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Comparison of static and dynamic biomechanical models

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To quantify spinal stress biomechanical models are often used. Static models reveal the postural effects due to gravity, while dynamic models also take into account inertial factors. We used both dynamic and static models to evaluate the lumbosacral compression when 20 subjects lifted a box weighing 15 kg from a 10 cm high shelf to knuckle height with four lifting techniques. The mean peak acceleration of the load was 4·9–6·3 ms^{-2}, thus increasing the force at the hands by over 50%. The static peak compression was 3989–4650 N and the dynamic 5866–6629 N, the increase due to inertial factors being 33–60% depending on lifting technique.

1. Introduction

Biomechanical models are used to quantify the spinal stress in manual materials-handling activities in order to compare different handling techniques, to evaluate the effects of job redesign, to predict the individual lifting capacity, etc. The models are based on the analysis of forces and torques acting on the musculoskeletal system of the human body.

Most models have been static (see for example, refs. [205, 408, 556]), thus revealing the postural stress due to gravity. When using static models the effects of acceleration are assumed negligible which leads to underestimation of the spinal stress in dynamic activities, such as lifting.

Grieve [255] and Troup [624] used force platform and accelerometers on the load to illustrate the dynamic aspects of lifting. In dynamic models the inertial forces and torques induced by the accelerations of the load and body segments are added to static forces and torques. Ayoub and El-Bassoussi [46] used in their model a mathematical simulation of acceleration, based on the initial and final positions, and the total displacement time of each joint. Garg *et al.* [242] used digitized photographic data of body movements in their comparison of static and dynamic analysis. Leskinen *et al.* [371, 372] recorded directly the movements of the body with an optoelectronic method and the accelerations of the load with accelerometers.

For this paper 80 lifts were analysed with both static and dynamic models.

2. Materials and methods

2.1. *Subjects*

Twenty male subjects took part. Their ages were 20–42 years (mean 28·3), heights 164–187 cm (mean 117·9) and weights 60–94 kg (mean 72·4). The subjects lifted a box from a shelf, 10 cm above floor level, to knuckle height. The total weight of the box was 15 kg and its handles were set 12 cm from the base.

2.2. *Recording methods*

Movements of the body were recorded using an optoelectronic Selspot system with four infrared light-emitting diode markers attached to the knuckle of the middle finger, shoulder, hip and ankle. A second order Butterworth low-pass filter with the cut-off frequency of 20 Hz was used for each channel.

The acceleration of the load was recorded with two triaxial accelerometers, attached on the front and rear of the box, respectively.

The vertical force at the feet was recorded using a force platform, 60 cm × 60 cm, constructed with strain gauges according to principles described by Terekhov [610].

An ABC 80 microcomputer with 32 kbytes program memory was used for data collection at 100 samples/s for 2·5 s, and for subsequent processing and analysis.

2.3. *Lifting techniques*

All subjects lifted the box using four lifting techniques: back lift (BL), leg lift (LL), load kinetic lift (LKL) and Trunk kinetic lift (TKL) (see the figure):

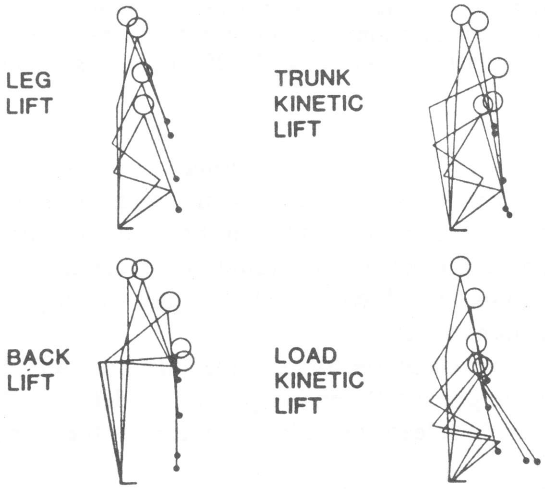

Figure 1. Diagrams of motions in the four lifting techniques (from Leskinen *et al.* [371, 372].

(1) The initial posture for BL was stooped with extended knees and flexed hips.
(2) For LL subjects squatted with flexed knees and hips, and the trunk as erect as possible.
(3) For LKL the initial posture was as for LL, but the box was pulled horizontally towards the body from a position 40 cm in front of the feet, and then swung upwards.
(4) For TKL the initial posture was as for LL, but the hips were first moved vertically by extending the knees, followed by trunk extension and raising of the load.

The subjects practised each lifting technique until they performed the lifts precisely as instructed and only after that were the lifts recorded for the analysis.

2.4. *Dynamic model*

For the analysis of biomechanical forces and torques the concept of the 'free-body-diagram' was employed in this model using only two body-segments: the upper limbs (segment 1), and the trunk above L5/S1 level including the neck and head (segment 2). This was justified for segment 1 as in all lifts from floor to knuckle-height the elbows remained more or less fully extended.

The position of the L5/S1 disc in relation to the hip and shoulder markers was determined using the assumptions: (i) that the pelvis does not rotate during the first 27° of trunk flexion, the initial flexion being spinal; (ii) that after 27°, rotation of the pelvis accounts for two-thirds of the motion and the spine for the rest; and (iii) that the distance from the hip to the disc is 19·2% of the distance between the hip and the shoulder [48].

Anthropometric data from Dempster [186] were used to determine the weights, centres of gravity and moments of inertia of the upper limbs as well as of the trunk, head and neck [660]. To determine these values for the head, neck and trunk above L5/S1 it was assumed that 65·5% of the weight of the trunk was above the L5/S1 level.

The linear (vertical and horizontal) and angular accelerations were obtained as the second derivatives of the positions and angles after determing the centre-points and angles of the body segments. A sliding average of eight successive samples of the position signals was used in addition to the filtering referred above in order to improve the calculation of accelerations.

The force–torque analysis began from segment 1 [371]. The components of the reactive force across and the torque around the shoulders (F_{sx}, F_{sy}, T_s) were obtained from equations (1)–(3):

$$F_{sx} = m_L a_{Lx} + m_1 a_{1x}, \tag{2}$$

$$F_{sy} = m_L(g + a_{Ly}) + m_1(g + a_{1y}), \tag{2}$$

$$T_s = m_L(g + a_{Ly})x_1 + m_L a_{Lx} y_1 + m_1(g + a_{1y})x'_1 + m_1 a_{1x} y'_1 + I_1 \alpha_1 \tag{3}$$

where

m_L = mass of the load,
m_1 = mass of segment 1,
a_{Lx}, a_{Ly} = horizontal and vertical accelerations of the load,
a_{1x}, a_{1y} = horizontal and vertical accelerations of segment 1,
g = acceleration due to gravity,
x_1, y_1 = horizontal and vertical distance from the knuckle to the shoulder,
x'_1, y'_1 = horizontal and vertical distance from the centre of gravity of segment 1 to shoulder,
I_1 = moment of inertia of segment 1 around the shoulders and
α_1 = angular acceleration of segment 1.

The components of force and torque acting at the L5/S1 disc (F_{bx}, F_{by}, T_b) were computed in a similar way, replacing the effects of the external load with the force and torque at the shoulder:

$$F_{bx} = F_{sx} + m_2 a_{2x}, \tag{4}$$

$$F_{by} = F_{sy} + m_2(g + a_{2y}), \tag{5}$$

$$T_b = T_s + F_{sy} x_2 + F_{sx} y_2 + m_2(g + a_{2y})x'_2 + m_2 a_{2x} y'_2 + I_2 \alpha_2, \tag{6}$$

where I_2 is the moment of inertia of the whole trunk + head + neck around L5/S1 and the other parameters corresponding to those previously defined, with the index 2 showing the body segment.

For the musculoskeletal system to be in equilibrium the spinal extensor muscles must create a torque equal and opposite to T_b. As the erector spinae muscle group acts upon the spine with a lever arm of about 5 cm, the muscle force required, F_m, is obtained by dividing the torque by 0·05 m:

$$F_m = T_b/0{\cdot}05\,\mathrm{m}. \tag{7}$$

The total compression at L5/S1, F_c, is obtained by combining the effects of muscle forces and reactive forces taking into account their directions:

$$F_c = F_m + F_{by}\sin\varphi + F_{bx}\cos\varphi, \tag{8}$$

where φ is the angle of the trunk from horizontal.

2.5. *Static model*

For static analysis the same body-segment data were used, but all accelerations were set to zero. Thus, equations (1)–(6) were replaced by the much simpler equations (9) and (10):

$$F_{by} = g(m_L + m_1 + m_2), \tag{9}$$

$$T_b = g(m_L(x_1 + x_2) + m_1(x_1 + x_2) + m_2 x_2). \tag{10}$$

Thus, the reactive force is the sum of the weights of the body above L5/S1 and the load; and the torque is the sum of the products of the weights of the load and body segments and their lever-arms in relation to the lumbosacral disc. Horizontal forces are zero.

2.6. *Parameters*

Two parameters of the L5/S1 compression were studied: the peak compression during a lift, and the compression × time integral for the period when the recorded vertical acceleration of the load was positive.

The inertial force at the feet was obtained by subtracting the weights of the subject and the load from the force-platform signal. The peak force and the force × time integral for the positive force period were the parameters detected.

Statistical analysis was based on one recording of each lifting technique from each subject. The results of each technique were compared to the three others in pairs, using the paired *t*-test. The same lifts were used in both the static and the dynamic analyses.

3. Results

3.1. *Load velocity and acceleration*

Table 1 shows the mean vertical peak velocities and accelerations. Acceleration was smaller in the kinetic lifts than in the 'non-kinetic', but there were no significant differences in peak velocity.

3.2. *Inertial force at the feet*

The peak inertial forces at the feet and the inertial force × time integrals are shown in table 2. Both parameters were highest in LL and lowest in BL.

3.3. *L5/S1 compression*

Peak compressions at the L5/S1 disc for both static and dynamic analyses are presented in table 3, and the compression × time integrals in table 4. The static compression peaks were smallest in BL, but the dynamic peaks of BL were significantly greater than those of LL and LKL. The compression × time integral was smallest in BL with both static and dynamic models.

Table 1. Peak vertical acceleration and peak vertical velocity of the load (from Leskinen *et al.* [372]).

	Acceleration ($m\,s^{-2}$)					Velocity ($m\,s^{-1}$)				
			Significance level					Significance level		
	Mean	S.D.	LL	LKL	TKL	Mean	S.D.	LL	LKL	TKL
BL	6·09	2·26	NS	*	*	1·28	0·14	NS	NS	NS
LL	6·29	1·82		**	***	1·28	0·15		NS	NS
LKL	5·17	0·94			NS	1·25	0·15			NS
TKL	4·94	1·73				1·29	0·21			

NS, non-significant, * $p<0·05$, ** $p<0·01$, *** $p<0·001$.

Table 2. Peak inertial forces at the feet and the inertial force × time integrals (from Leskinen *et al.* [372]).

	Peak (N)					Integral (N s)				
			Significance level					Significance level		
	Mean	S.D.	LL	LKL	TKL	Mean	S.D.	LL	LKL	TKL
BL	244	68	***	NS	NS	62	13	***	***	**
LL	362	105		***	***	119	24		NS	***
LKL	255	43			NS	112	26			***
TKL	258	145				82	28			

NS, non-significant, ** $p<0·01$, *** $p<0·001$.

Table 3. Peak compression at L5/S1 (N) computed with static and dynamic models (from Leskinen *et al.* [372]).

	Static					Dynamic					
			Significance level					Significance level			Dynamic increase (%)
	Mean	S.D.	LL	LKL	TKL	Mean	S.D.	LL	LKL	TKL	
BL	3989	372	NS	***	***	6365	824	**	*	NS	59·5
LL	4033	544		***	***	5866	807		NS	***	45·4
LKL	4548	581			NS	6042	868			**	32·8
TKL	4650	537				6629	867				42·5

NS, non-significant, * $p<0·05$, ** $p<0·01$, *** $p<0.001$.

4. Discussion

The acceleration of the load was on average more than g/2. This indicates clearly the importance of inertial factors when lifting activities are analysed biomechanically.

However, the lower peak accelerations and the smaller forces at the feet were not matched by the peak compressive loads at L5/S1, which was higher in TKL than BL or LL and higher in LKL than LL. Though the dynamic component of compression is

Table 4. Compression × time integral at L5/S1 (N s) computed with static and dynamic models (from Leskinen *et al.* [372]).

	Static					Dynamic					
			Significance level					Significance level			Dynamic increase
	Mean	S.D.	LL	LKL	TKL	Mean	S.D.	LL	LKL	TKL	(%)
BL	1666	296	**	**	***	2268	360	*	*	***	36·1
LL	1884	363		NS	***	2417	245		NS	***	28·2
LKL	1918	319			**	2451	390			***	27·7
TKL	2145	362				2770	443				29·1

NS, non-significant, * $p<0{\cdot}05$, ** $p<0{\cdot}01$, *** $p<0{\cdot}001$.

undoubtedly smallest in LKL, the static component appears to have been supplemented by the horizontal distance of the load from the body. There was a similar increase in the static component of compression in TKL arising from the fact that a lift which begins as LL is converted into a lifting technique resembling BL. The advantage of smaller dynamic components of compression in the 'kinetic lifts' therefore appears to be dissipated. The probable reason is that subjects were unable to perform the lifts with the box as close to the body as in the 'non-kinetic lifts'.

The dynamic component of spinal compression was greatest in BL: arising probably from the comparatively greater angular acceleration of the trunk as compared with the other lifts.

Lanshammar [358] proved that velocities and accelerations calculated by differentiation of noisy movement data may be noisy and erroneous, the errors increasing with lower sampling rate and higher signal bandwidth. This phenomenon was clearly observed in our experiments, too, leading us to use as high a sampling rate and as low a cut-off frequency as possible, without a loss of information.

5. Conclusion

The clear differences in the results calculated with static and dynamic models show that dynamic aspects should be taken into account when lifting work is analysed with biomechanical models. The comparison between static and dynamic components of spinal compression yields valuable information about the lifts.

Afin de quantifier la contrainte spinale, on utilise souvent des modèles biomécaniques. Les modèles statiques révèlent les effets posturaux attribuables à la gravité, alors que les modèles dynamiques prennent également en compte les facteurs d'inertie. Nous avons utilisé, à la fois des modèles statiques et dynamiques pour évaluer la compression lombosacrée auprès de 20 sujets qui devaient soulever une caisse de 15 kg depuis un rayon situé à 10 cm du sol jusqu'à la hauteur de la jointure des doigts, en utilisant quatre techniques de levage différentes. L'accélération de crête moyenne de la charge était comprise entre 4,9 et 6,3 m s^{-2}, ce qui augmente les forces sur les mains de plus de 50%. La compression statique de crête était comprise entre 3989 et 4650 N et la compression dynamique entre 5866 et 6629 N, avec un accroissement dû aux facteurs d'inertie entre 33 et 60% selon la technique de levage utilisée.

Zur Quantifizierung der Wirbelsäulenbelastung werdenoft biomechanische Modelle benutzt. Statische Modelle verdeutlichen die Wirkung der Erdanziehung auf die Körperhaltung, während dynamische Modelle auch Inertialfaktoren berücksichtigen. Wir benutzten beide,

dynamische un statische Modelle, um den Druck im Lendenwirbelbereich zu bewerten. Hierzu haben 20 Versuchspersonen ein 15 kg schweres Gewicht mit vier verschiedenen Techniken von einem 10 cm hohen Sines bis in Kniehöhe. Die Hauptbeschleunigungsspitze beim Anheben betrug 4,9–6,3 ms^{-2}, was einen Anstieg der Handkräfte von über 50% bedeutete. Der Spitzenwert des statischen Druckes betrug 3989–4650 N; im dynamischen Fall waren es 5860–6629 N_0. Der Anstieg entsprechend den Intertialfaktoren betrueg je nach Hebetechnik 33–60%.

脊椎応力の定量化のために生物力学的モデルがしばしば用いられる。静的モデルにより重力による姿勢効果が明らかとなり，動的モデルにより慣性因子を考慮できる。被験者20名が4種類の持上げ法により15kgの箱を床上10cmの棚から指関節高まで持上げた時の腰仙椎応力を動的モデルと静的モデルの両方を用いて評価した。負荷の加速度の平均ピーク値は4.9-6.3 m/s^2で，これは手において50%以上の力の増加を示すものである。静的ピーク応力は3989-4650 N，動的ピーク応力は5866-6629 N，慣性因子による増加は持上げ法に関係し，33-60%の値をとった。

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