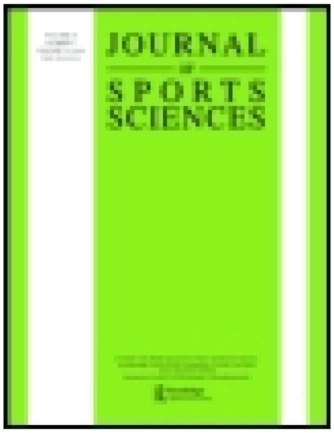

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Muscle activity and spine load during anterior chain whole body linkage exercises: the body saw, hanging leg raise and walkout from a push-up

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Abstract
This study examined anterior chain whole body linkage exercises, namely the body saw, hanging leg raise and walkout from a push-up. Investigation of these exercises focused on which particular muscles were challenged and the magnitude of the resulting spine load. Fourteen males performed the exercises while muscle activity, external force and 3D body segment motion were recorded. A sophisticated and anatomically detailed 3D model used muscle activity and body segment kinematics to estimate muscle force, and thus sensitivity to each individual's choice of motor control for each task. Gradations of muscle activity and spine load characteristics were observed across tasks. On average, the hanging straight leg raise created approximately 3000 N of spine compression while the body saw created less than 2500 N. The hanging straight leg raise created the highest challenge to the abdominal wall (>130% MVC in rectus abdominis, 88% MVC in external oblique). The body saw resulted in almost 140% MVC activation of the serratus anterior. All other exercises produced substantial abdominal challenge, although the body saw did so in the most spine conserving way. These findings, along with consideration of an individual's injury history, training goals and current fitness level, should assist in exercise choice and programme design.

Keywords: *anterior chain exercises, suspension strap training, labile contact surfaces*

Introduction

Exercises involving the full body linkage have been advocated to enhance functional strength. In this sense, "functional strength" involves strength that is created and transmitted through the body linkage. This is in contrast with strength challenges that are focused around a joint, and the forces leave an adjacent or nearby segment via contact with an object such as a bench or chair. This concept is sometimes referred to as open and closed chain exercises (Graham, Gehlsen, & Edwards, 1993). Training with labile systems such as suspension straps has been documented to offer unique opportunities for "open chain" training challenges (for example, Beach, Howarth, & Callaghan, 2008, who documented higher muscle activation levels with labile surfaces). Likewise, exercises performed while hanging from a bar could influence muscle activation and joint loads throughout the body linkage possibly making them more suitable for some people who would benefit from joint sparing approaches. However, more investigation of these exercise approaches is needed to understand their influence on muscle activation and joint load levels.

Many exercises are designed to create three-dimensional joint moments that prevent motion, essentially creating a stiffened and stabilised torso segment. Stiffness, and hence stability, enhances two elements: first, a stiffer spine is more resilient to buckling, allowing it to safely bear more load; and second, proximal stiffness, i.e., stiffness proximal to the shoulder and hip, fixates the proximal muscle attachment, so the mechanical effect is focused on the distal attachment creating faster limb movements with more power in the arms and legs (McGill, 2014). In contrast, other people may not benefit from stiffness for load bearing but rather utilise stiffness as a controller of motion (Brown & McGill, 2009). Consequently, the stiffness is "tuned" to seek the optimum between load bearing ability, power production in the limbs and movement control. To date, there appears to be little quantification of many of these whole body exercises which motivated this study.

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Specifically, this study investigated muscle activation and spine load during the body saw, hanging leg raise and walkout from a push-up. These exercises were chosen as representative of a spectrum of whole body linkage exercises including a three-point bend, distraction tension and a bend with high compression, respectfully. It was hoped that the resulting descriptive data could provide guidance to those designing exercise programmes with objectives to match appropriate exercise progressions with individuals based on muscle activation targets, and spine load and possibly for those with an injury history compromising load bearing ability. Low back loading was a focus of this study given the prevalence of back pain among those who train.

Methods

Participants

Fourteen male participants, mean (SD) age 21.1 years (2.0), height 1.77 m (0.06) and mass 74.6 kg (7.8) recruited from the university population comprised a convenience sample for this study. They were healthy with no previous history of disabling back or musculoskeletal pain. The study was approved by the Office of Human Ethics at the University, and all participants signed an informed consent form.

Instrumentation

Each participant was instrumented with electromyography electrodes monitoring muscle activity together with markers for 3D body segment movement tracking. These data were processed and input to a sophisticated and anatomically detailed 3D model that used muscle activity and body segment kinematics to estimate muscle force. In this way the model was sensitive to the individual choice of motor control selected by each person and for each task. Muscle forces and linked segment joint loads were used to calculate spine loads (Figure 1).

Electromyography (EMG). Fifteen channels of EMG were collected by placing electrode pairs over the following muscles on the right side of the body: rectus abdominis 3 cm lateral to the navel; external oblique approximately 3 cm lateral to the linea semilunaris at the same level as the rectus abdominis electrodes; internal oblique at the level of the anterior superior iliac spine (ASIS) and medial to the linea semilunaris, but superior to the inguinal ligament; latissimus dorsi inferior to the scapula over the muscle belly when the arm was positioned in the shoulder mid-range; upper (thoracic) erector spinae 5 cm lateral to the spinous process of T9; lumbar erector spinae 3 cm lateral to the spinous process of L3; rectus femoris midway between the patella and the ASIS over the belly of the muscle; gluteus maximus approximately 6 cm lateral to the intergluteal cleft; gluteus medius approximately 5 cm lateral to the posterior inferior iliac spine; biceps brachii with the elbow flexed at 90°, two-thirds of the way down the anterior aspect of the arm between the acromion process and the cubital fossa; triceps brachii posterior aspect of the arm at the same level as biceps brachii; anterior deltoid with the shoulder flexed to 90°, approximately 3 cm inferior to the acromion process; upper trapezius midway between the acromion and C7; pectoralis major with the arm abducted and elbow flexed to 90°, midway between the axilla and the areola; serratus anterior with the arm abducted and elbow flexed to 90°, over the attachment to the seventh rib. Note that recording from one side of the body allowed a greater number of muscles to be monitored. Motor control symmetry was assumed between left and right sides of the body. Before the electrodes (Meditrace, Mansfield, MA) were adhered to the skin, the skin was shaved and cleansed with Nuprep™ abrasive skin prepping gel. Ag-AgCl surface electrode pairs were positioned with an inter-electrode distance of approximately 2.5 cm and were oriented in series parallel to the muscle fibres. The EMG signals were amplified and analogue to digital converted with a 16-bit converter at a sample rate of 2160 Hz using the VICON Nexus™ (Los Angeles, CA, USA) motion capture system software. Though multiple muscles were collected, not all were incorporated into the modelling analysis (see *Kinetic and Kinematic Data to predict back loads* below).

Each participant performed a maximal voluntary isometric contraction (MVC) of each muscle for normalisation (Brown & McGill, 2009). These normalisation techniques have been developed over 30 years in our lab to achieve isometric activation in ways that minimise the risk of back injury and muscle avulsion. Dynamic contractions create higher levels of motor unit activity according to known force–velocity relationships – these are incorporated into the modelling approach to estimate muscle force. Specifically, for the abdominal muscles (rectus abdominis, external oblique, internal oblique), participants adopted a sit-up posture with the torso at approximately 45° to the horizontal with the knees and hips flexed at 90°. Manually braced by a research assistant, the participant was instructed to produce a maximal isometric flexion moment followed sequentially by a right and left twisting moment and a right and left lateral bending moment. Latissimus dorsi was normalised to maximum activation achieved during the static phase at the top of the pull-up exercise. For the spine

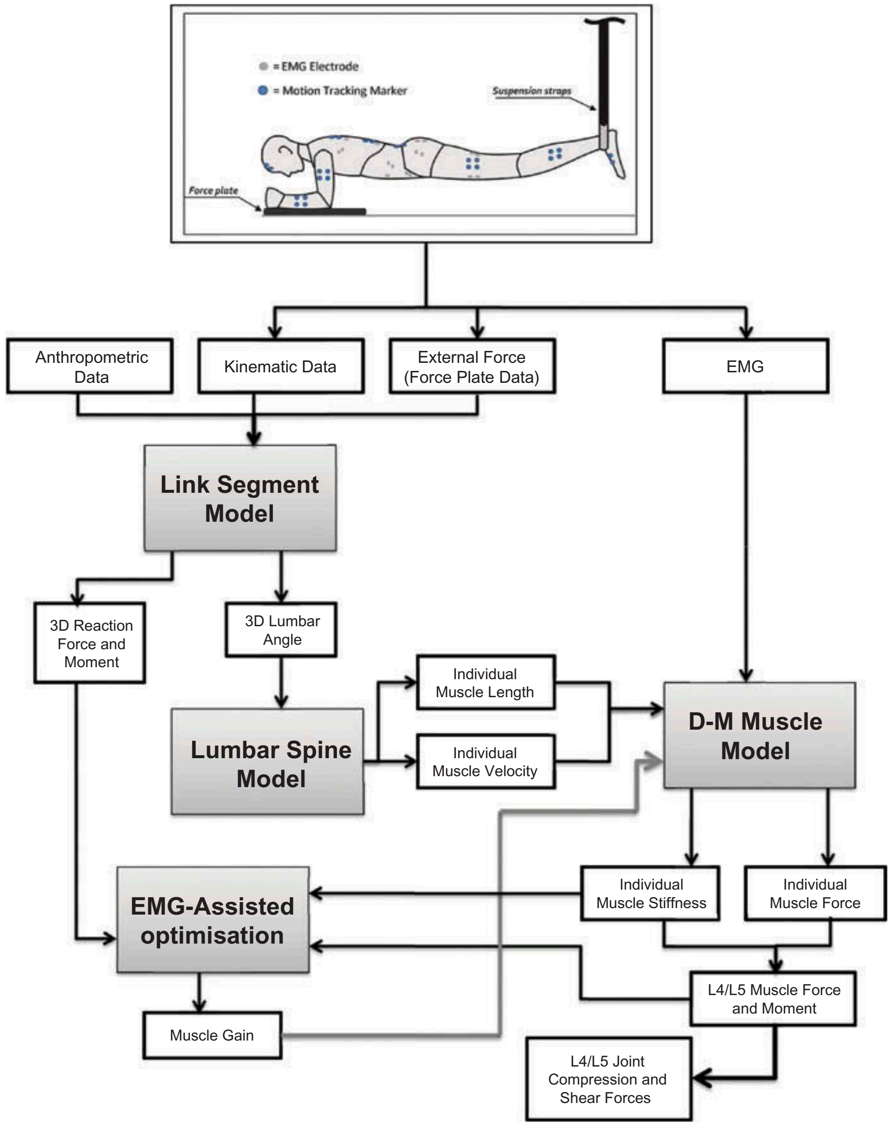

Figure 1. This flow diagram shows the data collected from the participant and input to the EMG processor and link segment model. The vertebral angles drive the lumbar spine model that prepares muscle forces for the EMG optimisation processor that balances predicted and measured moments.

extensors (lower erector spinae, upper erector spinae) and gluteus maximus, a resisted maximal extension in the Biering-Sorensen position was performed for normalisation. Gluteus maximus was cued to aid in extension at the hip. MVC for quadriceps involved the participant sitting on a therapy bed with his/her legs hanging over the edge. The participant grasped the edge of the bench behind them for support and performed a knee extension and hip flexion moment while being resisted by a research assistant. Gluteus Medius trials were performed in a side lying position during hip abduction, together with cued hip external rotation and extension (i.e., a lateral straight leg raise). Biceps brachii MVC was taken from a standing bilateral elbow flexion trial, resisted with straps that were secured to the ground at an angle that the participant felt he/she could elicit maximal muscle activation. The trapezius MVC trial made use of a set of straps similar to the biceps brachii MVC; however, participants were instructed to perform a maximal shoulder elevation effort. The MVC protocol for triceps, anterior deltoid, pectoralis major and serratus anterior were done from a supine push effort. Straps were secured to the ground at the participant's head and adjusted to a length the participant felt he/she could achieve maximal activation. With the straps at full length, the elbows were slightly flexed from full extension. The push was done isometrically, with the triceps cued to extend the elbow at the top of the push. Each maximum effort was conducted in a ramp fashion, up and down, while being coached. Practice contractions were allowed to enhance the technique. The maximal amplitude observed during the normalising contraction for each muscle was taken as the maximal activation for that particular muscle.

Body segment kinematics and marker placement. Eighteen reflective markers for tracking linked segment kinematics were adhered to the skin with hypoallergenic tape over the following landmarks bilaterally: first metatarsal head, fifth metatarsal head, medial malleoli, lateral malleoli, medial femoral condyles, lateral femoral condyles, greater trochanters, lateral iliac crests and acromia. Ten rigid bodies moulded from splinting material were adhered to the skin with hypoallergenic tape over the following areas: right and left feet, right and left shins, right and left thighs, sacrum, 3 cm medial to the right ASIS, inferior to the left scapula at the level of T12 and sternum. At least four reflective markers were adhered with tape to each rigid body (thigh clusters were comprised of six markers). The VICON Nexus™ motion capture system tracked the three-dimensional coordinates of the reflective markers during the various trials at a sample rate of 60 Hz (Figure 2).

Exercise description

Participants were asked to perform the exercises with some being variations of one another. A metronome set to 1 Hz (1 beat per second) was used to maintain consistent movements throughout all exercises, except for the walkout. A research assistant counted out loud to help participants maintain a steady pace. Three repetitions of all exercises were performed. All exercises are shown in Figure 3.

Exercises

1. Body saw – with the feet suspended in the labile suspension straps, knees bent and the

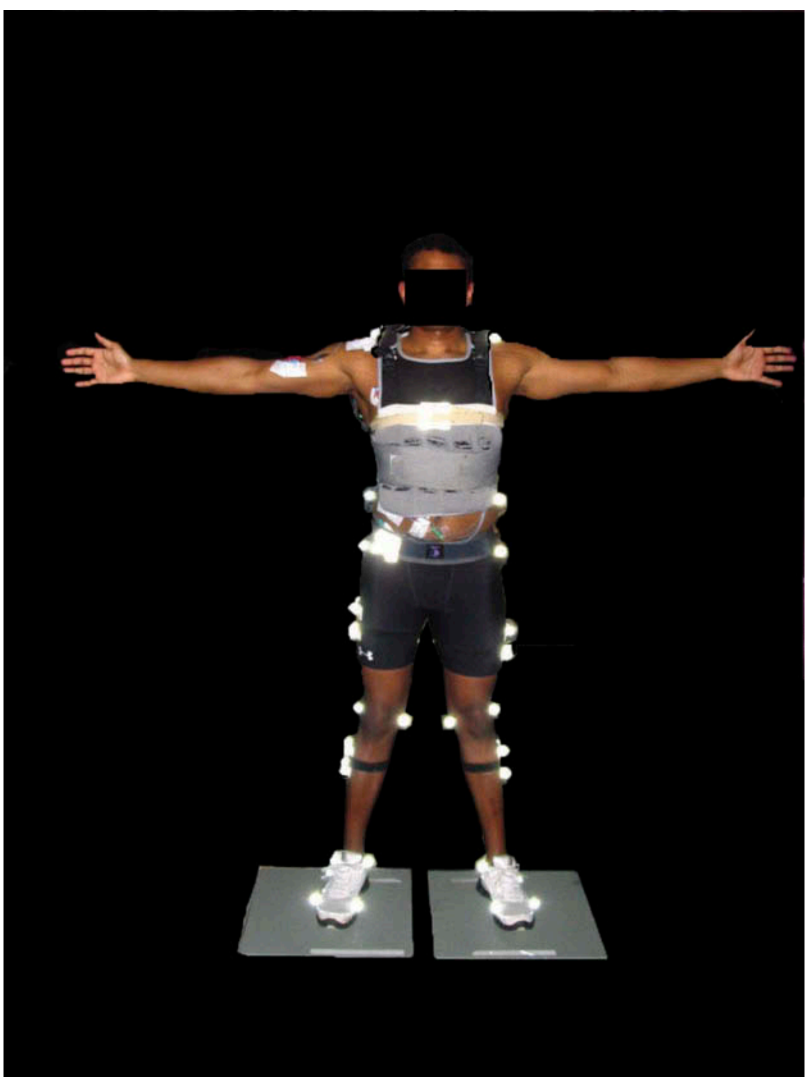

Figure 2. EMG and reflective marker placement in the calibration pose.

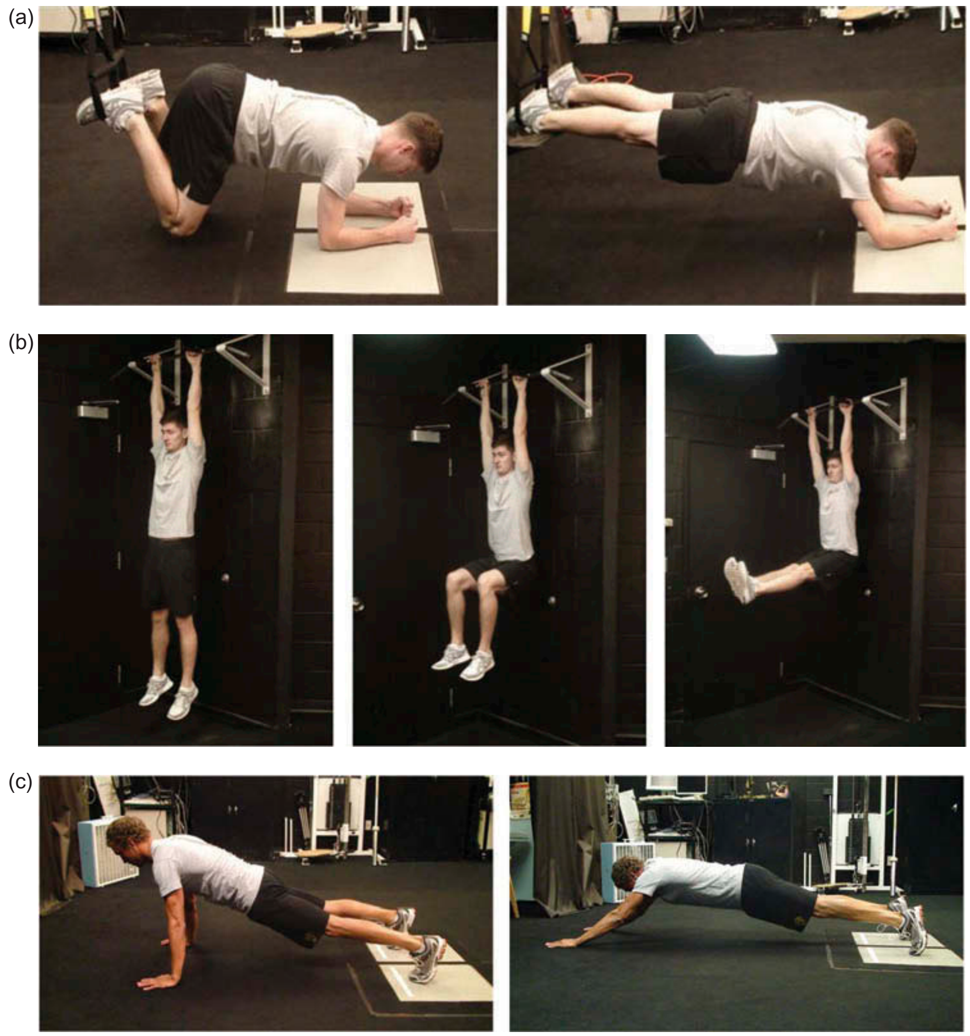

Figure 3. The exercises performed: (a) Body saw, (b) Leg raises, (c) Walkout.

forearms on the ground to support the weight of the body, participants were asked to straighten their legs and "saw" back and forth as far as possible over 2 s (i.e., 2 beats of the metronome). Once at full extension, the position was held for 1 beat before the participant "sawed" back to the original knees-bent position over 2 beats. The starting position was held for 1 beat before the next repetition began.

2. Leg raise, knees bent – hanging from an overhead bar, participants were instructed to raise their knees to create a 90° angle at the hips and knees over 2 beats while attempting to keep their spine in a neutral position. They held this position for 1 beat before descending over 2 beats and holding at the bottom for 1 beat.
3. Leg raise, knees straight – with the same start position and movement as the knees bent trial, participants were asked to raise their legs with knees straight to create a 90° angle at the hips while maintaining a neutral spine. The tempo of this exercise was the same as the leg raise, knees bent trial.
4. Walkout – from a push-up position, participants walked their hands forward as far as possible and were told to hold that position for 1 s. They then walked their hands back to a push-up position. Participants performed this task at their own pace.

Participants were taken through a familiarisation process before data collection began. They were instructed on how to generally position themselves for each task and were provided the opportunity to try some of the exercises. Each exercise was thoroughly explained and demonstrated immediately before it was performed. The order of exercises were randomised.

Data analysis

EMG to capture muscle activation for the spine model. The EMG data were band pass filtered between 20 and 500 Hz, full wave rectified, low pass filtered with a second order Butterworth filter at a cut-off frequency of 2.5 Hz (to mimic the frequency response of torso muscle, Brereton and McGill (1998)), normalised to the maximal voluntary contraction of each muscle to enable physiological interpretation, and down sampled to 60 Hz using custom LabVIEW™ software.

Kinetic and kinematic data to predict back loads. The three-dimensional coordinates of the markers were entered into a software package (Visual3D™, C-Motion, Germantown, MD, USA) which calculated the spine curvature angles as well as the reaction moments and forces about the lumbar spine (represented by the L4/L5 joint). Normalised EMG signals and lumbar spine position data were entered into an anatomically detailed model of the lumbar spine. Specifically, the modelling process proceeded in four stages:

1. The three-dimensional coordinates of the joint markers drove a linked segment model of the arms, legs and torso constructed with Visual3D™. This package output the lumbar spine postures described as three angles (flexion/extension, lateral bend and twist), bilateral hip angles and bilateral knee angles together with the reaction moments and forces about the L4–L5 joint.
2. The reaction forces from the link segment model above were input into a "Lumbar Spine model" that consists of an anatomically detailed, three-dimensional ribcage, pelvis/sacrum and 5 intervening vertebrae (Cholewicki & McGill, 1996). Over 100 laminae of muscle, together with passive tissues represented as a torsional lumped parameter stiffness element, were modelled about each axis. This model uses the measured 3D spine motion data and assigns the appropriate rotation to each of the lumbar vertebral segments (after values obtained by White and Panjabi (1978)). Muscle lengths and velocities are determined from their motions and attachment points on the dynamic skeleton of which the motion is driven from the measured lumbar kinematics obtained from the participant. As well, the orientation of the vertebral segments along with stress/strain relationships of the passive tissues were used to calculate the restorative moment created by the spinal ligaments and discs. Some recent updates to the model include a much improved representation of some muscles (documented by Grenier & McGill, 2007).
3. The third model, termed the "distribution-moment model" (Guccione, Motabarzadeh, & Zahalak, 1998; Ma & Zahalak, 1991), was used to calculate the muscle force and stiffness profiles for each of the muscles. The model uses the normalised EMG profile of each muscle along with the calculated values of muscle length and velocity of contraction to calculate the active muscle force and any passive contribution from the parallel elastic components.
4. When input to the spine model, these muscle forces are used to calculate a moment for each of the 18 degrees of freedom of the 6 lumbar intervertebral joints. The optimisation routine assigns an individual gain value to each muscle force in order to create a moment about the intervertebral joint that matches those calculated by the link segment model to achieve mathematical validity (Cholewicki & McGill, 1994). The objective function for the optimisation routine is to match the moments with a minimal amount of change to the EMG-driven force profiles. The adjusted muscle force and stiffness profiles are then used in the calculations of L4–L5 compression and shear forces.

In this way, the model was sensitive to the different muscle activation strategies and movement patterns of each participant.

Averages of muscle activation (EMG), spine angles and L4–L5 compression forces (spine load) were calculated at 4 phases for the 3 repetitions of each exercise:

1. M1 – Midway between rest and the peak of the exercise, the point where they were halfway to full extension (Body Saw and Walkout) or legs raised (Leg raise exercises) was selected.
2. P – At the peak of the exercise, at full extension or with legs completely raised. An average was taken over the time that the participant held this position.
3. M2 – Midway between the peak and returning to a rested position, halfway back to a push-up, or knees tucked for the body saw or legs hanging position.
4. E – Rested position at the end of each exercise, body saw with legs tucked or with legs hanging. An average was taken over the time that the participant held this position.

Results

There were gradations of muscle activity and spine load characteristics to every task. Interestingly, the hanging straight leg raise created the highest challenge to the abdominal wall (>130% of a statically determined maximum voluntary contraction (MVC) in rectus abdominis, and 88% MVC in external oblique) and approximately 110% MVC in pectoralis major. (Note that the normalising contraction was isometric and it is common to measure levels much higher than 100% during dynamic contractions. While this is observable in the muscle activation levels, the model modulates the dynamic activation levels with muscle force/length and force/velocity relationships when predicting individual muscle forces. Thus, in terms of force, dynamic muscle contractions remain within the physiological limits of the muscle). As expected, all of the anterior chain exercises activated rectus abdominis to very high levels (110% MVC for a walkout and 103% for the body saw) (Table 1). The external oblique muscles were activated more than the internal oblique muscles in every task. Interestingly, the pectoralis major was highly activated (on average) during the hanging straight leg raise while the body saw resulted in almost 140% MVC activation of the serratus anterior. In general, the hanging straight leg raise created approximately 3000 N of spine compression while the body saw created less than 2500 N (Table 2). All other exercises produced substantial abdominal challenge although the body saw did so in the most spine conserving way.

Discussion

This report presents the biomechanical demands of some anterior chain full body linkage exercises. There are gradations of muscle activity and spine load characteristics with every task. In general, all exercises were quite conservative in terms of spine load, and certainly lower than in pulling tasks that would be considered posterior chain exercises involving external loads (e.g., Callaghan, Gunning, & McGill, 1998; Fenwick, Brown, & McGill, 2009). Nonetheless, all of the exercises tested here resulted in substantial challenge to the rectus abdominis. The anterior chain exercises also produced more challenge to the external obliques than the internal obliques. Comparing with studies of torso loading in upright exercises (McGill, Karpowicz, & Fenwick, 2009a, 2009b), it appears as though the external obliques are generally more involved in pushing and torso flexion torque production while the deeper obliques are more for creating stiffness necessary for load bearing. Other studies on the loading associated

Table I. Muscle activation, % MVC (mean and standard deviation) at the peak (P-phase) phase of torso exercises. REO – Right External Oblique, RIO – Right Internal Oblique, RLD – Right Latissimus Dorsi, RLES – Right Lower Erector Spinae, RRA – Right Rectus Abdominis, RUES – Right Upper Erector Spinae.

	RLD	RUES	RLES	Biceps	Triceps	Anterior Deltoid	Upper Trapezius	Pectoralis Major	RRA	REO	RIO	Quadriceps	Gluteus Maximus	Gluteus Medius	Serratus Anterior
Leg raise, knees straight	32.60	14.04	13.65	8.70	11.88	8.01	3.59	109.36	130.88	87.90	52.23	49.49	16.56	11.12	91.04
	27.84	11.92	10.87	7.93	13.13	8.92	2.62	95.78	43.85	61.11	30.76	25.03	41.33	6.54	124.56
Walkout	23.95	4.03	4.78	3.01	28.99	8.90	6.71	46.67	110.40	57.68	30.33	32.53	5.53	9.44	110.38
	28.63	2.83	5.75	3.25	19.78	10.82	3.58	35.65	59.07	52.71	17.31	18.88	12.05	8.02	110.39
Leg raise, knees bent	12.20	15.87	12.13	5.80	5.22	7.70	1.33	51.85	70.23	41.55	37.49	14.79	6.99	7.57	70.41
	10.41	11.27	14.92	4.96	4.83	10.74	1.51	45.93	38.52	30.64	29.11	17.80	15.27	4.80	118.66
Body saw	11.94	4.43	6.05	1.34	7.12	7.99	5.05	23.40	103.14	56.98	24.10	30.98	7.11	5.00	138.84
	11.47	5.84	6.81	0.98	5.15	6.19	6.10	14.29	49.68	51.91	18.33	13.24	16.81	2.34	132.81

Table II. Rank of mean spine compression at the peak phase (P-phase) of each exercise.

Exercise	Rank	Mean spine compression (*N*)	SD
Leg raise, knees straight	1	3057.8	1333.3
Walkout	2	2718.7	1051.3
Leg raise, knees bent	3	2530.1	1189.4
Body saw	4	2423.4	1043.4

with push-up variations (e.g., Freeman, Karpowicz, Gray, & McGill, 2006) showed similar loading to that observed in the body saw, although the saw produced cyclic hip motion and the correspondingly cyclic abdominal activity.

The limitations of this study include the sample population, who were healthy and relatively fit. However, only fit individuals could accomplish these exercises. Only 14 males participated. The data collection was complex requiring about 2 h instrumenting and calibrating each participant for 3D motion reconstruction and muscle activation normalisation. The data analysis to obtain joint loads was intensive and time consuming. Fortunately, the variability in the data was reasonably low suggesting that more participants would not have influenced the estimates of the "average" response. The anatomical model is of a 50% male, as such females were not recruited. A female model is planned for the future. Participants ranged in height from 1.62 cm to 1.84 cm, resulting in a slight discrepancy in joint moments when performing each exercise. Thus, interpretation of variance via the standard deviation values needs to be considered in terms of participant height non-homogeneity together with their response-producing exercise-specific variables. Finally, this was a descriptive study to report, for the first time, spine load and muscle activation magnitudes in these exercises. The intent was to create this database for comparison with loads created by other exercises that were reported elsewhere.

Exercise programme design is influenced by the context, where appropriateness of an exercise is guided by the individual in terms of injury history, training goals and current fitness level. The real expert in exercise prescription matches the training demand with the training goal while considering any special variables such as specific injury history. The data presented here can assist in this decision.

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