Clinical Studies

An electromyographic study of isokinetic axial rotation in young adults

Shrawan Kumar PhD., DSc.,* Yogesh Narayan BSc., Doug Garand BSc

Department of Physical Therapy, 3-75 Corbett Hall, University of Alberta, Edmonton, AB T6G 2G4, Canada

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Abstract

Background context: Trunk rotation is associated with over 60% of all low back injuries. However, there are gaps in the knowledge about trunk rotation.

Purpose: To study the axial rotation torque and electromyographs (EMGs) of seven trunk muscles bilaterally in static and isokinetic modes with increasing angular velocity to determine qualitative and quantitative muscle response.

Study design: An electromyographic study of seven trunk muscles bilaterally was carried out in 50 normal subjects during static and isokinetic axial rotation at varying angular velocity. The qualitative and quantitative force and EMG measures were made and analyzed phenomenologically and statistically.

Methods: Fifty normal young adults (27 men, 23 women) performed an isometric maximal voluntary contraction (MVC) from the neutral upright-seated posture to their right and left. Also, these subjects performed isokinetic axial rotation from neutral seated position to right and left, and from prerotated right and left “end of range posture” toward the neutral at 10, 20 and 40 degrees per second angular velocity. All experimental trials were made on axial rotation tester designed for the study. EMG was recorded from the erector spinae at L3 and T10 vertebral levels, latissimus dorsi, pectoralis major, rectus abdominis and external and internal obliques bilaterally. The torque and EMG in isokinetic condition were normalized against those of isometric condition. Descriptive statistics were calculated. Data were subjected to analysis of variance, and torque was regressed on EMG.

Results: The peak isokinetic torques were significantly lower than the isometric torques (p<.01), but the EMGs of the isokinetic conditions were significantly higher than those of isometric trials (p<.01). The isokinetic axial rotation torque decreased by 1 Nm to 1.5 Nm with per degree increase in velocity of rotation, with the values for 10, 20 and 40 degrees per second angular velocities being significantly different (p<.01). There was a significant increase (p<.01) in percent EMG (up to 28%) per unit torque with increasing angular velocity of rotation. The rotation torque from prerotated position to neutral was significantly higher than that of rotation away from the neutral (p<.001). The EMG magnitude increased significantly with increasing velocity (2% to 17% at 10 degrees per second), 21% to 28% at 20 degrees per second, 30% to 36% at 40 degrees per second). Regression analysis revealed poor predictability of torque based on EMG. The latter was suggested because of the role and behavior of the ligaments and joint capsules of the spine.

Conclusion: The axial rotation is initiated and maintained by the contralateral external obliques, ipsilateral latissimus dorsi and internal oblique. The ipsilateral erector spinae likely play a stabilizer role. The isometric torque is greater than the isokinetic, which decreases with increasing velocity. Even with decreasing torque, EMG progressively increases, indicating a disproportionally higher stress in the spinal connective tissues potentiating injury. The data presented here suggest that, for safety, load and velocity of rotation should be kept low. © 2003 Elsevier Science Inc. All rights reserved.

Keywords: Axial rotation; Isokinetic; EMG; Low-back injury; Trunk muscles; Trunk twisting

Introduction

Trunk rotation continues to remain an enigmatic phenomenon that has eluded full understanding [1–9] Rotation of the trunk is a common activity of daily living and is performed to varying degrees in walking, running, racquet sports,
golfing, hockey, baseball, cricket, lifting, carrying, shipping, warehousing and many other activities. It has also been assigned as the only factor in causation of 33% of back injuries [10]. Manning et al. [11] in their epidemiological study found that trunk rotation was associated with over 60% of all back injuries. Social and economic impact of this affliction in our society is well known. Because the torque generation in trunk twisting motion is considered hazardous, several studies have looked into the torque generational capacity in rotational motions [7,8,12–17]. Trunk strength profile with various degrees of asymmetry and various modes of lifting or strength exertion is described in these studies.

Anatomical and geometrical complexities of spine are extremely challenging to explain the entire aspect of rotation all at once. Many studies have tackled different aspects and have begun to furnish some data. Kumar and Panjabi [18] investigated the neutral zone of the thoracolumbar spine in vivo after application of incremental loads. With application of load, the range of rotation progressively increased but the neutral zone was not significantly affected up to application of 36 Nm torque. To describe the phenomenon of axial rotation away from and toward neutral posture, Kumar et al. [15] reported an electromyographic study of unrestressed trunk rotation with normal velocity in normal healthy subjects. They reported initiation of the trunk rotation by the activities of contralateral external oblique and ipsilateral latissimus dorsi. It was reported that a return from prerotated to neutral posture was primarily powered by the regulated release of stored elastic energy of the connective tissues by the very muscles that caused the rotation in the first place [15].

The spectral characteristics of erector spinae at T10 and L3 levels, latissimus dorsi, external and internal obliques, rectus abdominis and pectoralis bilaterally in axial rotation and combined motions (flexion-rotation and extension-rotation), respectively, have been reported [2,3,13]. The phenomenon of differential fatigue in the above-mentioned 14 trunk muscles in axial rotation has been reported by Kumar and Narayan [2] and Kumar et al. [3]. It was suggested that such a differential fatigue could destabilize the spine as a result of differential decay of forces in some muscles, causing sudden jerky motions precipitating injuries to connective tissues.

The torque and EMG of six muscles in dynamic torque generation was measured with a view to quantitatively explain the phenomenon of trunk rotation [7]. Despite careful and rigorous normalization of EMG by endeavoring to elicit maximal neural drive from each muscle in nonrotational activities, the results were far from conclusive. Other studies described the EMG of 10 trunk muscles with varying postural asymmetry and different angular acceleration [5,6]. These authors reported 50% of maximal EMG activity with minimal torque. Those muscles, which had greatest mechanical advantage in causing trunk rotation, showed the greatest increase in their activities as the trunk acceleration increased.

Despite the foregoing studies, the torque-producing behavior of the spine with increasing velocity of rotation and corresponding EMG is not known. Furthermore, the rate of muscle activity and relationship between torque and EMG in active thoracolumbar rotation is not known. This information will be helpful in designing human activities with consideration of the maximal voluntary contraction capabilities in an effort to avoid injury to the back. The current study was designed to address the foregoing issues.

Materials and methods

Experimental sample

Fifty normal young adults (27 men with mean age, weight and height of 21.1 years, 72.3 kg and 176.9 cm; 23 women with mean age, weight and height of 22.1 years, 58.5 kg and 165.4 cm) volunteered for this study. Low-back pain within the past year requiring 1-week vocational absence; other musculoskeletal, neuromuscular disorders and spinal and abdominal surgery were used as exclusion criteria. Ethics approval was granted, and informed consent was obtained from all subjects.

Tasks

These subjects were asked to assume their neutral spinal posture in upright sitting while appropriately stabilized in the axial rotation tester. After staying in this posture for a few seconds, the subjects were asked to exert their isometric MVC to left and right, and isokinetic MVC in left and right directions to and from neutral posture at 10, 20 and 40 degrees per second velocity in a random order. After each exertion, the subjects were allowed a minimum of 2 minutes of rest before the next experimental condition.

Equipment

Axial rotation tester

The Axial Rotation Tester (Fig. 1) was designed to study the coupled motions of axial rotation and lateral flexion with minimal flexion/extension at hip or in torso [1–3,12,13,15,18]. It did not restrict movement of the shoulder or affect the shape changes of the thorax. The Axial Rotation tester consisted of a rigid metal frame mounted on a metal base plate. Inside the frame, mounted on the base plate, was an adjustable chair that could be slid back and forth and adjusted vertically. It had velcro belts to stabilize the seated subjects. The backrest of the chair was sawed off to allow freedom to rotate. Directly above the chair, supported by a long bar, was an adjustable shoulder harness mounted on a circular plate. This plate, in turn, was attached to a spring-loaded rod sliding within a sleeve with a locking screw to position it rigidly at any chosen position. The rod could rotate when the positioned subject underwent axial rotation. This rotation was measured by a high-precision potentiometer attached to the rotating rod with a set of gear (Fig. 1).
Torque measuring system

The force exerted during the isokinetic maximal axial rotation was measured by an Intechnology load cell (I number 500) (Interface, Scottsdale, AZ) and a force monitor equipped with signal conditioner (Prototype Design, Ltd., Ann Arbor, MI). The circular plate of the axial rotation tester was attached to the static and dynamic strength tester [1–3,12–15,18–20] by means of airplane steel cable with an intervening load cell. The static and dynamic strength tester was a variable and controlled constant velocity (isokinetic) strength testing device [1–3,12–15,18].

EMG system

The EMG system consisted of surface electrodes, electrode cables, preamplifiers and amplifiers. Silver-silver chloride surface electrodes of 1 cm diameter with recessed pre-gelled elements (HP 144445) were used with interelectrode distance of 2 cm. These electrodes were connected to fully isolated preamplifiers by means of short cables and tip plugs. The low noise and low nonlinearity preamplifiers with a common mode rejection ration of 130 dB and a wide bandwidth were specially made. The preamplifiers fed to a low-power, high-accuracy instrumentation amplifier designed for signal conditioning and amplification (Measurement, Inc., Ann Arbor, MI). The amplifier had alternating current (AC) coupled inputs with single pole resistance capacitance (RC) filter with a cutoff frequency of 8 Hz to filter out motion artifacts.

Experimental procedures

Fourteen pairs of disposable pre-gelled surface electrodes (HP 144445) were applied to the subjects at an inter-electrode distance of 2 cm after suitable preparation of the skin with an alcohol-acetone mixture. These electrodes were placed on erector spinae leveled with spinous processes of T10 and L3 vertebrae bilaterally 4 cm lateral to the tips of the spinous processes. Surface electrodes were also applied to the left and right latissimus dorsi. On the ventral side, surface electrodes were applied bilaterally to the pectoralis major, rectus abdominis, external oblique and the internal oblique in the area of external oblique aponeuroses to minimize cross talk with it. A ground electrode was applied to the anterosuperior iliac spine.

Then the subjects were seated appropriately adjusted, aligned and stabilized in the chair of the axial rotation tester. The circular disc above the shoulder harness was attached to the static and dynamic strength tester by means of an airplane cable with no slack and an intervening load cell. The subjects were then asked to perform torso rotation to both sides, from the neutral posture and to the neutral posture from both pre-rotated positions in a random order, applying force through the shoulders on the harness causing it to rotate at predetermined fixed velocity. The MVCs were performed without jerking. Subjects were instructed to build their maximal force slowly in first second and carry on the contraction at maximal level until they reached the end of their range of motion, at which time the trial was terminated.

Data acquisition

Data were acquired using modular software designed for the project. The subject data were entered and data files created in the computer. At this stage, a random sequence of all
experimental conditions was generated. The sampling rate was set at 1 kHz, and recording duration was set for 7 seconds. One second into the data acquisition window, the computer generated the first beep, signaling the subject to initiate the contraction. After 5 seconds from the first beep, the computer beeped again, signaling the subject to relax. Data were acquired from 14 channels of EMG, load cell and the potentiometer of the axial rotation tester.

Data analysis

The data were windowed for analysis (Fig. 2). As soon as the torque began to increase from the baseline or returned to baseline within 1% of the maximum torque value, the computer software drew a vertical line marking the start and end of the cycle. A seven-point smoothing routine was used to smooth the EMG signals to reliably interpret the pattern. Using the start position as the point of reference, the onset and all other timings for the muscles were measured. The times of anticipation and onset of all 14 EMG channels with respect to the torque were determined. For this determination, each channel of EMG was plotted with the torque trace and task cycle duration individually (Fig. 2). The relevant sections were amplified in case of any ambiguity. Using a cursor, the points of beginning of anticipation and a clear onset were marked and stored in the memory. Using the start time previously determined, the specific times of all these events were obtained for deciphering phasic interrelationship.

Using the software, these points were joined to delineate the segments of anticipation and onset. From these lines, the slopes of the curves were calculated for these segments. A linear envelope detection of all EMG channels was performed from which the maximum EMG score of all channels, the time of peak EMG, the average EMG, the EMG per second and the EMG area of each of the channels of each of the activities were measured. Finally, each activity was divided in segments of 10% of the task cycle and the magnitude of EMG and time product; the amplitude of EMG at that percent of task cycle and the EMG area of that channel was also calculated using this analysis software.

A statistical analysis of these EMG parameters was carried out using SPSS software (SPSS Inc., Chicago, IL) to calculate descriptive statistics, analysis of variance and Pearson’s product-moment correlation between torque and EMG. A regression analysis was also carried out to determine the predictability of the torque from EMG measurements.
Results

Peak torque and total peak EMG

The peak torque, the EMG normalized torque and the normalized peak EMG are presented in Fig. 3. The peak torques recorded in isokinetic conditions were significantly lower than isometric values (p<.01). Furthermore, with increasing velocity of motion there was a progressive decrease in the peak torque value (Fig. 3). In efforts from neutral posture to either side at 10 degrees per second isokinetic motion, the torque decreased by approximately 28% to 36% compared with the isometric values. With increasing velocity from 10 degrees per second to 20 and 40 degrees per second, velocities in activities away from neutral posture, the peak torque decreased by approximately 1 Nm for every degree of increase in the velocity up to 20 degrees per second and an additional 0.5 Nm between 20 and 40 degrees per second. It is noteworthy that at 10 degrees per second velocity of rotation away from the neutral position in either direction, although the torque decreased by approximately 30%, the EMG per unit torque increased by 10% to 28%. With further increase in velocity of angular rotation (40 degrees per second), a decrease in absolute EMG value occurred, but per unit torque the EMG was higher than that of the isometric condition.

In rotation toward the neutral posture with slowest velocity the peak torques were approximately 10 Nm higher than those recorded in rotations away from neutral posture. With every successive increase of angular velocity, there was a decrement in torque of approximately 20 Nm (Fig. 3). Also,
the EMGs in “towards the neutral” efforts were significantly less from the corresponding rotations away from the neutral. In percentage terms compared with the corresponding isometric neutral posture rotations, with 10 degrees per second angular velocity the EMG ranged between 83% and 98%, with 20 degrees per second velocity it ranged between 72% and 79%, and with 40 degrees per second it ranged between 64% and 70% in left to neutral effort. Thus, in rotations toward the neutral an increase in torque was associated with decrease in EMG, and the increasing angular velocity of rotation was accompanied with decrease in total EMG.

**EMG magnitudes**

The normalized peak magnitudes of EMG of different muscles with increasing angular velocities for male and female samples for rotations from left to neutral and neutral to left are presented in Fig. 4. In both males and females, in isokinetic axial rotation from neutral to left or to right, the contralateral external oblique and ipsilateral latissimus dorsi and internal obliques acted as primary agonists. The contralateral pectoralis also registered high peak EMG activity. In addition to these muscles, the ipsilateral erector spinae had considerably elevated value in comparison to the contralateral side especially in the thoracic region. With increasing angular velocity, there was a small but consistent decrease in peak EMG activities of agonist muscles. By comparison, the peak EMG activities of erector spinae decreased very little and maintained nearly a consistent level of contraction. The EMG pattern and values for contractions from right to neutral and neutral to right were mirror images of contractions from left to neutral and neutral to left; hence, they were not presented.

In axial rotation when subjects returned from prerotated posture to the neutral position with varying isokinetic velocities, the contralateral external oblique and ipsilateral latissimus dorsi and internal obliques with respect to motion direction acted as agonists. These were assisted by the contralateral pectoralis. Here also the contralateral erector spinae at both thoracic and lumbar levels demonstrated considerably increased EMG activity. In these activities and with increasing angular velocity of rotation there was a progressive decrease in peak EMG activity. Here the decrement in EMG activity of erector spinae was also substantial. The average EMG and the EMG area (magnitude and time product) revealed similar patterns as peak EMG for all axial rotational activities (Fig. 5). When the task cycle was divided into 10 equal segments and the EMG area of each segment of each channel was examined, it revealed a consistent pattern. Each muscle up to 20% of the task cycle and beyond 80% of the task cycle produced 5% to 7% of the total EMG area of its respective muscle. Between 20% and 80% of the task cycle with 10 degrees per second angular velocity, each segment represented between 10% and 13%. With increasing velocity, this middle section shrank in duration but in magnitude increased up to 15%.

The peak EMG values expressed as percentage of their...
representative isometric axial rotation for all agonists and
most antagonists were higher in rotation away from the neu-
tral. These values ranged between 95% and 111% except
rectus abdominis for 10 degrees per second angular veloc-
ity. With increasing velocity, there was a decreasing per-
centage value for peak EMG activity of all muscles. When
axial rotation was performed toward the neutral posture
from a prerotated position, the percentage values were con-
siderably less than “away from neutral posture efforts.”
Also, the percentage magnitude of the EMG activity with
increasing velocities decreased much faster than that ob-
tained in activities toward the neutral.

**Slope of EMG activities**

The slopes of the onset of EMG activities in axial rotation
away from the neutral posture in both directions as well as to-
ward the neutral posture from prerotated postures had similar
patterns. The highest slope values were recorded from the con-
tralateral external obliques followed by the ipsilateral latissi-
mus dorsi, contralateral pectoralis, ipsilateral thoracic erector
spinae, lumbar erector spinae and ipsilateral internal obliques
in men. Among women, the steepest slope was for ipsilateral
latissimus dorsi, followed by contralateral pectoralis, contralat-
eral external oblique and ipsilateral internal oblique. Also there
was a pronounced increase in slopes of all muscles with in-
creasing angular velocity of axial rotation. When slope values
were normalized against the increase of slopes for different
muscles in corresponding isometric axial rotation, the slopes
for isokinetic rotations were 200% to 500% greater. The slopes
continued to increase rapidly with increasing angular velocity.

**Phasic relationships**

Except contralateral erector spinae and sometimes ipsi-
lateral pectoralis, all muscles began their activity before the
onset of torque. Generally, there was no repeatable and con-
sistent time relationship between different muscles in rota-
tions toward and away from neutral posture. Many times
there were no sizeable differences between right and left
muscle pairs. However, with increasing velocities of rota-
tion, the onset of most to all muscles occurred progressively
sooner, increasing the time gap between the onset of muscle
activity and the onset of the torque.

Fig. 5. Neutral to left. The decrement in electromyograph area with increasing velocity of the isokinetic trunk axial rotation. LEO = left external oblique; LIO = left internal oblique; LLD = left latissimus dorsi; LL3 = left lumbar erector spinae at L3 level; LP = left pectoralis; LRA = left rectus abdominus; LT10 = left thoracic erector spinae at T10 level; REO = right external oblique; RIO = right internal oblique; RLD = right latissimus dorsi; RL3 = right lumbar erector
spinae at L3 level; RP = right pectoralis; RRA = right rectus abdominis; RT10 = right thoracic erector spinae at T10 level.
The relationship between torque and EMG demonstrated an interesting pattern, especially when compared with isometric condition. A transition from isometric to a very slow and controlled isokinetic velocity of motion resulted in a 40% decrease in the torque and a 10% to 28% increase in EMG. It would appear that for any linked biological segments, any forceful motion is of considerable threat to safety and comes at a significant cost in force generation. Perhaps it is the result of the torso structure where the passive elements (connective tissue) act as resistance in any motion away from the neutral. From the observed data in this experiment, even the slowest motion generates twice as much EMG output (an index of muscle work). Between 10 and 20 degrees per second motions, the ratio of torque to EMG remained the same as for 10 degrees per second. However, with further increase in velocity to 40 degrees per second, the torque decreased to 40% and EMG remained at approximately 100%. This clearly means that with increasing velocity of forceful rotary motion, more and more muscle force is absorbed in deforming connective tissues to allow rotation and provide stability to the spinal joints, hence lower torque values. An increasing velocity of motion is clearly a greater risk to the safety of the viscoelastic soft tissue, which is strain rate dependent. Furthermore, while executing the motion, a decrement in velocity is not achieved by morphological restrictions until one reaches close to the end of the range of motion. Because the range of motion in axial rotation is small, it is perhaps for this reason that the ability of humans to develop torque in axial rotation is considerably lower, compared with sagittally symmetrical activities [1,12,14].

McGill [7] normalized asymmetric activity against sagittally symmetrical activities, and he reported some of the limitations of this method. Even the muscles that are ideally oriented for rotational activities generated greater EMG in symmetrical activities compared with the rotational activities. It may appear that the functions of muscles may have a variable relationship between torque generation and provision of stability. The quantitative nature of this varying relationship is not known or understood. Thus, normalization of asymmetric activities against sagittal plane reference activity may not be appropriate, because a bilateral recruitment of even rotary muscles may generate forces in the sagittal plane. It is unclear if such a position is mechanically more stable, thus allowing most of the muscle activity to be directed toward torque generation. In rotary motions the anatomical or behavioral end point is less well defined because of interplay between multiple joints. Relatively small antagonistic activity during the execution of an asymmetric motion may be an built-in mechanism to further impede rotation (which is also opposed by the connective tissues) and to prevent it from becoming very strong. Perhaps it is for this reason that the maximal torque in axial rotation was found to be approximately 25% of the torque generated in extension [1,14]. Marras and Mirka [4,5] also reported that 50% of maximal EMG was generated with minimal torque in rotational activity. Thus, it may be that in certain types of asymmetric motion the coupling of torque and EMG may be controlled and regulated differently and be at a different ratio from that observed in sagittal motion. Unusual low correlation and poor regression observed in this study are consistent with this logic. These reasons would lead one to deduce that an activity-relevant normalization may perhaps be more valid when considering normalization of

### Statistical results

The peak and average EMG amplitudes and the EMG area under the curve showed identical results. For all of them, the magnitude of EMG was significantly affected by the muscle, and angular velocity, the direction of rotation and starting position (p<.001; Table 1). The gender did not have significant effect when agonists were compared with agonists and antagonists were compared with antagonists (Table 1). In addition to these main effects, there were two-way and three-way significant interactions (Table 1). The two-way significant interactions were between muscle on the one hand and gender and the direction of rotation on the other (p<.001). This meant that the muscles generated different amplitudes in different starting angles and the direction of rotation. All individual muscles except the right rectus abdominis had significantly different scores for rotations in different directions (p<.02).

The correlations between EMG and torque, both peak and average, were weak but significant (p<.01). These correlations ranged between r=.184 and r=.194. However, when torques were correlated with EMGs of individual muscles, the correlation coefficients ranged between r=.21 and r=.41 (p<.01). Thus, only low correlation values were observed throughout. Similarly, significant linear regressions were also obtained for predicting peak and average torques from EMG parameters (p<.001). However, only 3.6% to 5.9% of variance in torque values could be explained by these variables. In a forward stepwise regression, 28.6% to 30.8% of variances could be explained by selecting from all EMG parameters for average and peak torques, respectively.

### Discussion

The relationship between torque and EMG demonstrated an interesting pattern, especially when compared with isometric condition. A transition from isometric to a very slow and controlled isokinetic velocity of motion resulted in a 40% decrease in the torque and a 10% to 28% increase in EMG. It would appear that for any linked biological segments, any forceful motion is of considerable threat to safety and comes at a significant cost in force generation. Perhaps it is the result of the torso structure where the passive elements (connective tissue) act as resistance in any motion away from the neutral. From the observed data in this experiment, even the slowest motion generates twice as much EMG output (an index of muscle work). Between 10 and 20 degrees per second motions, the ratio of torque to EMG remained the same as for 10 degrees per second. However, with further increase in velocity to 40 degrees per second, the torque decreased to 40% and EMG remained at approximately 100%. This clearly means that with increasing velocity of forceful rotary motion, more and more muscle force is absorbed in deforming connective tissues to allow rotation and provide stability to the spinal joints, hence lower torque values. An increasing velocity of motion is clearly a greater risk to the safety of the viscoelastic soft tissue, which is strain rate dependent. Furthermore, while executing the motion, a decrement in velocity is not achieved by morphological restrictions until one reaches close to the end of the range of motion. Because the range of motion in axial rotation is small, it is perhaps for this reason that the ability of humans to develop torque in axial rotation is considerably lower, compared with sagittally symmetrical activities [1,12,14].

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### Table 1

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NS = not significant.
EMG for any activity. Such a conclusion is supported with the findings of this study. Also, the relative amount of EMG in relation to torque continued increasing with increasing velocity of axial rotation. Such a pattern of strong interrelationship was clear in this study, which investigated only modest angular velocity of activities (maximum 40 degrees per second). Because of the progressive nature of this relationship within the experimental parameters of this study, it will be reasonable to speculate that a further increase in angular velocity of the motion will further widen this gap, obscuring the torque–EMG relationship to an even greater extent. The deformation of several spinal ligaments and joint capsules, all of which display viscoelastic properties, at progressively faster angular velocity of rotation will offer increasing resistance to rotation, hence a poorer correlation and regression between EMG and torque.

The foregoing logic supports the results of the current study, which shows that in asymmetric motion, from a pre-rotated position to the neutral posture, there was significantly higher torque and a correspondingly decreased EMG when compared with motion in the opposite direction. The torque generation would have been significantly assisted by release of stored mechanical energy in the deformed spinal connective tissues. It is also suggested here that motion away from the neutral posture involves concentric contraction of agonistic muscles. Thus, with progressive shortening of the muscle length, the torque begins to decrease and the stretched connective tissues offer progressive resistance to motion with increasing deformation. Therefore, in order to register increased torque, these agonistic muscles will have to further stretch the already stretched viscoelastic connective tissues. Toward the higher levels of deformation, the elastic stiffness increases on top of the viscous resistance. Such a combination of mechanisms causes considerably greater effort to realize even a small increase in the torque value. However, when people rotate toward the neutral posture from an already pre-rotated posture, the mechanical energy stored in the passive connective tissues is released, which assists the torque magnitude on top of what is being generated by the muscles. These observations are in agreement with literature [15].

The decrease in EMG magnitude with increase in angular velocity of rotation was in absolute terms only. When compensated for reduction in the torque, the ratio of EMG to torque increased with increasing velocity. This phenomenon would also indicate that perhaps there is a complex linking between the torque, velocity and EMG, which may be effective in maintaining the safety of a joint by keeping the resultant from increasing beyond the margin of safety [21]. The observation in this study that the magnitude of EMG of erector spinae remained relatively unchanged gives credence to the concept that such an adaptation may be guided by principles of safety. Those muscles not positioned to provide rotary torque in axial rotation appear to be providing stability to the spinal joints. Thus, the relationship between torque and EMG may be modulated by the safety requirement of the system.

**Conclusion**

The study proved that there was an inverse relationship between the axial rotation torque and the velocity of rotation. Also, there was a progressive increase in the EMG per unit torque with increasing velocity of rotation, thereby obscuring the EMG–torque relationship. In trunk rotation, the erector spinae likely plays the role of stabilization. The significance of the findings is that rotational jobs can be designed to control rotational back injuries, based on the quantitative data presented.

**References**