrically demonstrated that the resultant joint force did not deviate from an area near the center of the vertebral body within a certain radius (figure 3). Based on the anatomical data [18], [21] showing that $\boldsymbol{L_f}$, the distance between two adjacent vertebral body centers, and $\boldsymbol{L_s}$, the diameter of the vertebral body, are approximately 30~40 mm,

$$\frac{(\boldsymbol{L}_s/2)/2}{\boldsymbol{L}_f} \approx 0.25 \quad \text{and} \quad \frac{\boldsymbol{L}_s/2}{\boldsymbol{L}_f} \approx 0.5$$

could be obtained. The shear force allowance α, within 0.2~0.5 indicates that the resultant joint force aims at the area near the center of a vertebral body, between a half-radius and full radius of a vertebral body. Therefore, the modified FL allowed the statement that the resultant joint force is directed to a certain region within the vertebral body, from a half body to a whole body.

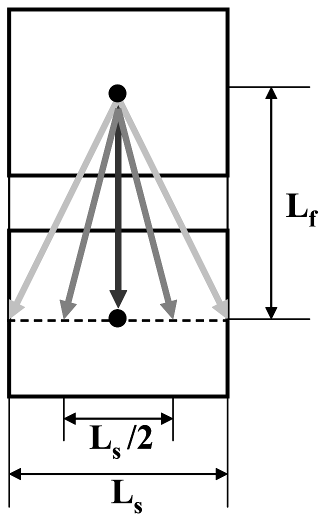


Fig. 3. A diagram explaining the modified follower load concept that the compressive load aims at the area near the body center within a certain radius from a sagittal viewpoint

There were a few restrictions and limitations in this study. The finite element model of the lumbar spine used in this study utilized beam elements of one-dimensional structures as in [4], [8], [10], [11], [14] because long computational time and high cost were required when dealing with a large number of muscles. In addition, the intra-abdominal pressure which also stabilizes the spine [22]–[24] was not considered. Therefore, the validity of our study could be improved by modelling vertebrae, ligaments, and intervertebral discs with realistic anatomical geometry and by taking into account the intra-abdominal pressure. Finally, α needs to be investigated in a variety of postures, such as flexion–extension, lateral bending, and axial torsion, with various external loads during daily life in order to establish the modified FL concept.

5. Conclusion

The shear force allowance was defined and the clinically feasible range was obtained by utilizing the modified FL concept, which allowed for a shear force in a neutral standing posture. The clinically feasible range of shear force obtained could be interpreted as the resultant joint force directed to a certain region between a half body and a whole body. The findings of the present study could be useful for gaining more relevant understanding of the redundant problem in musculoskeletal models.

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