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Original article

Trunk muscle response to various protocols of lumbar traction

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ABSTRACT

The purpose of this study was to compare trunk muscle activity, spinal decompression force, and trunk flexibility resulting from various protocols of spinal traction. Four experiments explored the effects of (1) sinusoidal, triangular, square, and continuous distraction-force waveforms, (2) 0, 10, 20, and 30 degrees of pull angle, (3) superimposed low, medium and high frequency force oscillations, and (4) sham traction. Nineteen healthy subjects volunteered for this study. Surface EMG was recorded during traction and later used in a biomechanical model to estimate spine decompression force. Trunk flexibility was measured before and after each treatment. There were no differences in muscle activity between any of the experimental conditions except the thoracic erector spinae muscle, which had lower EMG during continuous compared to sinusoidal distraction-force waveform ($p = 0.02$). Thoracic and lumbar erector spinae muscles were significantly less active during sham than real traction ($p = 0.01$ and $p = 0.04$, respectively). The estimated L4–L5 spine compression force was 25 N. Trunk flexibility decreased after each experimental session ($p = 0.01$), and there were no differences between sessions. Our results suggest that the trunk muscle activity is minimal and point toward fluid exchange in the disc as one of the key biomechanical effects of spinal traction.

1. Introduction

With low back pain (LBP) remaining one of the most prevalent and costly health problems in Western Society (Andersson, 1999), the search continues for an effective treatment. Because spinal surgery is expensive and not always effective, the management of LBP begins usually with a conservative approach. One such conservative approach is mechanical spinal traction. This type of treatment relies on the application of a continuous or intermittent distraction-force between the pelvis and ribcage. Over 30% of physical therapists surveyed in Ontario, Canada, used spinal traction as the preferred treatment for subacute LBP and acute LBP with sciatica (Li and Bombardier, 2001), which represents the trends in North America. Similarly, lumbar traction is frequently used in the UK despite numerous recommendations suggesting it is ineffective (Harte et al., 2003). These recommendations, based on comprehensive reviews of randomized clinical trials, state that lumbar traction cannot be recommended as a single therapy for LBP with or without sciatica (Harte et al., 2003; Airaksinen et al., 2006; van Tulder et al., 2006a,b; Clarke et al., 2007). However, these reviews also state that the literature does not allow for a firm negative conclusion to be made due to the small number of high quality studies published. Most of the studies had too few subjects, mixed patient population, and other methodological flaws.

The exact mechanism through which traction might be effective is not known. It has been suggested that spinal elongation, by increasing intervertebral space, inhibits nociceptive nerve activity, improves mobility, reduces muscle spasm, relieves nerve root compression, and lessens adhesions around the facet joints. None of these mechanisms have been supported sufficiently by empirical data (van der Heijden et al., 1995; Clarke et al., 2007). However, all of these possible mechanisms depend on adequate distraction-force being transmitted directly to lumbar segments. During traction, muscle tension and friction between the body and the support surface should be taken into account in the form of counterforces (van der Heijden et al., 1995). While the counteractive friction force can be eliminated with various technological solutions, such as a split and sliding table, the effects of trunk muscle response to lumbar traction are unknown (van der Heijden et al., 1995; Krause et al., 2000; Clarke et al., 2007). Two previous studies looked only at EMG of sacrospinalis muscles (Hood et al., 1981; Letchuman and Deusinger, 1993). Thus, relaxation of spinal muscles appears to be

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the most important prerequisite for spinal traction to be mechanically effective.

The most recent developments in spinal traction involve new technologies that allow for varying angles of pull, varying load duty cycles; waveforms; their frequency; and concurrent application of superimposed oscillations (Shealy et al., 2005). Such a treatment, named Intervertebral Differential Dynamic (IDD®) therapy, claims to be more effective in treating patients with LBP than a standard traction technique (Shealy et al., 2005). However, further refinement of IDD therapy requires quantification of trunk muscle activity and the resultant spinal loads under various waveforms, angles of pull, and oscillations. Currently, no studies comparing trunk muscle response to these protocols exist. Therefore, the purpose of this study was to compare trunk muscle activity, spinal decompression force, and trunk flexibility resulting from various protocols available with the Accu-Spina device (North American Medical Corporation, Marietta, GA) used for IDD therapy. The premarket approval for this device was granted by the FDA in 2005 (510(k) #K033231).

2. Methods

2.1. Study design

The entire study consisted of four separate experiments, each exploring changes in trunk muscle activity, spinal decompression force, and trunk flexibility during various treatment options available with the Accu-SPINA device (Fig. 1). Each experiment lasted between 24 and 28 min, per manufacturer's recommendations, and contained all experimental conditions presented in random order:

(1) The effects of various distraction-force waveforms (sinusoidal, triangular, square, and continuous). The angle of pull was kept at 10°.
(2) The effects of various angle of pull (at 0, 10, 20, and 30°) using sinusoidal distraction-force waveform.
(3) The effect of force oscillations (low, medium and high frequency) superimposed on the square distraction-force waveform. The angle of pull was kept at 10°.
(4) The effects of sham traction consisting of lying supine without any distraction-force.

It should be noted that it was not possible to investigate all of the independent variables in one experiment because we did not want to expose the subjects to traction longer than the recommended 30-min limit.

The difference in sit-and-reach tests performed before and after each experiment served as an indicator of possible changes in the fluid content of intervertebral discs. Because in addition to hip and hamstring, this test also measures low back flexibility; and because the range of motion of the back (i.e. modified Schröber test) reflects diurnal changes in disc hydration, (Wing et al., 1992), we included the sit-and-reach test as one of the outcome measures. Trunk muscle activity was monitored with surface EMG, which was later used in an EMG-assisted spine model to estimate net forces acting on the osteoligamentous spine during traction.

Fifteen subjects were tested in each experiment. However, most of the subjects volunteered for more than one experiment and were thus tested multiple times on separate days. In total, 13 males and 6 females, each without a history of LBP, were recruited for all experiments. On average (standard deviation) they were 26.4(6.2) years old, 1.76(0.10) m tall, and weighed 74.3(13.3) kg. All subjects read and signed an informed consent form prior to testing. The protocol for this study was approved by Yale University's Human Investigation Committee.

2.2. Procedures

Prior to traction treatment, all subject performed three trials in a sit-and-reach flexibility test according to a standard protocol (Allen, 1988). This protocol involved sitting on the floor with straight legs braced against a box. With palms facing down, the subject reaches forward along the measuring line on the box as far as possible. The maximum reach was held for 3 s and all three trials were averaged to obtain a flexibility score. The flexibility test was repeated at the end of each traction experiment.

After appropriate skin preparation, Ag–AgCl, bipolar, disposable surface EMG electrodes were placed over the following muscles on

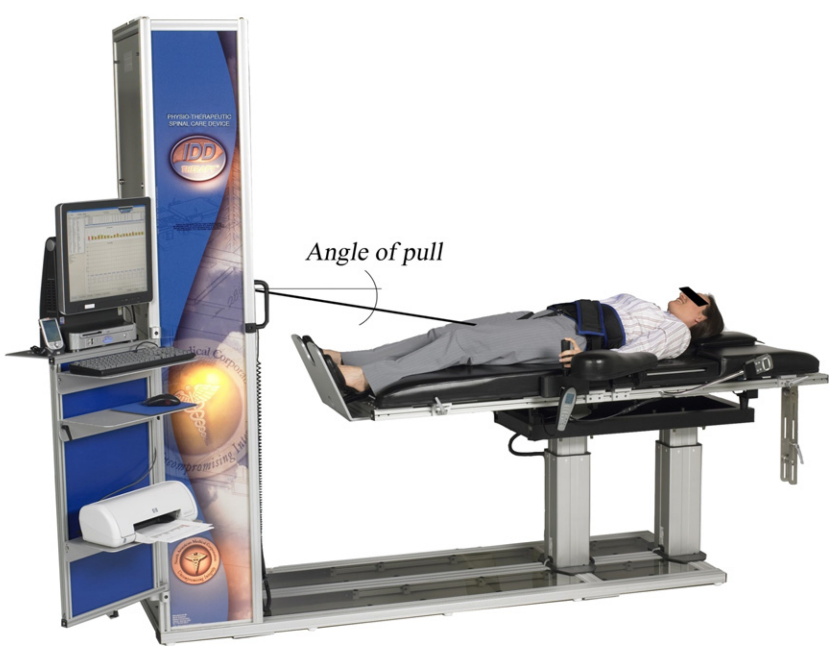

Fig. 1. The Accu-SPINA device used in this study (North American Medical Corporation, Marietta, GA). The table was split, such that the lower body of a subject moved with the bottom part of the table on linear bearings during traction.

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the right side of the body: rectus abdominis (RA, 3 cm lateral to the umbilicus), external oblique (EO, medial to the mid auxiliary line at the level of the umbilicus), internal oblique (IO, approximately midway between the anterior superior iliac spine and symphysis pubis, above the inguinal ligament), latissimus dorsi (LD, lateral to T9 over the muscle belly), thoracic erector spinae (TE, 5 cm lateral to T9 spinous process), and lumbar erector spinae (LE, 3 cm lateral to L4 spinous process) (Cholewicki and McGill, 1996). Each pair of electrodes was spaced 3 cm center-to-center along the muscle belly. A reference electrode was placed over the 10th rib on the right side. After verifying the quality of EMG signals on an oscilloscope, subjects performed maximum isometric exertions in trunk flexion, extension, and lateral bending on an examination table against the resistance provided manually by one of the investigators. These tasks were designed to elicit maximum voluntary activation (MVA) levels from trunk muscles, for the purpose of EMG normalization (McGill, 1991). For the abdominal muscles, an exertion in a sit-up position was modified from McGill (1991) in that the subjects produced a sequence of maximal efforts in trunk flexion as well as trunk flexion with superimposed left and right torso twists.

Next, subjects donned chest and pelvic harnesses and lay supine on the Accu-SPINA table (Fig. 1). The chest harness was affixed to the immovable part of the table, while the pelvic harness was attached to the motorized traction assembly. This assembly moved up or down for adjusting the angle of pull, which was verified with an inclinometer. At this point, 3 s of EMG data were recorded while subjects lay fully relaxed to obtain a baseline EMG value.

The exact shapes of all force waveforms applied are presented in Fig. 2. According to the manufacturer's recommendations, the peak force was set at half body weight plus 44.5 N (10 lb), while the low force was set at half of the peak value. Each traction experiment began with a 60 s ramp-up to the peak force followed by two cycles of a given force waveform application. The bottom part of the split table was then released to slide freely on linear bearings. This release reduced the friction between the person and the table and allowed the distraction-force to be transmitted to the trunk. The release also marked the beginning of the treatment, which consisted of three cycles of each experimental condition applied consecutively. The EMG data and the distraction-force were recorded with the same data acquisition board on the third cycle of each condition using 1 kHz analog-to-digital conversion (A/D). A 60 s ramp-down concluded each condition. In the sham experiment, EMG data were collected every 5 min. Prior to the A/D conversion, the EMG signals were band-pass limited between 20 and 450 Hz and differentially amplified (input impedance = 100 GΩ, CMRR > 140 dB).

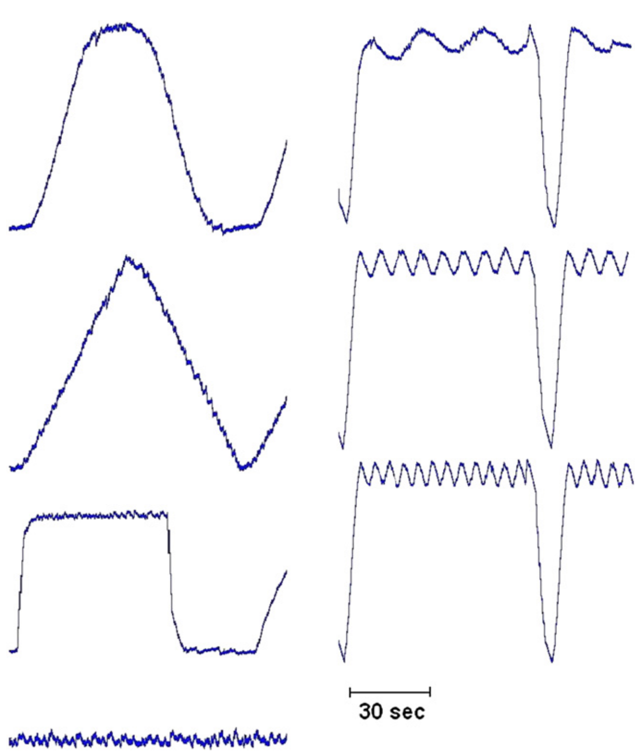

Fig. 2. Various waveforms of distraction-forces applied via Accu-SPINA device (left panel). Low, medium and high frequency oscillations are presented in the right panel.

2.3. Data analysis

Mean absolute values of EMG signals were computed between heart beats (QRS waves) in epochs corresponding to the peaks and troughs of the force waveforms. The data were examined for normality using the Anderson–Darling test and corrected with the Box–Cox transformation prior to the statistical analyses, if they were not normally distributed. Repeated measures ANOVAs and Tukey's post hoc tests ($p \leq 0.05$) were used to evaluate differences in muscle activities. First, the comparison was made between EMG corresponding to peaks and troughs of the distraction-force. Next, EMG data corresponding to peak force were compared between all experimental conditions in the first three experiments (various waveforms, angle of pull, and oscillations). Finally, we compared the sham and real traction using the EMG collected during the last time point for the sham and the last experimental condition from experiment 1 (various waveforms). Because the data for this comparison came from different testing sessions, we normalized the EMG using the baseline EMG value obtained from the relaxed lying condition. Because these data were not normally distributed, even after the transformation, a non-parametric Kruskal–Wallis test was used. A nested repeated measures (subjects nested within each experiment) ANOVA was used to compare sit-and-reach flexibility before and after each experiment. Before and after condition served as a within-subjects factor and four experiments constituted a between-subjects factor. All analyses were performed using the Minitab statistical software (Minitab Inc., State College, PA). All data were presented as % MVA.

The net decompression force transmitted to the osteoligamentous spine was computed as the difference between the sum of all trunk muscle forces and the distraction-force applied to the trunk by the Accu-SPINA device. Muscle forces were estimated based on the level of their EMG activation using the biomechanical model of a lumbar spine system. A detailed description of this model has been previously published (Cholewicki and McGill, 1996). It consists of a rigid pelvis and sacrum, five lumbar vertebrae separated by a lumped parameter disc and ligament equivalent, rigid ribcage and 90 muscle fascicles. Each muscle consists of an active contractile part, a passive parallel elastic element and a passive nonlinear tendon. Forces in all 90 muscle fascicles were calculated with the help of EMG and the cross-bridge bond distribution moment approach (Cholewicki and McGill, 1995). As in the original work, assumptions were made regarding the neural activation of deep muscles not accessible via surface EMG. Psoas and quadratus lumborum were driven with the EMG signals of their synergists (IO and LE, respectively). Left/right muscle activation symmetry was also assumed.

3. Results

There were no differences between EMG activity corresponding to the peaks and troughs of the distraction-force in any of the six muscles tested ($p > 0.50$, DF = 1, $F < 0.5$). Therefore, only peak force EMG was used for subsequent analyses.

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Within the three traction experiments, no effects of angle of pull ($p > 0.06$, DF = 3, $F < 2.6$) or superimposed oscillations ($p > 0.36$, DF = 2, $F < 1.1$) were found in any of the six trunk muscles. With respect to waveform, however, a significantly lower EMG activity was present in the TE muscle during constant compared to a sinusoidal distraction-force waveform (ANOVA: $p = 0.02$, DF = 3, $F = 3.6$; Tukey's post hoc: $p = 0.02$, $T = 3.0$) (Fig. 3).

A comparison between sham and real traction was made using EMG collected at the end of the sham traction and the EMG obtained from the last waveform tested in experiment 1, which gave similar duration of treatment in both cases. Both TE and LE were significantly less active during sham than during real traction ($p = 0.01$, DF = 3, $H = 6.2$ and $p = 0.04$, DF = 3, $H = 4.1$, respectively) (Table 1).

To compute spine decompression force, the counter force (spine compression force) stemming from the activity of all trunk muscles was estimated with a biomechanical model. Because overall muscle activity was very low with little differences between various experimental conditions, two representative cases were considered: sham and sinusoidal traction. The input to the model consisted of the across-subjects average EMG data expressed as % MVA (Table 1). The L4–L5 spine compression force was 218 N for sham and 434 N for the sinusoidal waveform traction. Considering that on average 409 N of peak distraction-force was applied, the spine was decompressed to 25 N during the sinusoidal waveform traction.

Trunk flexibility decreased after all of the four experimental sessions (main effect: $p = 0.01$, DF = 1, $F = 7.2$). There was no significant interaction between the sessions and flexibility ($p = 0.90$, DF = 3, $F = 0.2$), suggesting that flexibility decreased similarly after each session. On average, subjects lost 6 (SD = 2) mm in their reach during a traction or sham session.

4. Discussion

The main finding of this study was that the overall trunk muscle activity is very low during traction and varies very little between different protocols of applying distraction-force in healthy subjects. For example, the average overall activity during sinusoidal waveform traction was 0.65% MVA. As expected, this value is lower than 1.7% MVA reported during upright standing (Cholewicki et al., 1997), because the demands on spine stability are lower when lying as compared to standing postures. These results agree with the only two previous studies that looked at EMG activity of sacrospinalis muscle. Hood et al. (1981) found no difference in EMG in healthy subjects between lying supine on a table and applying traction.

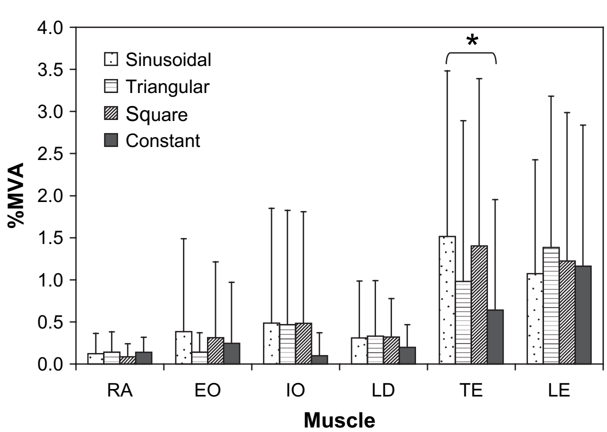

Fig. 3. Comparison of trunk muscle activities (mean (SD)) during traction using various force waveforms. An asterisk indicates significant difference ($p < 0.05$).

Table 1
Average trunk muscle activity (% MVA, mean (SD)) during sham and traction

	RA	EO	IO	LD	TE*	LE*
Sham	0.14 (0.32)	0.07 (0.12)	0.73 (2.41)	0.01 (0.03)	0.08 (0.31)	0.13 (0.39)
Traction	0.18 (0.20)	0.29 (0.90)	0.17 (0.30)	0.27 (0.22)	1.06 (1.65)	0.84 (1.06)

*, Significant difference between two conditions ($p < 0.05$).

Letchuman and Deusinger (1993) recorded approximately 4% MVA of EMG activity in patients with LBP during traction, but there is always a doubt whether these patients were able to produce true maximum voluntary contractions. Both of these studies recorded higher EMG during the initial traction cycle. After approximately 4–6 min, this activity returned to baseline (Hood et al., 1981; Letchuman and Deusinger, 1993). Because we pre-conditioned the subjects before data collection with a 60 s ramp-up and two cycles (4 min total), we did not find any differences in EMG activity between cycles during the subsequent treatment part.

Both studies found less sacrospinalis activity during continuous traction than during intermittent traction, although these differences were not statistically significant (Hood et al., 1981; Letchuman and Deusinger, 1993). These results are again consistent with our finding of significantly lower TE activity during continuous traction compared to the traction with a sinusoidal waveform. No other differences between waveforms, angle of pull, or superimposed oscillations existed in our study. Any possible cumulative effects of EMG responses were circumvented by randomizing the order of conditions tested within each experiment.

It is quite likely that patients with LBP would demonstrate different muscle response to traction and this should be the focus of a future study. Patients with LBP demonstrate trunk muscle recruitment patterns that enhance spine stiffness, including greater antagonist co-activation (van Dieën et al., 2003). Therefore, it is also possible that in the face of reduced demands for spine stability during traction, patients would relax their muscle co-activation to some extent. Because prolonged muscle co-activation levels exceeding 5% MVA could lead to muscle fatigue and pain, such relaxation would have a positive result and could be one of the mechanisms by which traction might relieve back pain symptoms. This mechanism was proposed earlier for lumbosacral orthoses (Cholewicki, 2004; Cholewicki et al., 2007).

The estimated spine compression force was only 434 N during sinusoidal waveform traction. This compressive force was comprised of a passive elastic muscle force component and a very low active component, which some may call muscle tonus (Walsh, 1992). Combined with the peak distraction-force of 409 N, the spine was almost completely decompressed during traction. Ramos and Martin (1994) measured negative 100 mmHg pressure in a few patients' discs during the application of approximately a 100 lb (445 N) distraction-force. Taking 1500 mm^2 as a disc's cross-sectional area, this distraction-force would produce −55 mmHg in our experiment ((434 N–445 N)/0.0015 m^2/133 Pa $mmHg^{-1}$). Therefore, both the documented muscle activity and the estimated spine decompression forces appear reasonable in our study.

Despite the relatively short duration (approximately 0.5 h) of each experimental session, a significant loss in trunk flexibility occurred. This was likely due to an increase in disc hydration (Adams et al., 1990; Wing et al., 1992). Such changes increase disc height and decrease flexibility of the lumbar spine (Adams et al., 1990; Wing et al., 1992). These phenomena are well documented as diurnal changes during sleep and are considered an important mechanism for nutrient transport to the intervertebral discs (Grunhagen et al., 2006).

Although there was no difference in flexibility between real and sham traction, the intermittent force application might be more advantageous for maximizing fluid exchange and nutritional transport. This could be another biomechanical effect of spinal traction. If differences in fluid flow exist between various distraction-force waveforms used in the Accu-SPINA device, it is possible that they could be detected with MRI modalities. The short treatment duration and rapid effects of fluid flow in our study should not be surprising, because the greatest increase in hydration of the unloaded disc takes place within the first hour of load removal (Costi et al., 2002).

In summary, our results suggest that overall trunk muscle response to traction does not pose a great problem for mechanically decompressing the intervertebral disc. The significant changes in trunk flexibility point toward fluid exchange as one of the key biomechanical effects of spinal traction, but this study did not address the overall effectiveness of traction as a treatment for LBP.

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