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Clinical Biomechanics 15 (2000) 703–717

CLINICAL BIOMECHANICS

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Review paper

The effects of motion on trunk biomechanics

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Received 15 June 2000; accepted 16 June 2000

Abstract

Objective. To review the literature that evaluates the influence of trunk motion on trunk strength and structural loading.

Background. In recent years, trunk dynamics have been identified as potential risk factors for developing low-back disorders. Consequently, a better understanding of the underlying mechanisms involved in trunk motion is needed.

Methods. This review summarizes the results of 53 studies that have evaluated trunk motion and its impact on several biomechanical outcome measures. The biomechanical measures consisted of trunk strength, intra-abdominal pressure, muscle activity, imposed trunk moments, and spinal loads. Each of these biomechanical measures was discussed in relation to the existing knowledge within each plane of motion (extension, flexion, lateral flexion, twisting, and asymmetric extension).

Results. Trunk strength was drastically reduced as dynamic motion increased, and males were impacted more than females. Intra-abdominal pressure seemed to only be affected by trunk dynamics at high levels of force. Trunk moments were found to increase monotonically with increased trunk motion. Both agonistic and antagonistic muscle activities were greater as dynamic characteristics increased. As a result, the three-dimensional spinal loads increase significantly for dynamic exertions as compared to isometric conditions.

Conclusions. Trunk motion has a dramatic affect on the muscle coactivity, which seems to be the underlying source for the decrease strength capability as well as the increased muscle force, IAP, and spinal loads. This review suggests that the ability of the individual to perform a task "safely" might be significantly compromised by the muscle coactivity that accompanies dynamic exertions. It is also important to consider various workplace and individual factors when attempting to reduce the impact of trunk motions during dynamic exertions.

Relevance

This review provides insight as to why trunk motions are important risk factors to consider when attempting to control low-back disorders in the workplace. It is apparent that trunk motion increases the risk of low-back disorders. To better control low-back disorders in industry, more comprehensive knowledge about the impact of trunk motion is needed. A better understanding of muscle coactivity may ultimately lead to reducing the risk associated with dynamic exertions.

Keywords: Electromyography; Spinal loading; Trunk strength; Intra-abdominal pressure; Trunk moments

1. Introduction

Despite mechanization increases in the workplace, manual material handling has remained a vital component of work. As electronic commerce becomes more prevalent, more frequent lifting (often with lower force) must occur to process on-line orders. More frequent lifting often is associated with more rapid movements [1,2]. Hence, modernization has impacted the work by making lifts more dynamic. Similarly, in the manufacturing environment, better methods and faster machines have increased output, requiring the worker to keep pace with the process. Thus, a better understanding of how trunk dynamics may impact the worker is becoming more important.

Trunk motion has recently been identified as a potential risk factor for low-back disorders (LBD). Three-dimensional trunk velocity has been found to significantly increase LBD risk [1–3]. Marras and associates [1,2] reported odds ratios (comparing "high risk" to "low risk") for sagittal, lateral, and twisting trunk velocity on the job between 1.34 and 3.3, 1.04 and 1.73, and 1.09–1.66, respectively. Norman et al. [3] found an odd ratio of 1.9 for peak sagittal trunk velocity when

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comparing cases to randomly selected controls. In all three of these studies, trunk velocity variables were included in the multivariate models, indicating that these motion characteristics may play an important role in the development of LBD. In addition, high values of combined trunk velocities (e.g. simultaneous lateral flexion and twisting velocities) were found to occur more frequently in high and "medium" risk jobs than low risk jobs [4]. Thus, dynamic motion plays a dominant role in development of LBD, particularly when the motion occurs in multiple planes simultaneously. However, our knowledge of the underlying injury mechanisms associated with dynamic trunk motion is limited.

1.1. Potential mechanisms leading to LBD

Many factors may contribute to the relationship between dynamic motion and LBD risk. Two major categories of such factors include: trunk strength and spine structural loading. While both factors may potentially cause LBD, each typically is thought to be associated with different mechanisms of LBD. LBDs that result from lack of strength would exceed muscular tolerance while structural loading LBDs would result from forces being placed on the spinal structures exceeding the tolerance limits. In either case, LBD risk is viewed in a load-tolerance perspective, regardless of whether it is muscular or skeletal in nature.

Dynamic strength of the trunk has typically been measured with sophisticated dynamometers that control the motion while measuring the force being exerted. Typically, the motion has been isolated to one plane of motion (e.g. sagittal, lateral, or twisting). The premise behind strength capacity is that the closer the required strength (load) is to the strength capacity (tolerance), the greater the risk of injury [5,6].

Structural loading factors would include the biomechanical factors that contribute to loading on the spinal structures such as intra-abdominal pressure (IAP), muscle activity, and the imposed trunk moment, as well as the actual loads on the structures of the spine. IAP has traditionally been thought of as a mechanism that assists in generating an extensor moment, thus, reducing the loading on the spine [7]. Others have suggested that IAP assists in maintaining the integrity of the abdominal muscles [8] and vertebral motion segments [9], as well as stability of the vertebral column [10,11]. Muscle activity has traditionally been used as an indirect indicator of the level of force generated by a particular muscle [12]. However, it should be noted that these measurements are not an indicator of muscle tension but rather the degree of muscular activation solicited from the muscle. In order to estimate the tension in the muscle, the signal must be adjusted (modulated) to account for the length and velocity of the muscle [12]. The trunk moment imposed on the spine (external load) needs to be offset by the trunk muscles (internal load) [13,14]. Thus, as the imposed trunk moment increases, there would be a corresponding increase in muscle activity. As a result of the muscle activity and, potentially, IAP, loads on the spine are generated in the form of compression, anterior–posterior (A–P) shear (front-to-back shear forces), and lateral shear (side-to-side shear forces) forces.

This review investigated the impact of trunk dynamics on trunk strength and structural loading variables and attempted to provide insight into how these factors may be the underlying causes for trunk motions that may be associated with risk of LBD. The literature was summarized across the all planes of motion providing a comprehensive evaluation for each of the potentially contributing factors. This provided opportunity to draw conclusions as well as identify major voids in the literature.

2. Methods

2.1. Selection of articles

The current review encompasses all articles published in English before February 2000 that evaluated various biomechanical measures during three-dimensional dynamic motions. An extensive literature search was conducted using Medline and Institute for Scientific Information databases as well as looking at the reference lists of the accumulated articles. Typically, there have been five groups of biomechanical measures that have been commonly investigated in literature: strength, IAP, trunk moments, spinal loads, and muscle activity (agonistic and antagonistic). Based on these areas, searches of the databases were accomplished using the following key words: trunk motion, strength, IAP, muscle activity, spinal loading, kinematics, low-back motion, and lifting. Inclusion criteria for the review required that the study to have at least two levels of trunk velocity, either uncontrolled (e.g. slow) or isokinetic (e.g. 30°/s or 10 cm/s). Studies were also required to have a minimum of six subjects. In all, 54 studies were found to evaluate multiple levels of trunk velocity.

For each study, the relative impact of increased velocity for the appropriate biomechanical measures was determined by computing the percent difference relative to the lowest velocity. For example, the reference value for a study that had both isometric and isokinetic exertions would be the static exertions while studies without isometric exertions used the slowest velocity level as the reference. When conclusions were drawn about the various biomechanical measures, the studies having a static exertion as the reference value were considered to provide the most information about the impact of increased dynamics. More credence was also given to studies that controlled for trunk velocity than studies relying on subjective motion (e.g. slow). Thus, the "high quality" studies were considered to be ones that evaluated controlled velocities and had an isometric reference exertion.

3. Results

3.1. Overview of the results

Tables 1–5 provide a summary of the studies that evaluated trunk velocity. For each study, the velocity conditions and study population are described along with the results for the corresponding biomechanical measures. The results were grouped by the primary direction of motion consisting of extension (Table 1), flexion (Table 2), lateral flexion (Table 3), twisting (Table 4), and extension with asymmetry (Table 5). Forty-three of the studies evaluated extension exertions, eight studies assessed flexion motion, four studies evaluated lateral flexion exertions, eight investigated twisting motion and nine studies provided information about extension exertions with asymmetry. Notice, several studies evaluated multiple directions of motion.

By comparing the tables, one can appreciate how the various studies have evaluated extension exertions and particularly dynamic strength. Of the 46 extension studies, 22 studies evaluated strength while less than 12 studies reported results within any of the other biomechanical measures. All of the flexion studies assessed strength, as did the majority of the twisting (78%) and lateral flexion (75%) studies. Few studies have investigated IAP, muscle activity and spinal loads for lateral flexion and twisting exertions. Twenty-two of the studies (out of 54) evaluated both genders while 29 of the studies evaluated males only. Two of the studies provided no information about the gender of subjects and one used an empirical model to evaluate the effect of trunk velocity.

As stated earlier, two criteria were used to rate the quality of the information derived from these studies. First, higher quality studies were considered those that controlled the trunk motion rather than relying on subjective measures of velocity (e.g. slow, fast, normal, etc). Seventy-six percent (41 studies) of the studies controlled the trunk velocity while the remaining 13 studies used subjective velocities. Second, studies that compared both isometric and dynamic exertions were felt to provide a better appreciation and quantification of the role dynamics and were considered higher quality. In these studies, isometric exertions were used as the reference to determine the impact of velocity while the other studies had the slowest dynamic exertion as the reference. Thirty-five studies (65%) evaluated both isometric and dynamic exertion. By combining the articles that satisfy each of these criteria, 32 studies (59%) were judged to provide the highest quality of information when interpreting the impact of dynamics on the spine (designated by bold letters in Tables 1–5).

3.2. Trunk strength

Dynamic motion generally decreased trunk strength in all directions by approximately 10–30%. Most of the high quality studies (14 studies) evaluating extension strength (Table 1) reported diminished strength of 10–30% for dynamic exertions compared to static conditions with only a few studies finding some postures having increased strength. The dynamic exertions with increased strength occurred during more awkward postures (e.g. load placed at a full reach away from the body) [38] or in an upright posture (as indicated by the upward pointing arrows in Table 1) [31,49]. It appears that some of the variability in the strength results for extension may be attributable to whether the exertion was isolated to the L5/S1 joint (e.g. lower body was constrained) or was a whole body assessment (e.g. able to move legs during exertion).

Flexion strength decreased for dynamic exertions by 5–80% for all but two studies (Table 2). Smith et al. [57] reported that flexion strength increased by 34–75% as compared to the isometric exertion. Khalaf et al. [31] found similar results for males but not for the females. The three studies that investigated dynamic lateral flexion strength reported a decrease of 11–88% for dynamic lifts (Table 3). Dynamic twisting strength was found to be 15–80% lower than for isometric twisting for the high quality studies (Table 4). Only two twisting studies reported an increase in strength for the dynamic lifts, one study reported increases for females only [57] while the other included only one condition [66]. The only two studies that evaluated asymmetric extension exertions found mix results ranging from a decrease of 63% to an increase of 91% (Table 5).

For the studies that evaluated both genders, an interesting trend in the impact of dynamics of the strength of the individuals emerged. Strength for the males decreased with dynamic exertions by about 30% while the females' decrease in capacity was about 15–20%. This would mean that trunk motion had more an impact on the male strength than the females by about 10%. This gender finding was relatively consistent across all exertion directions.

In summary, an individual's strength is reduced by 10–30% when exertions are performed dynamically as compared to isometric strength. In other words, given the same exertion level, a dynamic exertion would be closer to the tolerance of the muscle than during a static exertion, resulting in more risk of a muscular injury. Trunk motion appears to have a greater impact on the males' strength than on the females.

Table 1
Results of the studies that evaluated the impact of trunk dynamics on the trunk strength and structural loading factors for *Extension* exertions[a]

Study	Velocity	Study population	Strength (%)	IAP (%)	Trunk moments (%)	Agonist activity (%)	Antagonist activity (%)	Spinal loading (%)
Buseck et al. [15]	Fast, normal	10 M			↑6–18			
Bush-Joseph et al. [16]	Slow, normal, fast	10 M			↑3–49			
Cresswell and Thorstensson [17]	0.12, 0.24, 0.48, 0.72, 0.96 m/s	7 M	↓4–18	↑4–21		↑3–138	↑1–29	
Davis et al. [18]	5, 10, 20, 40, 80°/s	10 M						Fz ↑8–13 Fy ↑7–17
De Looze et al. [19]	0.2, 0.4, 0.8 m/s	5 M			↑5–57			
Dempsey et al. [20]	0, 0.1, 0.2, 0.4, 0.6, 0.8 m/s	25 M	↓10–31					
Dolan and Adams [21]	Slow, medium, fast	10 M/F			↑25–75			
Dolan et al. [22]	0, Slow, medium, fast	21 M 18 F			↑40–167			
Dolan et al. [23]	Slow, fast	8 M			↑17–32			
Fathallah et al. [24]	1, 1.5, 2 s/lift	11 M						No diff
Gallagher [25]	0, 30, 60, 90°/s	21 M	↓20			↓34–44	↓20–38	
Grabiner and Kasprisin [26]	60, 90, 120°/s	9 M	↑7			↓35–61		
Granata and Marras [27]	0, 30, 60, 90°/s	10 M				↑34–109	↑50–431	Fz ↑27–52 Fy ↑23–57
Granata and Marras [28][b]	0, 30, 60, 90°/s	10 M			↓21–30			Fz ↑75–120 Fy ↑25 Fx ↑100
Granata et al. [29]	Preferred, faster than preferred	10 M			No diff			No diff
Hall [30]	Slow, medium, fast	10 M/F		No diff	↑0–227			Fz ↑21–178 Fy ↑35–137
Khalaf et al. [31]	0, 10, 20, 40, 60, 80, 100°/s	10 M 10 F	↓66–↑33					
Kim and Marras [32]	Slow, medium, fast	8 M				↑25–33		Fz ↑8
Kingma et al. [33]	Normal, rapid	8 M			Sag ↑11–14 Lat ↑6–10 Twt ↑5–9			

Kishino et al. [34]	0, 18, 30, 36 in/s	23 M 42 F	↓5–23					
Kumar [35]	0, 60 cm/s	9 M	↓39–↓18					Fz ↓22–↑15
Kumar [36]	0, 30°/s	59 M 43 F	↓24–93					
Kumar and Davis [37]	0, Stoop not controlled	8 M		↑117–273		↑155–184		
Kumar and Garad [38]	0, 50 cm/s	18 M 12 F	↓22–↑45					
Kumar et al. [39]	0, 20, 60, 100 cm/s	10 M 10 F	↓3–18					
Langrana et al. [40]	0, 30°/s	50 M 26 F	↓11–20					
Lavender et al. [41]	Slow, fast	12 M 3 F			↑16			
Lindbeck and Arborelius [42]	Slow, fast	12 M			↑35–41			Fz ↑33–41
Mandell et al. [43]	0, 18, 30 in/s	14 M 7 F	↓23–33					
Marras [44]	0%, 33%, 67%, 100% of max velocity	10 M 10 F		↓16–25		↓26–88		
Marras et al. [45]	0%, 33%, 66%, 100% of max velocity	10 M 10 F	↓37–70	↓17–31		↓26–88	↓7–↑56	
Marras et al. [46]	0%, 33%, 66%, 100% of max velocity	10 M 10 F	↓37–70	↓4–45				
Marras et al. [47]	0,15, 30, 90°/s	45 M	↓13–69			↑7–33		
Marras et al. [48]	0,15, 30, 90°/s	45 M	↓8–49			↑16–↓41		
Marras and Mirka [49]	0, 10, 20, 30°/s	34 M 10 F	↓31–↑25					
Marras and Mirka [50][b]	0, 10, 20, 30°/s	34 M 10 F		No diff		↑11–44	↑0–150	
Marras and Mirka [51][b]	0, 15, 30°/s	20 M				↑4–66		
Marras and Mirka [11]	0, 10, 20, 30°/s	94 M 20 F		↑0–160		↑5–46[a]		
Marras and Sommerich [52]	0, 10, 20, 30°/s	11 M						Fz ↑3–11
Newton et al. [53]	60, 90, 120°/s	35 M 35 F	↓5–17					
Parnianpour et al. [54]	Not controlled	42 M	↓3.8 (per 10°/s)					
Reilly and Marras [55]	0%, 25%, 50%, 75%, 100% of max velocity	10 M						Fz ↑11–21 Fy ↓61–↑90 Fx ↓31–62
Shirado et al. [56]	30, 60°/s	20 M 16 F	↓11–13					
Smith et al. [57]	0, 30, 60, 90, 120°/s	62 M 63 F	↓1–20					
Sutarno and McGill [58]	0, 3, 5, 9.7, 14.9, 20.1, 26.9°/s	10 M				↑14–171		
Thorstensson and Nilsson [59]	0, 15, 30°/s	14 M	↓4–18					

[a] Bold studies indicate high quality studies.
[b] Studies reported results across all asymmetry conditions since the asymmetry by trunk velocity interaction was not significant.

Table 2
Results of the studies that evaluated the impact of trunk dynamics on the trunk strength and structural loading factors for *Flexion* exertions[a]

Study	Velocity	Study population	Strength (%)	IAP	Trunk moments	Agonist activity	Antagonist activity	Spinal loading
Khalaf et al. [31]	0, 10, 20, 40, 60, 80, 100°/s	10 M 10 F	↓21–↑47					
Kumar [36]	0, 30°/s	59 M 43 F	↓5–82					
Langrana et al. [40]	0, 30°/s	50 M 26 F	↓6–↑5					
Newton et al. [53]	60, 90, 120°/s	10 M 16 F	↓3–11					
Parnianpour et al. [54]	Not controlled	42 M	↓4.8 (per 10°/s)					
Shirado et al. [56]	30, 60°/s	20 M 16 F	↓13–16					
Smith et al. [57]	0, 30, 60, 90, 120°/s	20 M 16 F	↑34–75					
Thorstensson and Nilsson [59]	0, 15, 30°/s	14 M	↓20–30					

[a] Bold studies indicate high quality studies.

Table 3
Results of the studies that evaluated the impact of trunk dynamics on the trunk strength and structural loading factors for *Lateral Flexion* exertions[a]

Study	Velocity	Study population	Strength (%)	IAP (%)	Trunk moments (%)	Agonist activity (%)	Antagonist activity (%)	Spinal loading (%)
Kumar [36]	0, 30°/s	59 M 43 F	↓16–88					
Kumar et al. [60]	0, 30°/s	41 M 32 F	↓16–88					
Marras and Granata [61]	0, 15, 30, 45°/s	12 M				↑46–116%	↑20–133%	Fz ↑10–23% Fy ↑50% Fx ↑100–325%
Thorstensson and Nilsson [59]	0, 15, 30°/s	14 M	↓11–30					

[a] Bold studies indicate high quality studies.

Table 4
Results of the studies that evaluated the impact of trunk dynamics on the trunk strength and structural loading factors for *Twisting* exertions[a]

Study	Velocity	Study population	Strength (%)	IAP (%)	Trunk moments (%)	Agonist activity (%)	Antagonist activity (%)	Spinal loading (%)
Kumar [36]	0, 30°/s	59 M 43 F	↓17–86					
Kumar et al. [60]	0, 15°/s	41 M 32 F	↓17–86					
Marras and Granata [62]	0, 10, 20°/s	12 M	↓43–50			↑56–157	↑64–252	Fz ↑50 Fx ↑15–45
Marras et al. 1998 [63]	0, 10, 20°/s	12 M			↓4–↑30	↓23–↑60	↑6–55	
Mayer et al. [64]	30, 60, 120, 180°/s	25 M 42 F	↓15–25					
McGill [65]	0, 30, 60°/s	10 M 5 F				↓8–27	↓3–58	
McGill and Hoodless [66]	0, 30, 60°/s	10 M 21 F	↓49–↑5					
Newton et al. [53]	60, 120, 150°/s	35 M 35 F	↓6–10					

[a] Bold studies indicate high quality studies.

Table 5
Results of the studies that evaluated the impact of trunk dynamics on the trunk strength and structural loading factors for *Asymmetric Extension* exertions[a]

Study	Velocity	Study population	Strength (%)	IAP (%)	Trunk moments (%)	Agonist activity (%)	Antagonist activity (%)	Spinal loading (%)
Fathallah et al. [24]	1, 1.5, 2 s/lift	11 M						No diff
Jager and Luttmann [67]	0, 0.75, 1.0, 1.5 s	Model						Fz ↑1–188 Fy ↑1–250 Fx ↑1–333
Kingma et al. [33]	Normal, rapid	8 M			Sag ↑5–22 Lat ↓5–↑10 Twt ↓6–↑10			
Kumar and Garad [38]	0, 50 cm/s	18 M 12 F	↓24–↑91					
Lavender et al. [41]	Slow, fast	12 M 3F			Sag ↑16 Twt ↑8			
Marras and Mirka [49]	0, 10, 20, 30°/s	34 M 10 F	↓63–↑14					
Marras and Mirka [11]	0, 10, 20, 30°/s	94 M 20 F		↑0–500				
Marras and Sommerich [52]	0, 10, 20, 30°/s	11 M						Fz ↑3–6
Smith et al. [57]	0, 30, 60, 90, 120°/s	25 M 42 F	↓10–↑33					

[a] Bold studies indicate high quality studies.

3.3. Intra-abdominal pressure

The studies evaluating IAP provided very inconclusive results when comparing dynamic trunk motion to isometric exertions. In the high quality studies that evaluated extension exertions (Table 1), IAP was found to decrease in four studies, no difference in one, and increased in one when dynamic motion was involved. No studies have evaluated IAP while performing dynamic flexion, lateral flexion, or twisting exertions. Marras and Mirka [11] combined the results of four individual studies evaluating IAP and concluded that IAP increased monotonically with trunk velocity. However, this increase in IAP appeared to be only at high levels of exerted force. To date, there has been no comparison of the impact of dynamics on IAP between males and females.

In summary, the results for IAP were inconclusive. The impact of dynamics on IAP seems to be greatest at high exertion levels. There is a need for a better understanding of the influence of trunk dynamics on IAP, particularly differences between genders, as well as an improved appreciation of the biomechanical role of IAP.

3.4. Trunk moments

Overall, the moments imposed on the trunk have been observed to increase by 15–70% during the dynamic exertions. The majority of the studies that evaluated trunk moments were not considered to be in the high quality category since most of them did not control velocity. For extension exertions (Table 1), studies that used subjective motions found increases ranging from 3–227% while controlled motion exertions actually saw the trunk moments decrease with the introduction of dynamics (by 12–69%). In the study that found a decrease in trunk moment, the lower torso was restricted which may have influenced how the exertions were performed (e.g. subject pulled the weight closer to the body during the dynamic conditions). No difference between trunk moments was found in one study [29] but there was no difference in the actual trunk velocity between the two subjective velocity categories (preferred and faster than preferred). Similarly, McIntyre et al. [68] reported that individuals gravitated towards a preferred trunk velocity when exerting below 25% of their maximum effort, which may provide insight into the results of Granata et al. [29]. Similar increases in trunk moment were reported for asymmetric extension exertions (Table 5). No studies evaluated the imposed trunk moments for flexion and lateral flexion. The only study to evaluate imposed twisting moments was Marras and associates [63] who found the trunk moments to increase 15–25% in most conditions. To date, differences in the impact of trunk motion between males and females have yet to be explored.

In summary, imposed trunk moments during dynamic exertions were found to be greater than during isometric exertions, monotonically increasing with faster motions. Since much of the research evaluating dynamic trunk moments have relied upon uncontrolled velocities, there is a need for further investigation of how imposed trunk moments are affected by trunk motion under more controlled conditions, especially in the non-extension exertions.

3.5. Muscle activity

In general, both agonistic and antagonistic muscle activity increased with increased dynamic trunk motion. Agonistic activity increased by approximately 10–40% while antagonistic activity increased by as much as 450%. There have been no studies that have evaluated muscle activity while performing dynamic flexion exertions and few evaluating twisting or lateral flexion. For the 10 high quality extension studies (Table 1) evaluating agonistic muscle activity, most (all but four) found an increase in activity with dynamic motion (range of 4–109% increase). Gallagher [25] reported decreases in the activity of the Latissimus Dorsi, which accompanied small (non-significant) increases in erector spinae muscle activity. Decreases in agonistic activity were also found for very fast extensions (e.g. 90°/s) [48]. The only study that evaluated lateral flexion and satisfied the inclusion criteria found agonistic muscle activity to increase by 46–116% with dynamic motion [61]. Two of three studies found increases in agonistic activity in twisting (6–157%) (Table 4). Four studies found decreases in agonistic activity with trunk motion [25,44,45,65]. The exerted force for dynamic condition velocities also decreased as compared to the isometric exertions, thus, the decrease in activity may be more reflective of the force exerted. Marras et al. [63] reported a decrease in agonistic activity when the subjects twisted in awkward postures only.

Few studies (9) have investigated the impact of trunk dynamics on the antagonistic muscle activity. Since these muscles are not required for force generation, increased activity indicates more coactivity of the trunk musculature system and less efficiently but probably related to stability requirements. Results from the high quality studies indicated that antagonistic activity might be affected more by dynamics than the agonistic muscles. For the extension exertions (Table 1), dynamic motion increased antagonistic activity by about 50–450% while dynamic lateral flexion exertions (Table 3) increased the antagonistic activity by 20–133%. Although McGill [69] did not directly compare the antagonistic muscle activity during dynamic lateral flexion to static, he noted that dynamic exertion resulted in moderate levels of co-contraction, in other words, presence of activity in the antagonistic muscles. Dynamic antagonistic muscle activity increased in two of the three twisting studies (by 6–252%) (Table 4).

In summary, trunk dynamics dramatically influenced both the agonistic and antagonistic muscle activity. Since the antagonistic muscle activity was impacted more, the overall coactivity of the trunk musculature was significantly increased without resulting in any additional force generating capability. Few studies have evaluated the muscle activity while performing flexion, lateral flexion, and twisting exertions. Future studies are also needed to investigate how trunk dynamics might influence muscle activity for males and females differently.

3.6. Spinal loading

In general, trunk dynamics was found to increase the three-dimensional spinal loads with increases of 10–50 % for compression, 50–325% for lateral shear, and 10–30% for A–P shear. The high quality extension studies (Table 1) reported an increase of 3–120% in compression force when dynamic trunk motion was compared to isometric exertions. Two of these studies reported increased A–P shear forces during dynamic exertions (by 23–57%). Reilly and Marras [55] (the only other study to report on A–P shear force), found a decrease in shear for the slowest velocities (less than 30°/s). Only two of the studies evaluating extension exertions (Table 1) evaluated lateral shear forces with one reporting a decrease and one reporting an increase with the dynamic conditions. This apparent conflict in results may be a direct reflection of the low levels of expected lateral shear force during sagittally symmetric exertions. In the one study that reported lateral shears in asymmetric extensions (Table 5), the lateral shears increased by 1–333%. However, this study was not considered to be in the high quality study group.

Only one study found in the literature [61] investigated spinal loads during lateral flexion (Table 3). These authors reported an increase in compression, lateral shear, and A–P shear force by 10–33%, 50%, and 100–325%, respectively. The only study included in the present review to evaluate spinal loads during twisting exertions reported a 50% increase in compression and a 15–45% increase in the lateral shear forces during the dynamic exertions [62]. Combining the results of these studies with the one asymmetric exertion study, non-sagittal trunk motions had a large impact on the lateral shear forces. No studies could be located that evaluated spinal loads during flexion exertions.

In summary, trunk dynamics had a significant impact on three-dimensional spinal loads. Dynamic sagittally symmetric exertions affect spinal compression and A–P shear forces while non-sagittal (lateral flexion and twisting) exertions result in higher compression and lateral shear forces. The impact of trunk dynamics on spinal loads for females has yet to be investigated. Since females would be expected to have different trunk anatomies [70–75], the impact of trunk dynamics on spinal loads may be different for females as compared to males. More studies are needed to evaluate the spinal loads during dynamic flexion, lateral flexion, twisting, and asymmetric extension exertions.

4. Discussion

Based on this review, it is clear that trunk dynamics has a significant impact on how the individual performs an exertion and how the trunk musculoskeletal system behaves. Trunk motion severely reduces the individual's ability to generate force. This would mean that when a task requires a certain level of force to be employed, more dynamic motions would hinder the ability of the worker to meet the demands of the job. Potentially, the mismatch between the individual's strength capability and the job demands may result in increase risk of the LBD in industry, specifically, muscular injuries.

One factor that is highly dependent upon dynamic strength is the dynamic functional capability of the individual. Several studies have attempted to quantify the functional capacity of an individual without an LBD [76–84]. In general, these studies report the dynamic functional capability for sagittal, lateral, and twisting motion to be 40–140°/s, 70–120°/s, and 65–170°/s, respectively, which was dependent upon the force exerted. This indicates that high risk jobs found in the Marras studies [1,2] and the cases in the Norman study [3] would have velocities that were much closer to their expected dynamic capabilities as compared to the corresponding low risk and control jobs.

In addition, trunk strength would also be directly linked to muscle coactivity. Decreases in strength capacity during dynamic exertions suggest that more muscle force would be required to respond to the external load demands. In most cases, the activity of the agonistic muscles as well as the antagonistic muscles increased during dynamic motions. This indicates that there is not only more activity in the primary force generating muscles present, but also more overall coactivity of the trunk musculature in general.

Higher levels of coactivity have a significant impact on the spinal loads since increased antagonistic muscle activity must be offset by the agonistic forces. In other words, the muscle activity from the antagonistic muscles produces more loading on the spinal structures without contributing to the ability to offset the external moment imposed on the spine. In the studies that evaluated spinal loads, trunk dynamics were found to significantly increase the compressive forces on the spine and seemed to be independent of the type of motion used during the exertion. However, the impact of dynamics on the shear forces was much more dependent on the exertion type. Dynamic sagittally symmetric extensions increased the A–P shear forces while non-sagittal motions impacted the lateral shear forces. In all types of dynamic exertions, there was more three-dimensional or complex loading on the spine.

The spine tolerance literature suggests that disc strain and vertebral segment failure increases significantly when loading occurs in multiple directions simultaneously [85–89]. This indicates that trunk dynamics places the spinal structures at increased risk of failure since spinal loads increased in multiple directions for all types of motions. Fathallah et al. [24] found that the load rate for compression and A–P shear increased with trunk motion during sagittally symmetric motion and for all three directions (compression, A–P shear, and lateral shear) for asymmetric lifting. The rate at which the load is applied to the spine also influences the mechanical properties of the disc [90–95]. Wang et al. [94,95] predicted that the stresses in the annulus fibers increased with load rates using a finite-element model. However, Yingling and associates [96,97] found that the ultimate strength of the spinal motion segments increased with faster load rates. These authors also indicated that while the magnitude of the loads affects the tolerance of the spine, the rate of the load actually influences the site of the failure. Thus, both the magnitude of the loads as well as the rate of the loads has the potential of being the underlying mechanisms that explains the relationship between trunk motion and LBD risk [1–3]. Both of these factors have been found to be predictive of high-risk of LBD [98]. These indices provide a measure of how likely a task resembles a high risk job.

The load imposed on the spine may also have had an impact on the resulting muscle activities and subsequent spinal loads. Some of the increase in muscle activity (especially for the agonistic muscles) and spinal loads may have resulted from the higher imposed trunk moments that accompany dynamic motions. When trunk moments are increased, the trunk muscles have to increase their output to offset the moment, which result in the higher spinal loads.

The importance of IAP in the reduction of the trunk moment was not substantiated entirely with many studies finding conflicting results. The relationship between IAP and trunk motion is poorly understood since most of the studies were confined to the extension exertions (either sagittally symmetric or asymmetric). With no clear relationship between trunk motion and IAP, it appears that IAP may be a by-product of the muscle activity rather than an active contributor to offsetting of the imposed trunk moments [11].

Trunk motion impacts the musculoskeletal system by altering the recruiting patterns of the trunk muscles. Muscle coactivity appears to be the driving force behind the diminished strength and functional capability that accompanies trunk dynamics as well as the increase spine structural loading (IAP, trunk moments, and spinal loads). The increased co-activation of the trunk musculature associated with increases in trunk motion may have resulted from programming of the neurological pathways that control the muscles and are fine-tuned through experience [99]. McIntyre et al. [100] provides evidence of motor recruitment programs by reporting that individuals adopt a preferred trunk velocity when performing an exertion. In other words, an individual will adopt a certain motion profile that is accompanied by a specific co-activation pattern based on previous exposure to similar motions and these motor recruitment programs are constantly updated.

Based on this premise, trunk dynamics provides a concise representation of the current status of the trunk musculoskeletal system. Many studies have found trunk velocity and acceleration to be better discriminators between individuals with and without LBD than range of motion measurements (Table 6). When an individual becomes injured, the musculoskeletal control program must be adjusted to compensate for limitations related to diminished muscle functioning, structural restrictions, and guarding behavior [81,99]. These adaptations to the motor control program are manifested in the high-order motion profiles (velocity and acceleration) for the individual with LBD. Although the recruitment pattern will be changed as a result of the injury, the pattern would be expected to be consistent. Marras and associates [99] reported that impairment magnification (insincere effort) resulted in more variability in the motion profile, providing further evidence of a central motor recruitment program that guides trunk motion. Collectively, these studies point toward trunk motion as a biomarker of the status of the lumbar spine.

4.1. Other factors that influence trunk motion

Several other factors may influence the trunk motions of the individual as well as the corresponding biomechanical outcomes. One factor that may impact how the exertions are performed is gender. The only studies in the current review that have evaluated gender differences were limited to strength assessments. These studies found males to be impacted more by trunk dynamics than females, that is, males had larger decreases in strength for dynamic exertions. Males have also been reported to have higher levels of functional capability to perform dynamic motions, that is, males are able to move up to 40% faster than females [79,80]. It may be this discrepancy in functional capability (strength) that results in trunk dynamics impacting males more. It would also be possible that the muscle activity patterns

Table 6
Summary of studies evaluating the impact of trunk motion on the ability to discriminate between LBP and non-LBP individuals

Study	Range of motion	Velocity	Acceleration
Bishop et al. [100]	*	**	**
Esola et al. [101]	NS[a]		
Ferguson et al. [76]	*	**	**
Gomez [102]	*		
Kaigle et al. [103]	**		
Klein et al. [104]	*		
Langrana et al. [40][b]	*		
Mandell et al. [43]	NS[a]		
Marras et al. [79]		**	**
Marras et al. [80]	*	**	**
Marras et al. [81]	*	**	**
Marras and Wongsam [105]	*	**	
Masset et al. [82]	NS[a]	**	
Mayer et al. [106]	**		
McClure et al. [107]	NS[a]	*	
McIntyre et al. [108]			**
Pope et al. [109]	*		

[*] Indicates either mixed results or weaker LBP and non-LBP discrimination ability (e.g. P-values below 0.05 or error rates above 40%).
[**] Indicates stronger LBP and non-LBP discrimination ability (e.g. P-values below 0.01 or error rates below 25%).
[a] Indicates no significant LBP and non-LBP discrimination ability (e.g. P-values >0.05 or error rates above 50%.
[b] No statistical test performed.

and subsequent spinal loads would be different between genders, and thus, the impact of trunk motion on the loads may also be affected. Differences in muscle activity and spinal loads may result from strength differences as well as muscle anatomy variations [70–75]. Strength capability differences between the genders may also influence the dynamic trunk moments imposed on the spine.

Trunk dynamics has also been influenced by several workplace factors such as box weight, task asymmetry, handles on the box as well as how the lifting task was performed (e.g. mode of lift). Typically, when the amount of box weight increased, the trunk motion was found to decrease by 1–27% in the sagittal plane [110–115]. Allread et al. [116] found that increased box weight resulted in larger lateral trunk velocities during asymmetric lifting. Task asymmetry has a major effect on the trunk motions with more asymmetric tasks having greater three-dimensional trunk velocities. The sagittal and lateral trunk motions were found to increase by 20–50% while a much greater impact was seen for twisting velocities (100–300% increase in motion) [110,116,117]. Conversely, one study found no effect of asymmetry of the trunk motions [29]. Sagittal trunk velocity was found to be higher when handles were not present on the box by about 3°/s [118]. Thus, workplace factors have the potential to alter the motions of the worker, ultimately influencing the spinal loads and subsequent risk of LBD.

The mode in which the exertion is performed impacts the trunk motion adopted. Gagnon and Smyth [113] found that individuals lifted faster as compared to lowering exertions. The biomechanics during lifting and lowering were also found to be different with lowering having greater strength, lower muscle activities, and higher spinal loading [18,50,119–122]. Marras and Mirka [49] found strength to increase with lowering exertions (up to 14%). The number of hands used during lifting has also been found to influence the trunk motions with two-handed lifts having more sagittal motion (10–30% more) and one-hand lifts having more lateral motion (about 50%) [116,117]. There would appear to be the potential for the lifting technique adopted to influence the trunk motion, although the current results are conflicting [123,124]. While the results to date are less than conclusive, the way the exertions are performed has an impact on the motion within the trunk.

Individual factors such as anthropometry or LBD status also can influence trunk motions. Many studies found the functional capacity of individuals with a LBD to be diminished by as much as 70% [76,79–81] and had lower dynamic strengths of about 20–40% [34,40,53,56,60,64]. The impact of trunk motion on the strength for these individuals was less than for asymptomatic individuals [40,60]. Dynamic strength decreased by 27% for extension, 5% for lateral flexion, and 15% for twisting exertions. On the other hand, Lagrana et al. [40] actually found increased strength in flexion for the dynamic exertions (about 45%). Body compositions may also have a role in how an individual performs a certain exertion. Factors such as body weight and height may alter the trunk motions. For example, taller individuals may have to bend farther forward when lifting, possibly causing them to move faster during the exertion. However, research indicating how body dimensions influence trunk motions is very limited.

Finally, fatigue of the musculature has also been found to influence the trunk motions, muscle activities, and subsequent spinal loading. Several authors have reported that trunk motion in the primary plane (e.g. flexion and extension) decreases as an individual becomes fatigued while motion in the off-planes (e.g. lateral flexion and twisting) increased [125,126]. Sparto et al. [127,128] found dynamic strength to decrease and a more stoop lifting style was adopted as the subject fatigued. Marras and Granata [129] have also reported that lifting motion changed over a 5-h lifting session, resulting in decreases in trunk motion but increase in hip motion. While these studies have provided insight into how fatigue influence trunk motion, our review failed to yield any studies that compared the effect of fatigue during dynamic and static exertions, simultaneously. A better understanding of the impact of fatigue during dynamic and isometric on trunk strength, muscle coactivity, and spinal loads is needed.

4.2. Future research

As can be seen from the Tables 1–5, evaluation of trunk motion has predominantly been performed for extension exertions or assessing trunk strength. A complete understanding of the influence of trunk dynamics will require more studies evaluating the other biomechanical measures, especially for the non-extension exertions. More complex motions will also need to be evaluated. Some of the best insight may come from assessments that include all directions of exertion within the same study. There is also a need for additional epidemiological studies to further establish trunk motion as a risk factor for LBD since few of these types of studies exist currently.

Additionally, the development of more easy-to-use techniques for the quantification of the trunk motion and the corresponding biomechanical measure needs to be undertaken. Many of the techniques employed for the studies in this review require elaborate measurement devices such dynamometers, motion analysis systems, and electromyography. While it is important to have sophisticated measurements to gain an in-depth understanding, simpler measurement tools will allow assessment of trunk motion in the actual workplace. Through improved ability to provide quicker and more versatile assessment of the workplace, a better understanding regarding links between trunk dynamics and risk of LBD may be obtained. Future research needs to focus on the pathways in which trunk motion may lead to LBD, particularly linking it to actual incidences of LBD.

4.3. Potential limitations

Many of the summaries for the various biomechanical outcomes were based on few studies. While this review is the first attempt to understand the impact of trunk motion on common biomechanical measures, the actual number of high-quality studies was particularly limited for the non-extension exertions. In several cases, there were not any studies under a given biomechanical factor. However, these recommendations do represent the state of the literature and what is currently known.

With any review, the conclusions drawn are only as strong as the studies being evaluated. Since different researchers adopt vastly different measurement techniques (e.g. different strength assessment devices, controlled the exertion differently, various processing and testing procedures, etc.) as well as different experimental conditions, total consensus within the results becomes difficult. With this in mind, the review attempted to address this issue by only including studies that control the trunk motion and have an isometric reference. However, more rigorous experimental methods may provide a more definitive account of the impact of trunk motion on the musculoskeletal system.

5. Conclusion

Trunk motion had a dramatic affect on the muscle coactivity, which may dictate other biomechanical outcomes. It appears the increased muscle coactivation that accompanies increased trunk motion may be the underlying source for the decrease strength capability as well as the increased muscle force, IAP, and spinal loads. Muscle coactivity may also influence the physical capability of the individual. Based on the results in the current review, the ability to perform a task without risk may be significantly compromised by the muscle coactivity that accompanies more dynamic exertions. In addition, many workplace and individual factors have been found to influence the trunk motions during dynamic exertions. It is apparent that trunk motion increases the risk of LBD and to better control LBDs in industry, more attention to trunk kinematics due to the job is needed.

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