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TECHNICAL NOTE

A METHOD FOR MEASURING EXTERNAL SPINAL LOADS DURING UNCONSTRAINED FREE-DYNAMIC LIFTING

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Abstract—Biomechanical lifting models often require the knowledge of the applied trunk moments and forces for model validation purposes and/or to determine loading levels experienced at various joints of the body. Trunk kinetic data under dynamic exertions are commonly difficult to attain without restrictive anatomic/anthropometric assumptions and cost or constraining body motion. The main objectives of the study were to present a new technique for determining continuous three-dimensional forces and moments about the L5/S1 spinal joint, and to validate the technique and assess its applicability under lifting situations. A combination of a force plate and two electrogoniometers facilitated the determination of trunk kinetics about L5/S1. An apparatus was devised to allow the application of various *actual* moments that were compared to their corresponding *predicted* moments. The results showed that, over all the conditions considered, the average percent error in estimating the actual applied moment(s) was about 4% (2.3 S.D.), with a test–retest reliability approaching unity. Given such agreement, along with the relative ease and directness of the method, it is believed that this approach should be applicable under most lifting conditions. The technique offers a fairly accurate measure of trunk moments without the need for constraining the motion of *any* body joint. © 1997 Elsevier Science Ltd

Keywords: Spinal load, Lifting; Force plate; Trunk kinetics.

INTRODUCTION

Several modeling approaches use external loads (net reaction moments and forces) for quantifying and validating the stresses experienced by the spinal structure during task execution (Granata and Marras, 1995; Plamondon *et al.*, 1996). External loads are commonly determined via various techniques such as dynamometers, video-based motion systems, and 'force plates'. Force plates measure the net reaction forces and moments at the feet rather accurately without the extensive data reduction effort common to video-based techniques or the restrictions of the dynamometers (e.g. joint motion restriction, motion type, etc.).

Granata *et al.* (1996) presented a technique that uses the 3-D net reaction loads provided by the force plate to estimate the moments and forces at the L5/S1 joint. In that study, the L5/S1 joint was assumed to be at a *fixed position* relative to a force plate. However, in order to investigate *unconstrained* free-dynamic conditions, the pelvis is expected to be freely moving with respect to the force plate. Hence, it is important to continuously document the relation of the moving (L5/S1) coordinate system to that of the fixed system (force plate). The two main objectives of this study were (1) to develop a technique for determining the continuous 3-D forces and moments about the L5/S1 spinal joint, and (2) validate the technique and assess its applicability to lifting conditions.

METHOD

A 'free body diagram' of the lower body segment below the L5/S1 joint is represented in Fig. 1(a). The distal end of the segment was considered at the force plate (FP) (Bertec 4060A; Bertec, Worthington, Ohio) and its proximal end at the L5/S1 joint. The forces and moments measured by the FP can then be written as

$$\mathbf{M}_{FP} = \mathbf{M}_{L5} + \mathbf{M}_{L} + [(m\,\mathbf{a}_{Lcm}) \times \mathbf{R}_{FPL}] + (\mathbf{F}_{L5} \times \mathbf{R}_{FPL5}),$$

$$\mathbf{F}_{FP} = \mathbf{F}_{L5} + (m\,\mathbf{a}_{Lcm}), \qquad (2)$$

where

$\mathbf{M}_{FP}$ = moment vector acting at the FP; distal end,
$\mathbf{M}_{L5}$ = moment vector acting at the L5/S1 joint; proximal end,
$\mathbf{M}_{L}$ = moment vector acting at the center of mass of the lower body segment (L) due to angular acceleration of the lower extremity,
m = mass of the lower body segment,
$\mathbf{a}_{Lcm}$ = acceleration of the center of mass of the lower extremity,
$\mathbf{R}_{FPL}$ = position vector between the FP and L
$\mathbf{R}_{FPL5}$ = position vector between the FP and the L5/S1 joint.
$\mathbf{F}_{FP}$ = force vector acting at the FP,
$\mathbf{F}_{LP}$ = force vector acting at the L5/S1 joint.

Equations (1) and (2) can be rewritten as follows:

$$\mathbf{M}_{L5} = \mathbf{M}_{FP} - \mathbf{M}_{L} - [(m\,\mathbf{a}_{Lcm}) \times \mathbf{R}_{FPL}] - (\mathbf{F}_{L5} \times \mathbf{R}_{FPL5}), \qquad (3)$$

$$\mathbf{F}_{L5} = \mathbf{F}_{FP} - (m\,\mathbf{a}_{Lcm}). \qquad (4)$$

The forces and moments acting at the L5/S1 joint include those generated by the upper body segment and any applied external loads (e.g. lifted box). The weight of the lower body segment (and upper body) can be systematically eliminated using the *zeroing* option on the FP (Bertec 4060A user's manual). Therefore, the moment vector about L5/S1 would be overestimated by a magnitude of $(\mathbf{M}_L + [(m\,\mathbf{a}_{Lcm} \times \mathbf{R}_{FPL}])$. Similarly, the force vector F_{L5} would be overestimated by a magnitude of $m\,\mathbf{a}_{Lcm}$. However, several studies have demonstrated that the *inertial* effects of the lower body segments played a minimal role in the net moment generated, especially when 'back lifts' are performed (Lindbeck and Arborelius, 1991; Plamondon *et al.*, 1996). Hence, in this study, it was assumed that the unaccounted moment and force vectors due to lower body dynamics would constitute minimal contribution to the total moment and force generated at L5/S1. Therefore, equations (3) and (4) would be reduced as follows:

$$\mathbf{M}_{L5} \approx \mathbf{M}_{FP} - (\mathbf{F}_{L5} \times \mathbf{R}_{FPL5}), \qquad (5)$$

$$\mathbf{F}_{L5} \approx \mathbf{F}_{FP}. \qquad (6)$$

The FP measures the two vectors F_{FP} and M_{FP}, hence, the only unknown is the position vector between the FP and the L5/S1 joint (R_{FPL5}). Two goniometers were developed in order to determine this vector: (1) The L5/S1 3-D Locator, and (2) the Pelvic Orientation Monitor (POM) [see Fig. 1(b)].

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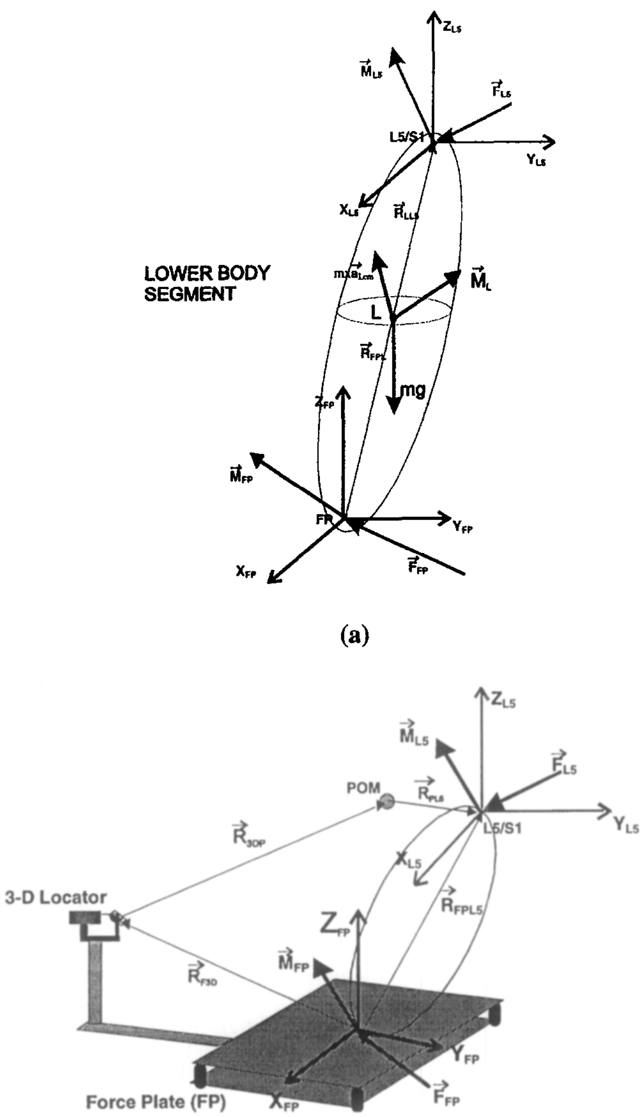

Fig. 1. Free body diagram of the lower body segment (a). The global coordinate system was considered at the distal end (FP), with the local coordinate system at the proximal end (L5/S1) joint. The position vector between the segment's ends (R_{FPL5}) is determined from the position vectors between the FP, the 3D locator, the POM, and the L5/S1 joint (b).

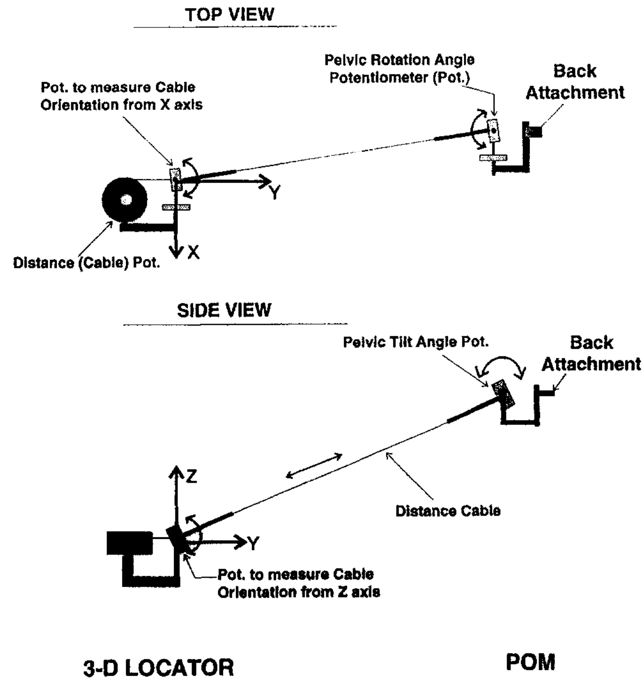

Fig. 2. The L5/S1 3-D locator and the pelvic orientation monitors shown attached to each other. The POM is attached to the subject's lower back region.

The 3-D locator is a device placed at a fixed known distance from the center of the FP (R_{F3D}). The device consisted of three potentiometers that allowed the determination of the continuous 3-D position vector between a point of attachment on the subject's back and the device's center (R_{3DP}) (see Fig. 2). There is a cable that extends from one potentiometer to the subject's lower back at a fixed position from the L5/S1 joint. As the subject moves his/her trunk, the distance between him/her and the location of the goniometer results in a change in the length of the cable, and hence, a change in the potentiometer's voltage. The other two potentiometers were necessary to determine the angular orientation of the cable (angle between the cable and the X-axis, and angle between the cable and the Z-axis). Using a calibration scheme, voltages from the three potentiometers were converted to a distance and two angles (spherical coordinates), which in turn were used to determine the 3-D position vector.

In a neutral posture, the center of the L5/S1 joint is anterior to the subject's skin (R_{PL5}). However, since the pelvis is expected to move with respect to the trunk, the orientation of the 3-D coordinate system at L5/S1 changes with such movement. A Pelvic Orientation Monitor (POM) was developed to monitor the pelvis (L5/S1) orientation. The POM consists of two independent potentiometers designated to detect anterior/posterior pelvic tilt and pelvic axial rotation (about Z-axis), respectively. The device is attached to the subject at the L5/S1 level. It was assumed that pelvic motion in the frontal plane (lateral tilt) is expected to be minimal, especially if the subject maintained both feet on the FP. Using calibration equations, pelvic angular motions were determined through converting voltage readings from the potentiometers to angular positions.

Hence, the position vector R_{FPL5} (FP to L5/S1) can be determined as the vector sum of the three vectors described above [see Fig. 1(b)]:

$$\mathbf{R}_{FPL5} = \mathbf{R}_{F3D} + \mathbf{R}_{3DP} + \mathbf{R}_{PL5}. \tag{7}$$

A static validation was performed to test the accuracy of the method described above in representing moments applied at a given point in space (e.g. L5/S1) (see Fig. 3). A known weight was suspended on a bar attached to a frame which in turn was connected to the FP. An adjustable fixture attached to the frame allowed the bar a limited amount of tilt and rotation, simulating pelvic motion. The POM was attached to the adjustable fixture, and to the 3-D locator via the distance cable. The weight was moved to different points on the bar to obtain different moment arms and/or different moments. For cases where M_Z was applied, a known force (using a force dynamometer) was applied horizontally at given points on the bar.

RESULTS

The average absolute error in the X-axis was 2.21 N m (0.82 S.D.) [Table 1(a)]; whereas in the Z-axis the average error was 1.35 N m (0.17 S.D.) [Table 1(b)]. In the combined X and Y moment conditions, the average absolute error in the X-axis was 2.66 N m (1.54 S.D.), and 1.3 N m (1.4 S.D.) in the Y-axis [Table 1(c)]. Over all conditions (trials), the average percent error was 4.17% (2.3 S.D.). In addition, the test-retest reliability measure was 0.99 ($p < 0.001$) for cases where repeated conditions were performed (see Table 1).

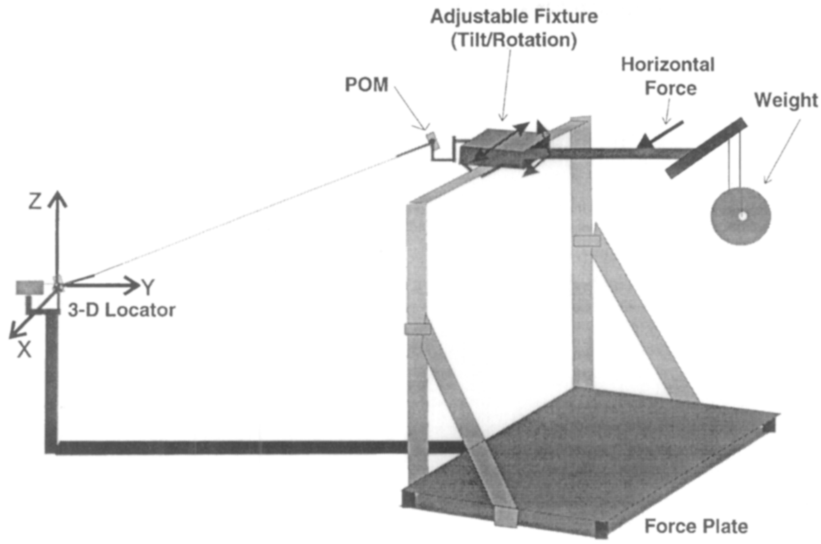

Fig. 3. The apparatus used to apply various external moments on a point equivalent to the L5/S1 location.

Table 1. Actual and predicted moments about the X-axis (a), the Z-axis (b), and the X- and Y-axes combined (c)

Trial	Pelvic rotation (deg)	Pelvic tilt (deg)	$M_{X\mathrm{act}}$	$M_{X\mathrm{prd}}$	e_X $(M_{X\mathrm{act}} - M_{X\mathrm{prd}})$	% e_X $(e_X/M_{X\mathrm{act}})$
(a)						
1	0	44	47.99	46.67	1.31	2.74
2	0	44	47.99	51.01	−3.03	6.31
3	0	0	63.77	65.76	−1.99	3.13
4	0	0	38.26	40.58	−2.32	6.06
5	21	0	61.80	58.87	2.94	4.75
6	21	0	36.30	33.16	3.13	8.64
7	−30	0	63.77	64.65	−0.88	1.39
8	−30	0	39.24	41.33	−2.09	5.34

Trial	Pelvic rotation (deg)	Pelvic tilt (deg)	$M_{Z\mathrm{act}}$	$M_{Z\mathrm{prd}}$	e_Z $(M_{Z\mathrm{act}} - M_{Z\mathrm{prd}})$	% e_Z $(e_Z/M_{Z\mathrm{act}})$
(b)						
9	0	0	81.60	83.16	−1.56	1.91
10	0	0	40.13	41.28	−1.15	2.87
11	0	15	52.55	53.96	−1.41	2.69
12	0	15	38.76	40.06	−1.30	3.34

Trial	Pelvic rotation (deg)	Pelvic tilt (deg)	$M_{X\mathrm{act}}$	$M_{X\mathrm{prd}}$	$M_{Y\mathrm{act}}$	$M_{Y\mathrm{prd}}$	e_X	e_Y	% e_X	% e_Y
(c)										
13	0	26	49.38	53.00	18.15	17.10	−3.62	1.05	7.34	5.76
14	0	26	49.38	51.46	−16.68	−15.94	−2.08	−0.74	4.22	4.42
15	0	0	66.71	69.63	18.15	18.59	−2.92	−0.44	4.38	2.42
16	0	0	66.71	68.50	−16.68	−17.22	−1.80	0.55	2.69	3.28
17	−30	0	66.71	69.58	−16.68	−15.04	−2.87	−1.64	4.30	9.81
18	−30	0	66.71	66.51	20.60	21.02	0.20	−0.42	0.30	2.04

Note: Errors (and % error) in predicted moments are shown for various combinations (trials) of pelvic rotation and tilt angles. The magnitude of the differences are considered negligible when compared to the range of moments experienced during lifting, making the approach applicable under most lifting conditions.
$M_{X\mathrm{act}}$, $M_{Y\mathrm{act}}$, $M_{Z\mathrm{act}}$ = Actual applied moment about the X, Y, and Z axes.
$M_{X\mathrm{prd}}$, $M_{Y\mathrm{prd}}$, $M_{Z\mathrm{prd}}$ = Predicted moment about the X, Y, and Z axes.
e_X, e_Y, e_Z = Error in actual applied moment about the X, Y, and Z axes (actual moment-predicted moment).
% e_X, % e_Y, % e_Z = Percent error in actual applied moment about the X, Y, and Z axes.

DISCUSSION

The results showed that there were some discrepancies between the actual (applied) moments and the predicted (calculated) ones, especially under some of the combined moments conditions. However, in general, the magnitudes of the differences are rather negligible relative to moments commonly experienced at the L5/S1 joint during common lifting conditions (100–300 N m.; e.g. Dolan *et al.*, 1994; Gagnon *et al.*, 1993; McGill and Norman, 1986).

This static validation of the measured moments should be applicable under dynamic conditions given that minimal inertial forces are generated by the lower limbs. As discussed earlier, these forces have minimal contribution to the total kinetics about L5/S1, especially when no squatting is involved (straight-leg lifting) (Lindbeck and Arborelius, 1991; Plamondon *et al.*, 1996). Additional apparatus could be devised to facilitate the measurement of these moments (in order to eliminate their contribution). In addition, future studies could further investigate this issue by validating the method presented here under actual lifting tasks (e.g. current method versus inverse dynamics), and/or under simulated dynamic moments conditions.

As outlined by Granata *et al.* (1996), one of the important advantages of this method was that external loads were computed directly. Although the equivalent L5/S1 moment due to the net applied force is very difficult to measure, the proposed method computed it from force and moment vectors recorded by the FP, and the position vectors determined by the two electrogoniometers. This would eliminate the error associated with double differentiation of noisy position data when estimating body segment masses and moments of inertia as commonly done in other types of kinetic analyses, (e.g. inverse dynamics).

This method provided the means to determine the kinetics about the L5/S1 joint, with reasonable accuracy, under unconstrained free-dynamic situations where the hip and legs are not constrained to a fixture (though both feet should be on the force plate; hence, limiting the motion space). This information is an integral part of several biomechanical modeling approaches that attempt to better understand the nature and complexities of the loading patterns experienced by the back. Although this approach was designed mainly to acquire information about kinetics at the lower back region, the method can be extended to other joints of the body.

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