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THE IMPORTANCE OF ANTAGONISTIC COCONTRACTION OF TRUNK MUSCLES FOR SPINAL LOADS DURING LIFTING AND PULLING TASKS: IMPLICATIONS FOR MODELING APPROACHES.

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INTRODUCTION

It has been suggested that tasks involving considerable trunk extensor efforts may entail compression forces on the lumbar spine which can reach a high enough level to cause damage to the vertebral endplates and consequent low-back pain (Dieën, Weinans, & Toussaint, 1999). However, since direct measurements of compression forces are impossible, such conclusions rely on estimates derived from biomechanical analyses. Therefore, the accuracy with which these compression forces can be estimated from a biomechanical analysis deserves attention. The basic problem in this type of analysis is the fact that the many muscles spanning the lumbar joints constitute a mechanically indeterminate system. Consequently, a given net moment about a joint can be produced by a range of combinations of muscle forces. Therefore, assumptions with respect to the distribution of the net moments across muscles have to be made. Unfortunately, sensitivity analyses have shown that compression force estimates are quite sensitive to such assumptions. Among these, the assumption made regarding the presence and intensity of abdominal cocontraction during extensor efforts has the most prominent effect (Dieën & Looze, 1999).

Two common approaches in estimating spinal compression forces treat the issue of cocontraction quite differently. The first approach uses surface electromyography to obtain estimates of forces in individual muscles. The level of cocontraction in abdominal muscles in this case is assumed to be reflected by the electrical activity picked up from a limited number of superficial motorunits in the abdominal muscles. The second approach based on mathematical optimization estimates the distribution of the net moment by minimizing a function of the muscle forces or muscles activities. Since these cost functions are usually based on some form of efficiency criterion, cocontraction is assumed to be absent. The effect of cocontraction is often used as an argument for using the former approach (e.g. Cholewicki, McGill, & Norman, 1995). However, it should be kept in mind that this method is more cumbersome in terms of data acquisition. Furthermore, although it uses more of the available biological information, it still relies on several assumptions, e.g. regarding the activity in deep muscles and regarding the relationship between electrical activity and muscle force.

The aim of the present study, therefore, was to compare compression force estimates based on the two approaches in realistic manual materials handling tasks involving substantial extensor efforts (lifting and pulling).

METHODS

Kinematic, kinetic and EMG data of 8 pairs of trunk muscles were collected from two groups of ten and five subjects participating in experiments on lifting boxes and pulling four-wheeled carts, respectively. In the lifting experiment subjects lifted 6 and 16 kg boxes from a position 25 cm in front of their feet and 10 cm above the ground to waist level, at their preferred speed using a self-selected technique. In the pulling experiment subjects pulled carts weighing 85 and 320 kg with two hands over a distance of 4 m on a level rubber surface.

Standard rigid link inverse dynamics were used to calculate net moments acting on the lumbar spine. Subsequently, an EMG driven distribution model, was used to estimate spinal compression forces. The model containing 90 muscle slips crossing the L5S1 joint, has in part been described previously (Dieën & Kingma, 1999). Muscle forces were estimated as the product of maximum muscle stress, normalised EMG amplitude, and correction factors for instantaneous muscle length and contraction velocity plus the passive force developed by the muscle's connective tissue. The correction factors are based on dynamical properties of human and animal muscles as described by Zandwijk (1998) and the passive length tension properties were modeled after Woittiez et al. (1984). Maximum muscle stress was iteratively adjusted to obtain maximum agreement between the time series of muscle moments and net external moments (cf McGill & Norman, 1986). The optimization driven version of the model used the maximum muscle stress derived from the EMG driven analysis and as cost function the sum of the cubed muscle activation levels (Dieën, 1997). This cost function aims to minimise fatigue development, similar to the cost function proposed by Crowninshield and Brand (sum of muscle stresses cubed, Crowninshield & Brand, 1981), but recognises the fact that muscle length and contraction velocity influence the relation between fatigue development and muscle force.

Estimates of the peak muscular compression force at the L5S1 disc from both models were compared using an ANOVA with task (lifting / pulling) as a between subjects factor and task intensity (high or low weights of cart and box) and model version as within subject factors. Subsequently, relationships between peak compression force estimates from both model versions and peak net moments were determined using linear regression analysis with a forced zero-intercept.

RESULTS

Compression force estimates were significantly affected by the model version (EMG versus optimization), the task intensity, and the activity performed (lifting versus pulling). Overall compression estimates obtained using EMG were about 300 N higher than estimates based on optimization (table 1). In lifting the weight effect was much stronger compared to the effect of the estimation method. In contrast, a very limited effect of cart weight was found in pulling, comparable in magnitude to that of the estimation method.

Table 1. Mean compression forces for lifting and pulling tasks of high and low intensity estimated by EMG and optimization based methods.

	low intensity		high intensity	
	EMG	optimization	EMG	optimization
lifting	4994	4769	6253	5850
pulling	3832	3378	3705	3566

All compression force estimates were strongly interrelated and correlated to the net moment (correlation coefficients all above 0.90; figure 1). Slopes of the regression lines did not differ significantly between tasks performed or task intensities. The slope of the regression line describing the relationship between the net moment and the EMG based compression estimate was 25.2 m^{-1}, for the optimization based compression estimate it was 23.7 m^{-1}.

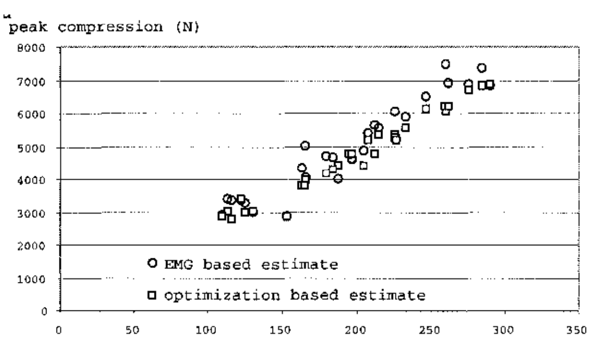

Figure 1. Peak net moments relate strongly to the estimates of compression force obtained by the two model versions.

DISCUSSION

The main finding of the present study was a significant difference between EMG based and optimization based estimates of spinal compression forces in lifting and pulling. This implies that cocontractions of abdominal muscles do influence the compression force during these extensor efforts to some extent. However, in line with a previous study (Dieën & Kingma, 1999), the effect was only small (about 300 N). In contrast with the present findings, Cholewicki et al. (1995) indicated that more substantial differences in compression force estimates between optimization and EMG based methods occur. Since the model geometry in these studies especially regarding the abdominal musculature appears fairly similar, the explanation for this disparity probably is the difference in cost functions used in the optimization. Cholewicki et al. used a cost function minimizing subsequently the maximum muscle stress and the compression force. This will in some tasks predict substantially lower compression forces than the cost function used in the present study (Hughes, 1995), while the latter was shown to make more valid predictions of muscle activity (Dieën, 1998). To our knowledge no previous studies have been performed on the effect of the estimation method on compression force estimates in pulling tasks.

Two explanations can account for the limited difference between the estimation methods. First, abdominal cocontraction influences spinal compression only to a limited extent during extensor efforts. This is in line with previous studies (Looze, Groen, Horemans, Kingma, & Dieën, 1999; Potvin, McGill, & Norman, 1991). However, Granata and Marras (1995) reported a strong influence of abdominal cocontraction on spinal compression in lifting. This disparity is probably accounted for by the difference in lifting task (lifting from the floor versus lifting from a position with 45 degrees inclined trunk; Looze et al., 1999) and the differences in model geometry. Second, the EMG based estimation of compression force caused by the extensor musculature is based on assumed equal activation of deep and superficial muscle slips, whereas the optimization approach predicts slightly higher activation of the superficial muscles (longer lever arms).

The results of this study indicate that optimization with the present cost function will provide estimates of peak compression force which are systematically but only slightly lower than the estimates obtained through an EMG driven model. Overall the correlation coefficient between the two compression estimates was 0.96, the regression coefficient obtained with a forced zero-intercept 0.94. This shows that optimization can be used as an alternative for EMG based modeling in many ergonomics applications, the difference between the outcomes being on average only 6%. Data presented in a related paper by Hoozemans et al. in these proceedings suggest that this conclusion will not hold for pushing tasks in which the extensor effort is less dominant. In asymmetric tasks cocontraction may also be more prominent.

A strong relationship was found between the peak net moments and peak compression force estimates. This suggests that a single-equivalent model (SEM) could be used to analyze these tasks. The regression coefficients of the relationships between net moment and compression estimate indicate that an effective moment arm of about 0.04 for this SEM would be appropriate. This is in line with previous suggestions based on the same model geometry (Dieën & Looze, 1999). In comparison to other studies (McGill & Norman, 1987; Potvin, Norman, Eckenrath, McGill, & Bennet, 1992) this effective moment arm is however rather small. Differences between tasks (degree of trunk flexion) may have contributed to this disparity, but in addition our model is rather small (Dieën & Looze, 1999). The latter implies that to obtain more reliable estimates of the absolute magnitude of the compression forces, subject specific scaling of the model should be performed. This does however not affect the comparisons made in this study. The conclusion that a SEM is useful for estimating spinal compression forces, should again be restricted to the type of tasks performed in the present study.

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