Three dimensional musculoskeletal modelling of the abdominal crunch resistance training exercise

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Abstract

The aim of this study was to evaluate the benefits and limitations of using three dimensional (3D) musculoskeletal modelling (LifeModeler^{TM}) in assessing the safety and efficacy of exercising on an abdominal crunch resistance training machine. Three anthropometric cases were studied, representing a 5th percentile female, 50th percentile and 95th percentile male. Results indicated that the LifeModeler^{TM} default model was capable of solving the forward dynamics simulations without adjustments. The modelling was able to indicate high risk for back injury when performing the abdominal crunch exercise as a result of the unacceptable intervertebral joint loading that occurs during the exercise. Individuals with small anthropometric dimensions such as some females and children cannot be accommodated suitably on the abdominal crunch machine which negatively impacts exercise posture and technique. Hip flexor muscle contribution in the execution of the exercise for the 5th percentile female was substantial thus reducing the efficacy of the exercise in isolating the abdominal muscles.

Keywords: Resistance training equipment, abdominal crunch, Lifemodeler^{TM}, inverse dynamics, forward dynamics

Introduction

The increased popularity of, and participation in resistance training worldwide is indicative of the level of interest in benefits derivable from this type of training (Vaughn, 1989). Ironically, participation in any type of physical activity places the exerciser in situations in which injury is likely to occur. Improvement in exercise equipment design could reduce the risk of injury (Dabnichki, 1998) as well as possibly increase the efficacy of the exercise.

Conceptual, physical and mathematical models have all proved useful in biomechanics (Alexander, 2003). Mathematical and computer modelling is suitable for a wide variety of applications such as the design, production and alteration of medical equipment as well as sports and exercise equipment (Alexander, 2003; Kazlauskiené, 2006). Capable of simulating musculoskeletal human models interacting with mechanical systems, three dimensional (3D) musculoskeletal modelling may be able to answer many questions concerning the effects of the resistance training equipment on the body. Thus we have previously shown that this method can successfully be used to evaluate a seated biceps curl resistance training machine (Nolte, Krüger, & Els, 2011).

This study presents the musculoskeletal modelling of three anthropometric cases while exercising on a commercially available seated abdominal crunch resistance training machine. The abdominal muscles are the major supporting muscles for the stomach area. They not only support and protect internal organs, but they aid the muscles of the lower back to properly align and support the spine for proper posture as well as in lifting activities (Beachle & Groves, 1992). There are several exercises for the abdominal muscles, such as bent-knee sit-ups, crunches, isometric contractions as well as exercises using specialized equipment and resistance training machines (McGill, 1995; Nieman, 2007). Two common types of abdominal crunch resistance training machines available include, machines that have the resistance at the back of the upper body and the exerciser has to grasp two handle bars in front of the chest as opposed to machines that have the resistance in the front of the chest in the form of a cushion or pad and the exerciser places his or her arms over the pad. In this study, the latter abdominal resistance training machine was utilized. Controversy remains as to which exercise method best activates the muscles of the abdomen and minimizes potentially harmful or excessive joint tissue loading (McGill, 1995). It is generally believed that a variety of selected abdominal exercises are required to sufficiently challenge the abdominal muscles and that these exercises will differ to best meet the different training objectives of the individual (Axler & McGill, 1997).

Evaluation methods are required to ensure equipment efficacy as well as the safety of the end-user. Thus, the primary aim of this study was to evaluate the benefits and limitations of using 3D musculoskeletal modelling in evaluating the abdominal crunch resistance training machine. We hypothesized that, 1) the Lifemodeler™ default model would be capable of solving the forward dynamics simulations without adjustments, 2) not all individuals (varying anthropometric dimensions) would be suitably accommodated by the abdominal crunch machine, 3) the abdominal crunch resistance training exercise places the exerciser at risk for back injury due to the fact regular abdominal exercises without resistance have been associated with large loads on the spine (McGill, 1995) and, 4) unsuitable accommodation on the abdominal crunch machine would negatively impact exercise safety and efficacy.

It should be noted that although the results from this study may be generalised to other computer modelling software, there may be certain aspects that are unique to the modelling software used in this study.

Methods

Equipment

Three sex specific 3D musculoskeletal full body models were created using LifeModeler™ software and incorporated into a multibody dynamics model of the abdominal crunch machine modelled in MSC ADAMS (Figure 1). The LifeModeler™ (San Clemente, USA) software runs as a plug-in on the MSC ADAMS software. LifeModeler™ software has previously been used in studies in the fields of sport, exercise and medicine (Agnesina et al., 2006; De Jongh, 2007; Hofmann, Danhard, Betzler, Witte & Edelmann, 2006; Olesen, Andersen, Rathleff, de Zee & Rasmussen, 2009; Nolte et al., 2011; Rietdyk & Patla., 1999; Schillings, Van Wezel & Duysens, 1996). Three default models, as generated through the software, were evaluated. These models consisted of 19 segments including a base set of joints for each body region. The spine does not consist of individual vertebrae but rather of various segments that represent different regions of the vertebral column with joints between these segments. The default models had a full body set of 118 muscle elements attached to the bones at anatomical landmarks, which includes most of the major muscle groups in the body (Biomechanics research group, 2006).

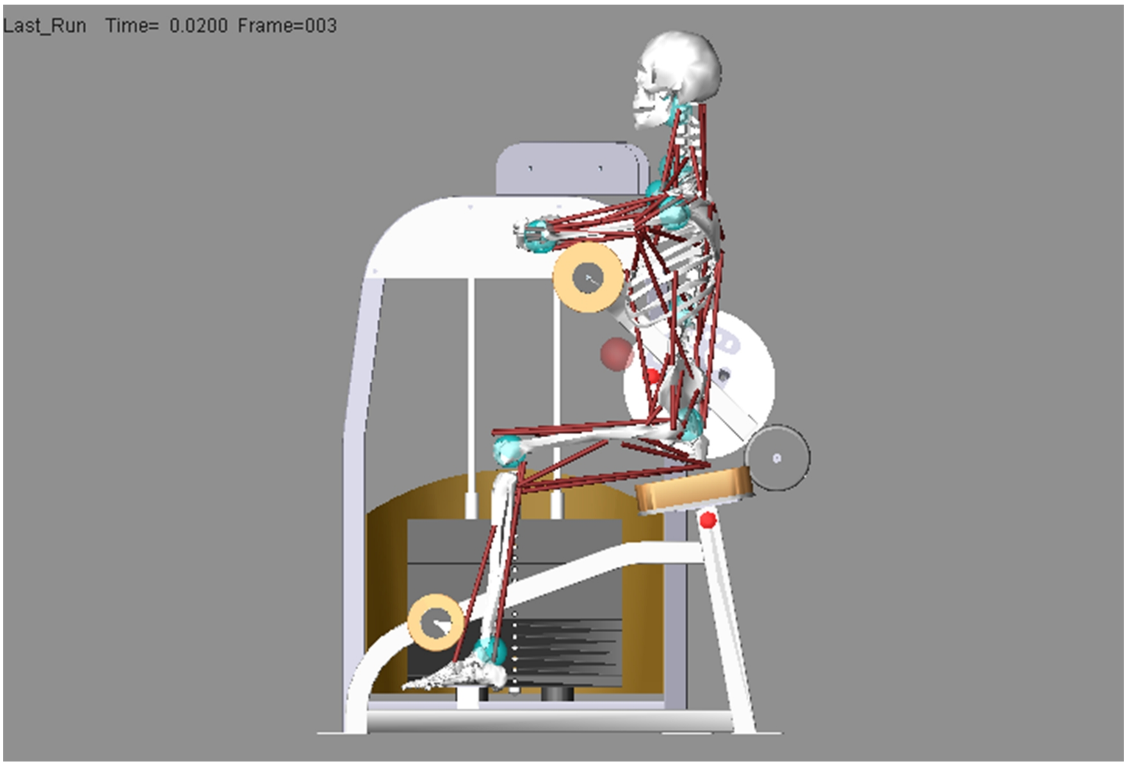

Figure 1: 3D musculoskeletal modelling of the abdominal crunch resistance training machine and 95th percentile male musculoskeletal model using LifeModeler™ and MSC ADAMS software.

Musculoskeletal full body human and the abdominal crunch computer aided design (CAD) models

Models for the three anthropometric cases were created. The human models were created using the GeBOD anthropometry database (default LifeModeler™ database) but were based on body mass index (BMI) data obtained from RSA-MIL-STD-127 Vol 1 (2004)(Table I). This standard is a representative of the South African National Defence Force (SANDF) which is kept current by a yearly sampling plan and can be considered an accurate representation of the broader South Africa population. A process described by Bredenkamp (2007) was followed to characterize the body forms of SANDF males and females found in RSA-MIL-STD-127 Vol 1. This process identified variances in body form as identified by principal component analysis. Two principal components (PCs) for the SANDF males and females were included in the modelling process and presented the positive boundary case (being tall and thin) and the negative boundary case (being short and heavy). Positive and negative boundary cases represent the boundary conditions to be accommodated in design (Gordon & Brantley, 1997). A “small” female, an “average” male, and a “large” male were the three anthropometric cases chosen for this study. They are traditionally known as a 5^{th} percentile female, 50^{th} percentile male and a 95^{th} percentile male based on the BMI. Thus, for the purpose of building these biomechanical models, a correlation between BMI and functional body strength was assumed. Similar assumptions have previously been made in biomechanics full body model simulations (Rasmussen et al., 2005). A study by Annegarn, Rasmussen, Savelberg, Verdijk & Meijer (2007) also verified scaled modelling strengths against actual functional body strengths and correlations ranged from 0.64 to 0.99.

This approach was followed in order to test whether the exercise machine could accommodate the full spectrum of the South African end-user population. A CAD model of the abdominal crunch resistance training machine was obtained from a South African exercise equipment manufacturing company (Figure 2). The model in a Parasolid file format was imported into the ADAMS simulation software.

The Adams software was used to create two design variables in order to adjust the external resistance (as selected by the amount of weights when using a selectorised resistance training machine) and to specify the radius of the cam over which the cable of an actual exercise machine would run in order to lift the selected resistance. This was possible since this machine employed a circular cam system. A special contact force (solid to solid) was created between the weights being lifted and the remainder of the weight stack during the simulation. A coupler joint was created linking the revolute joint (driver joint) of the lever arm attached to the abdominal crunch machine pad/cushion with the translational joint of the weight stack. The design variable created for the radius of the cam was then referenced as part of the function of the coupler joint in calculating the external resistance, taking into account the resistance selected as well as the radius of the cam on the machine. The design variable created for the mass of the weights was then adjusted according to the pre-determined resistance for each anthropometric case, explained in the next section.

The external resistance applied in the models was based on data obtained from Isokinetic testing results from trunk flexion (Perrin, 1993). Trunk flexion was selected as it most closely resembles the abdominal crunch movement. Torque (Nm) values obtained were converted to force values in Kilograms by adjusting for estimated isokinetic testing device lever arm length for each anthropometric case. Fifty percent of the functional strength one repetition maximum (1RM) for each anthropometric case was used, this can be considered a manageable resistance to perform an exercise with appropriate form and technique for four repetitions (Beachle & Groves, 1992) (Table I).

Table I. Anthropometric and user population strength data for population groups studied.

User population group	Body mass (kg)	Stature (mm)	User population group exercise resistance (50% 1RM) kg
5^{th} percentile female	49.5	1500	5
50^{th} percentile male	65.0	1720	14
95^{th} percentile male	85.0	1840	24

Simulation

Extreme care was taken with the positioning of the musculoskeletal model onto the abdominal crunch machine to ensure proper technique, posture and positioning according to best exercise principles (Table II). The engineered adjustability of the exercise machine was used in order to ensure correct positioning for each of the anthropometric cases. A bushing element was applied between the lower torso and the seat of the abdominal crunch machine as well as the two humeral bones and the abdominal crunch machine pad/cushion. Bushing elements were preferred to fixed joint elements because they allow for limited translational and rotational motion. Also, the amount of motion can be controlled by changing stiffness and damping characteristics in all three orthogonal directions. While compensatory movements should be limited during resistance training to ensure proper technique they do occur in most instances. Thus we applied and gradually increased the stiffness and dampening until we achieved visually acceptable kinematics in terms of such compensatory movements.

The inverse dynamics – forward dynamics method was applied during the simulations. Inverse dynamics simulations are performed on models which are being manipulated by the use of motion agents or motion splines. During the inverse dynamics simulation, a rotational motion was applied to the revolute joint of the lever arm attached to the pad/cushion of the abdominal crunch machine in order to generate the required movement of the resistance training machine. This movement replicated the pulling (concentric) and resisting (eccentric) phase of the exercise. The time for the concentric phase was set at 1.33 seconds and the eccentric phase slightly longer at 2.66 seconds to mimic conventional resistance training technique in which the eccentric phase is more deliberate to prohibit the use of momentum (Schilling et al., 2008). The 1.33 second concentric phase included a STEP function (ramp-up period) approximation over 0.5 seconds to ensure a gradual start to the movement. The muscles of the model were “trained” during the inverse dynamics simulation in order to calculate the changes in muscle lengths to result in the required machine movement. The movement replicated four repetitions of the exercise separated by a slight pause between repetitions.

After the inverse dynamics simulation was performed, the rotational motion was removed from the rotational joint of the lever arm of the abdominal crunch machine. The recorded muscle length changes and resulting joint movements were then used to drive the model during the forward dynamics simulation in the manner as developed through the inverse dynamics simulation. During the forward dynamics simulation the model is guided by the internal forces (muscle length changes resulting in joint angulations and torques) and influenced by external forces (gravity, contact and determined exercise resistance).

The muscle elements used during the modelling in this study are referred to as closed loop simple muscles. Closed loop muscles contain proportional-integral-differential (PID) controllers. The PID controller algorithm uses a target length-time curve to generate the muscle activation and the muscles follow this curve. The closed loop algorithm is governed by the following formula:,

$F – P_{gain} (P_{error}) + I_{gain} (I_{error}) + d_{gain} (d_{error})$ where:

- P_{error} is the target value – current value / range of motion
- D_{error} is the first derivative of P_{error}

- I_{error} is the time integral of P_{error} (Biomechanics Research Group, 2006).

All results presented are derived from the forward dynamics simulations.

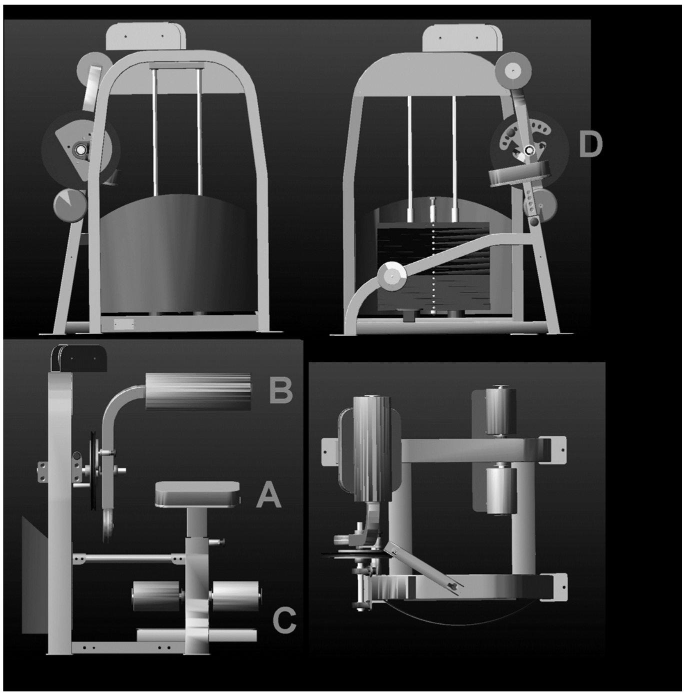

Figure 2: A side view from the right (top left), side view from the left (top right), front view (bottom right), top view (bottom right). Descriptions for the labelled parts are as follows: A = adjustable seat, B = abdominal crunch pad/ cushion, C = foot rest, D = circular cam.

Table II. Exercise starting posture for the 3 anthropometric cases on the abdominal crunch machine. Results are presented for the sagittal, transverse and frontal planes (degrees). Note that F = flexion, E = extension, and AB = abduction.

Joint	5th percentile female	50th percentile male	95th percentile male
Scapula	0.0; 0.0; .0.0	0.0; 0.0; 0.0	0.0; 0.0; 0.0
Shoulder	82.0(F); 0.0; 0.0	78.0(F); 0.0; 0.0	78.0(F); 0.0; 0.0
Elbow	90.0(F); 0.0; 90.0(F)	90.0(F); 0.0; 90.0(F)	90.0(F); 0.0; 90.0(F)
Wrist	0.0; 0.0; 0.0	0.0; 0.0; 0.0	0.0; 0.0; 0.0
Hip	40.0(F); 0.0; 7.0(AB)	63.0(F); 0.0; 7.0(AB)	77.0(F); 0.0; 7.0(AB)
Knee	20.0(F); 0.0; 0.0	55.0(F); 0.0; 0.0	70.0(F); 0.0; 0.0
Ankle	8.0(E); 0.0; 0.0	8.0(E); 0.0; 0.0	8.0(E); 0.0; 0.0
Upper neck	0.0; 0.0; .0.0	0.0; 0.0; 0.0	0.0; 0.0; 0.0
Lower neck	0.0; 0.0; .0.0	0.0; 0.0; 0.0	0.0; 0.0; 0.0
Thoracic	0.0; 0.0; .0.0	0.0; 0.0; 0.0	0.0; 0.0; 0.0
Lumbar	0.0; 0.0; .0.0	0.0; 0.0; 0.0	0.0; 0.0; 0.0

Data analysis

Firstly, it was determined if the forward dynamics simulations could be adequately solved by the Lifemodeler^{TM} default model. Kinematic data obtained from the inverse dynamics simulations was visually compared to that of the forward dynamics simulations in order to determine if the data was plausible.

Secondly, the anthropometric dimensions and exercise postures of the musculoskeletal human models were visually assessed in relation to the dimensions and adjustability of the resistance training equipment in order to determine if all three anthropometric cases representative of the South African end-user population could comfortably be accommodated on the abdominal crunch resistance training machine. Key aspects included start and end exercise posture as well as maintaining correct technique throughout the exercise during the simulations. Start and end exercise posture evaluation entailed positioning of the axilla and upper arms (humerus) on the top of the abdominal crunch pad touching the chest at the sternum. The trunk positioned at approximately 90º to the horizontal. The feet should be positioned on the provided supports with the hips flexed in order to protect the lower lumbar area from excessive strain during the exercise. Correct technique was assessed in terms of limited compensatory movements and performing the abdominal crunch through the full range of motion as determined by the inverse dynamics.

Lastly, in order to determine exercise safety and efficacy peak muscular force production and joint forces were evaluated. Specifically, peak muscular force production of the prime movers was analysed to determine exercise efficacy of the seated abdominal crunch. For the purpose of this study, efficacy of the equipment was assessed by evaluating whether the equipment exercised the muscles it was designed for, e.g. does the seated abdominal crunch machine exercise the prime flexors of the trunk / abdominal muscles? Furthermore, the risk of injury to the musculoskeletal system of the exerciser was ascertained by comparison of measured forces with safe loading limits for joints of the lumbar and thoracic spine. Risk to both these structures are real especially during exercises that require spinal flexion and extension (with and without resistance) and or during execution of exercise with poor postures. Different joint loading criteria were derived using biomechanical research taking into consideration the posture and anthropometry (Cooper & Ghassemieh, 2007). However, criteria for determining whether a particular task or exercise is "safe" based on tissue-level stresses or joint loading are available for only a small number of tissues and loading regimes (e.g. lower back motion segments in compression)(Wagner, Rasmussen & Reed, 2007). Therefore for this study anterior/posterior (A/P) shear forces and joint compression forces were used as safety criteria.

The basic descriptive statistical analysis of the results were completed using the STATISTICA© software package (Statsoft).

Results

Muscle force production (N) and muscle length (mm) for the right side are reported on. Theoretically, the results of the left and right side should be similar.

Force production (N) of the Erector spinae (ER), Rectus abdominis (RA), Oblique (O) as well as the hip flexor [Psoas major (PM) and Iliacus (I)] muscles are presented in Table III. Maximum force production was greatest for the O muscle in comparison to the RA muscle for all anthropometric cases (Figure 3). The 5th percentile female exerted the most force for all muscles analysed and the 50th percentile male the least, with the exception of the ES muscle which was slightly higher for the 50th percentile male in comparison with the 95th percentile male. The hip flexor muscles were only used by the 5th percentile female, specifically the PM muscle.

Muscle length results are presented in Table III. The mean muscle length for the ES, RA and O is greatest for the 95th percentile male and smallest for the 5th percentile female. The reverse is true for the PM and I muscles as the 5th percentile female measured the greatest mean muscle lengths. The mean muscle length is highest for the RA muscle in comparison with the O muscle and a similar trend was found with the PM muscle in comparison with the I muscle for the three anthropometric cases.

Due to the involvement of the spinal column in the abdominal crunch exercise, torque (Nm) for the T12/L1 intervertebral joint (thoracic) and the L5/S1 intertervertebral joint (lumbar) in the sagittal plane are presented in Table IV. For all anthropometric cases peak thoracic torque was greater than peak lumbar torque. The 5th percentile female’s peak thoracic torque was greater than that of the other two anthropometric cases as shown in Figure 4.

Results for the thoracic (T12/L1 intervertebral joint) and lumbar (L5/S1 intervertebral joint) spine compression and anterior/posterior (A/P) shear forces are presented in

Table V. The peak thoracic and lumbar spine joint compression forces are greatest for the 5th percentile female and least for the 50th percentile male. Peak thoracic spine joint compression forces are greater than the peak lumbar spine joint compression forces for all the anthropometric cases with the exception of the 5th percentile female whose peak lumbar spine joint compression forces exceed her peak thoracic spine joint compression forces.

Peak thoracic spine joint A/P shear forces are greater than peak lumbar spine joint A/P shear forces for all anthropometric cases (Table V). The 5th percentile female has the highest peak thoracic and lumbar spine joint A/P shear forces in comparison with the 50th and 95th percentile males.

Table III. Right Erector spinae, Rectus abdominis, Internal and External oblique, Psoas major and Iliacus (hip flexors) muscles force production (N) and muscle length (mm) results for the 3 anthropometric cases.

Musculoskeletal model	Muscles	Max muscle force production (N)	Mean muscle length (mm)
	Erector spinae (ES)	225.0	240.0
	Rectus abdominis (RA)	667.0	280.0
5th percentile female	Oblique (O)	1764.0	140.0
	Psoas major (PM)	1627.0	220.0
	Iliacus (I)	0.4	120.0
	Erector spinae (ES)	342.0	260.0
	Rectus abdominis (RA)	186.0	320.0
50th percentile male	Oblique (O)	503.0	190.0
	Psoas major (PM)	0.5	190.0
	Iliacus (I)	0.4	110.0
	Erector spinae (ES)	340.0	280.0
	Rectus abdominis (RA)	241.0	350.0
95th percentile male	Oblique (O)	618.0	200.0
	Psoas major (PM)	0.4	190.0
	Iliacus (I)	0.4	100.0

Discussion

The first relevant finding of this study was that based on a visual comparison of the kinematics the LifeModeler™ default models were adequate to solve the forward dynamics simulations for all the anthropometric cases. This was not the case for a previous study in which the seated biceps curl resistance training exercise was modelled (Nolte et al., 2011). Three adjustments had to be made to the musculoskeletal models on the seated biceps curl machine before the forward dynamics simulations could be solved namely; 1) increase the physiological cross-sectional area (pCSA) of the three default elbow flexor muscles, 2) manipulate the muscle origins and insertions and 3) decrease the joint stiffness in the forward dynamics simulations (Nolte et al., 2011). The reason for the adjustments not being necessary in this study could possibly be due to the fact that the trunk musculature of the default model is more comprehensive than that of the elbow and shoulder joints. The only relevant muscle that is omitted from the LifeModeler™ default model is the Transversus abdominis. Caution is advised when implementing such a visual assessment approach as in the case of the 5th percentile female one could argue that the resulting forward dynamics simulation was visually correct despite the kinetic results indicating implausible muscle and joint reaction forces.

The second relevant finding was that the software was able to sufficiently indicate anthropometric differences with regards to the abdominal crunch machine's engineered or manufactured adjustability. The anthropometric dimensions of the musculoskeletal models could be accommodated comfortably in relation to the dimensions and adjustability of the abdominal crunch machine except for the 5th percentile female (Figure 5). The small female's feet could barely reach the foot rest and the abdominal crunch pad/cushion was positioned too high and therefore could not be accommodated adequately under her axilla. Furthermore, her lumbar (L5/S1) spine joint could not be aligned properly with the axis of rotation of the machine. As a result her movement on the abdominal crunch machine was negatively impacted as her thoracic spine movement appeared to be exaggerated during the execution of the exercise to the point where it resulted in highly improbable joint loads, possibly an artefact of the modelling process.

The movement on the abdominal crunch machine could be compared to a bent knee sit-up movement, in a study conducted by McGill (1995) the analysis of a bent knee sit-up showed that most of the flexion rotation movement takes place about the hips and not the spine. Rather the spine remains close to the isometric flexed posture throughout the dynamic sit-up cycle. Thus, a sit-up exercise may be considered an isometric flexion exercise as far as the trunk musculature is concerned. The 50th and 95th percentile males appeared to have produced trunk flexion at the lumbar sacral region rather than the unnatural flexion of the thoracic region as demonstrated by the female model. Figure 5 illustrates that the mismatch between the female model anthropometry and machine adjustability resulted in excessive thoracic spine movement so that the thoracic joint reached its range of motion limits. While the results suggests that the female is at increased risk for injury due to poor accommodation by the machine it is possible that the values obtained for muscle tensions and joint loads are exacerbated by an artefact in the modelling process most probably caused by the thoracic joint movement exceeding the default range of motion. Furthermore, the large muscle lengths recorded specifically in the O muscle could also be an indication that there was exaggerated movement of the trunk rather than that of an isometric contraction in the small female although the other anthropometric cases recorded similar muscle lengths.

Thirdly, the following relevant findings were made regarding the biomechanical evaluation in terms of exercise efficacy and injury risk. The O muscles in comparison with the RA muscles exerted more force during the exercise for all anthropometric cases. This result was not entirely expected as the O muscles are traditionally exercised using trunk rotation or twisting to the left and right which bring the oblique muscles into more active contraction (Floyd, 2009). The O muscles however, also aid in lumbar flexion and posterior pelvic rotation and thus could explain its significant contribution to the execution of the movement of the abdominal crunch exercise. In addition in a study conducted by McGill (1995) it was found that the RA muscles activity to be slightly lower in bent knee sit-ups as opposed to the straight leg variety, while the O muscles were activated to a greater level presumably to make up the moment deficit. Similar results were obtained in this study in comparison with McGill (1995) with regards to abdominal RA and O muscle force production measured by means of electromyography (EMG) during the straight leg sit-up such as 206N and 236N respectively. However, the muscle force production results for both muscle groups in this study were higher for all anthropometric cases, specifically the 5th percentile female. It must be noted that Lifemodeler^{TM} default model only consists of 1 pair of oblique muscles, the orientation of the muscles appear to resemble that of the External obliques, this could have also contributed to the high recorded force production of the O muscles.

The ES muscle recruitment can be explained by means of its antagonistic role in relation to the RA and O muscles. In a study conducted on sit-ups it was found that the antagonist extensor moments are produced particularly by the thoracic extensors (Iliocostalis lumborum and Longissimus thoracis). Most of the extensor force was due to neural activation as well as due to passive elastic stretching (McGill, 1995). Higher levels of coactivity have a significant impact on the spinal loads since increased antagonistic muscle activity must be offset by the agonist forces. Thus, the muscle activity from the antagonistic muscles produces more loading in the spinal (compressive forces) structures without contributing to the ability to offset the external moment imposed by the spine (Davis & Marras, 2000). One could also argue that co-contraction of the antagonistic muscles serve a protective role by reducing shear forces, since the vertebra and discs can withstand compressive loads but are not well suited to large shear forces (Granata & Marras, 1999).

Usually when abdominal exercises are performed the exerciser tries to reduce the contribution of the hip flexors with regards to the execution of the movement. The most commonly recommended manner of reducing the contribution of these muscles is to bend or flex the hips as this shortens the iliopsoas muscle and other hip flexors thereby reducing their ability to produce force (Floyd, 2009). In addition, this action of the hips is supposed to reduce lumbar joint compression. However, Axler and McGill (1997) found this not to be the case as there were no differences observed in lumbar spine joint compression or the utilization of the hip flexor muscles in sit-ups performed with the legs bent versus with the legs straight. The positioning of the musculoskeletal model on the abdominal crunch resistance training machine in this study is such that the hips and knees are in a flexed position and results indicate that the Iliopsoas muscles did not significantly contribute to the movement with the exception of the 5th percentile female. The high recorded PM muscle force production in the small female appear unrealistic and could be due to a combination of an artefact as well as poor accommodation of the model. There was much less hip flexion for the 5th percentile female in comparison with that of the other two anthropometric cases. Therefore one could postulate that the exercise was not successful in isolating the abdominal muscles of the small female.

The 5th percentile females force production for all studied muscles was the greatest in comparison with the other anthropometric cases. This result is not unexpected as anatomical differences could be the reason for the greater force production in the small female such as a smaller lever arm as well as unfavourable positioning on the abdominal crunch machine, even although the resistance used for all three cases was proportionally calculated to correlate the anthropometric dimensions. In addition, it is possible that there may have been errors in the forward dynamics simulations kinetic data.

Joint torque values obtained for the thoracic and lumbar spine in the 50th and 95th percentile males as well as lumbar spine torque values of the 5th percentile female appear to be plausible when comparing the results to peak values obtained by means of isokinetic testing. Langrana and Lee (1984) report trunk flexion/extension values of 60 Nm and 95 Nm respectively in non-disabled female subjects and 136 Nm and 212 Nm respectively in non-disabled male subjects assessed in a seated position at 30 degrees per second. Bearing in mind that the values obtained in this study were not from maximal testing they were still substantially lower than the isokinetic values of Langara and Lee with the exception of the 5th percentile female's thoracic spine torque values which were considerably higher. This once again could have resulted due to her poor positioning, on the abdominal crunch resistance training machine.

Abdominal exercises are prescribed for both the prevention and treatment of low back injury. However, these exercises sometimes appear to have hazardous effects on the spine. A study conducted by Axler and McGill (1997) with the purpose of identifying abdominal exercises that optimize the challenge to the abdominal muscles but impose minimal load penalty to the lumbar spine found that no single exercise optimally trained all of the abdominal muscles while at the same time incurring minimal intervertebral joint loads. Accurate assessment of the risk of spinal injuries during occupational, athletic/exercise and daily activities as well as subsequent design of effective prevention and treatment programmes depend amongst others, on an accurate estimation of trunk muscle forces and internal spinal loads (i.e., intervertebral disc compression and shear forces)(Arjmand, Gagnon, Plamondon, Sharazi-Adl & Lariviére, 2009). Thus, an important aspect of this study involved assessing the intervertebral joint loads. The intervertebral discs work as a visco-elastic system that absorb and distribute forces acting on the spine. When submitted to compressive forces the collagen fibres of the annulus fibrosus are deformed radially expelling fluid from the nucleus pulposus of the discs (Adams & Hutton, 1985). It is important to bear in mind when making this analysis and applying the information that the spine of the default model does not consist of all the individual vertebrae but rather of various segments that represent the different regions of the vertebral column with joints between these segments. Individualised vertebra and corresponding joints might produce different results.

Previous research from the American National Institute for Occupational Safety and Health (NIOSH) recommends that spinal compression forces should not exceed 3.4 kN to avoid injury. However there is a very real threat of musculoskeletal injury before this failure limit value has been reached (Snook & Ciriello, 1991; Cooper & Ghassemieh, 2007, Knapik & Marras, 2009). British standards (BS EN 1005-3, 2002) recommend 600N as the cut-off point for carrying masses, no further recommendations other than "time of exposure needs to be minimised" and "a preferred system requires optimal ergonomic position with reduced back bending posture" are made. Therefore, the 5th percentile female's lumbar and thoracic spine joint compression forces were far above the recommended failure limit of 3.4 kN and therefore she would be at possible risk for a back injury bearing in mind that the high recorded values could have been an artefact of the modelling process. A possible cause of this artefact could be the improbably high muscle forces recorded for the O and PM muscles of this model. Future studies to validate the model is required in order provide clear design guidelines for the likely risk for females of small stature exercising on the equipment. However, the 50th and 95th percentile males' thoracic and lumbar joint spine compression forces were also high and therefore the results suggest that the exercise may pose a risk for back injury.

The thoracic spine joint A/P shear forces appear to be higher than the lumbar spine joint A/P for the three anthropometric cases. Both thoracic and lumbar spine joint A/P shear forces for all anthropometric cases are above the most commonly cited spine tolerance of 1000 N for shear force as stipulated by McGill (1996), with the exception of the 50^{th} and 95^{th} percentile males' lumbar spine joint A/P shear forces. Thus, this exercise appears to place all anthropometric cases at risk of injury. It is important to note that the modelling does not take conditioning differences between individuals of similar anthropometric dimensions into account which can protect the individual against spinal loading. Furthermore, increased strength of trunk flexors and extensors muscles are thought to raise intra-abdominal pressure and to decrease spinal loading (Aspden, 1988).

The results regarding the spine reaction forces are not surprising. Predictions of compressive load on the low back were found to be substantial during both isometrically held sit-ups and dynamic sit-ups with minimal acceleration components by Axler and McGill (1997). Therefore, forces on the back during a resistance exercise such as this can be expected to put substantial strain on the back especially if positioning is not adequate as with the 5^{th} percentile female.

Lastly, it should be noted when evaluating an exercise in terms of efficacy and injury risk it is sometimes useful to compare various exercise techniques, different exercises for the same muscle groups as well as different manufacturer's equipment for the same exercise.

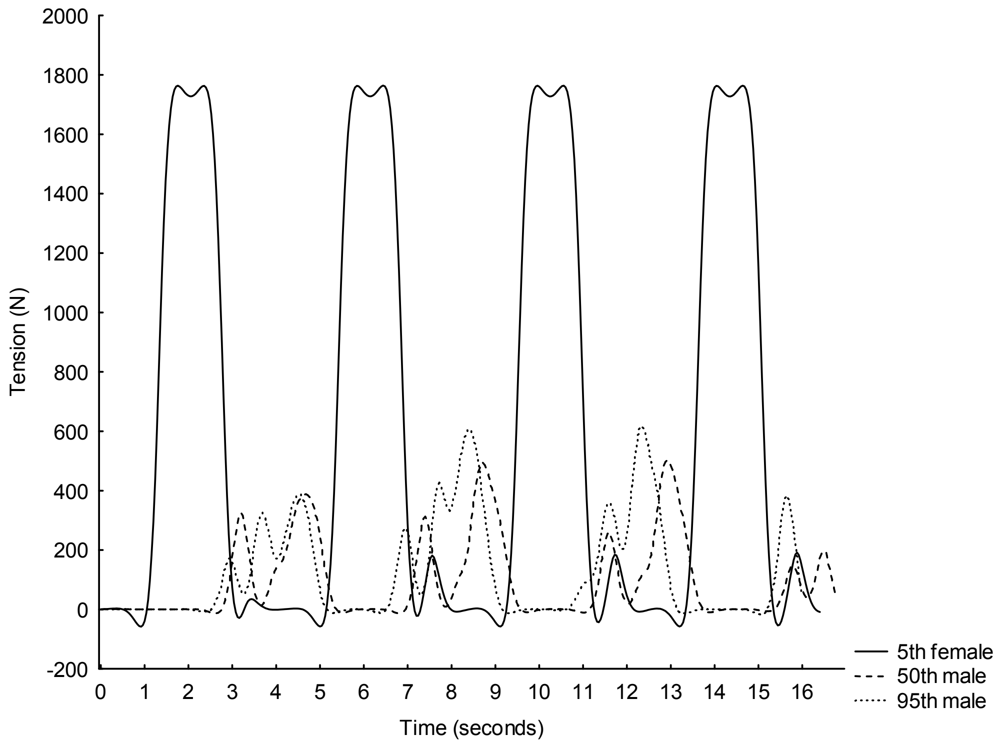

Figure 3: Right Oblique muscle force (N) for the 3 anthropometric cases (4 repetitions presented).

Table IV. Lumbar and thoracic joint torque (Nm) results in the sagittal plane for the 3 anthropometric cases.

Musculoskeletal model	Spinal joint	Mean (Nm)
5th percentile female	Thoracic spine	-257.0
	Lumbar spine	0.4
50th percentile male	Thoracic spine	-8.6
	Lumbar spine	-2.9
95th percentile male	Thoracic spine	-8.0
	Lumbar spine	-2.5

Table V. Thoracic and lumbar spine joint compression and anterior/posterior shear forces (N) for the 3 anthropometric cases. Note: for the compression forces, positive values indicate forces in a superior direction and negative values indicate forces in an inferior direction and for the anterior/posterior shear forces, positive values indicate forces in a posterior direction and negative values indicate forces in an anterior direction.

Musculoskeletal model	Spinal joint	Max compression forces (N)	Max anterior/posterior shear forces (N)
5th percentile female	Thoracic spine	11043.0	5827.9
	Lumbar spine	12580.2	5122.3
50th percentile male	Thoracic spine	4206.4	3201.3
	Lumbar spine	3388.6	559.9
95th percentile male	Thoracic spine	4673.9	3067.0
	Lumbar spine	3664.2	436.8

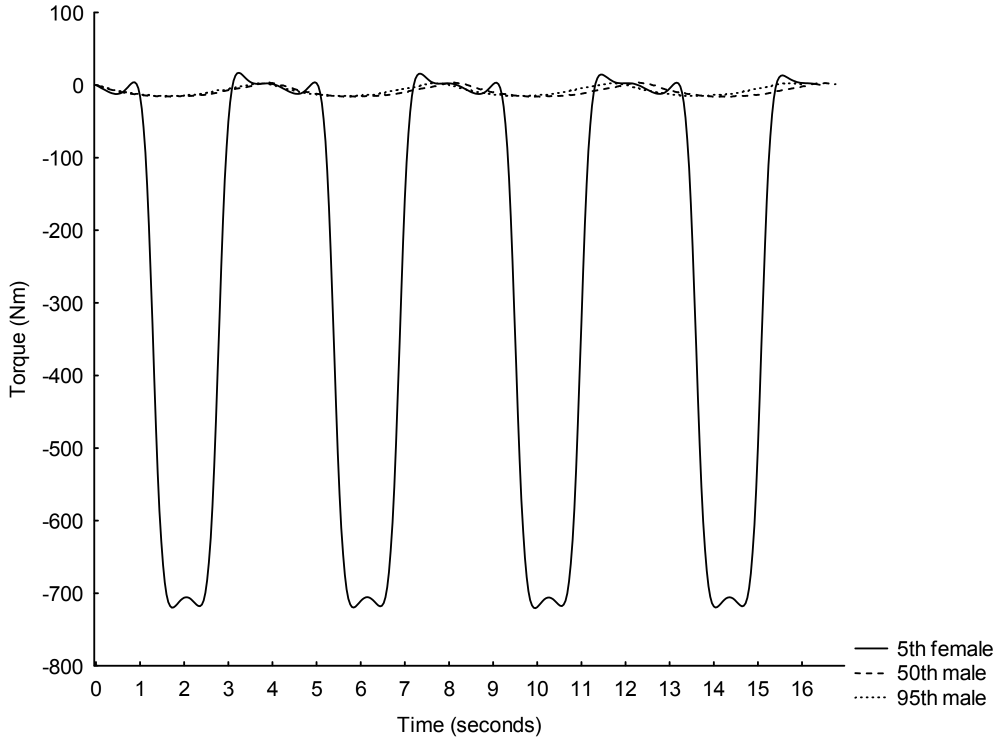

Figure 4: Thoracic spine joint torque (Nm) in the sagittal plane for the 3 anthropometric cases (4 repetitions presented). Note: negative joint angle indicates trunk flexion.

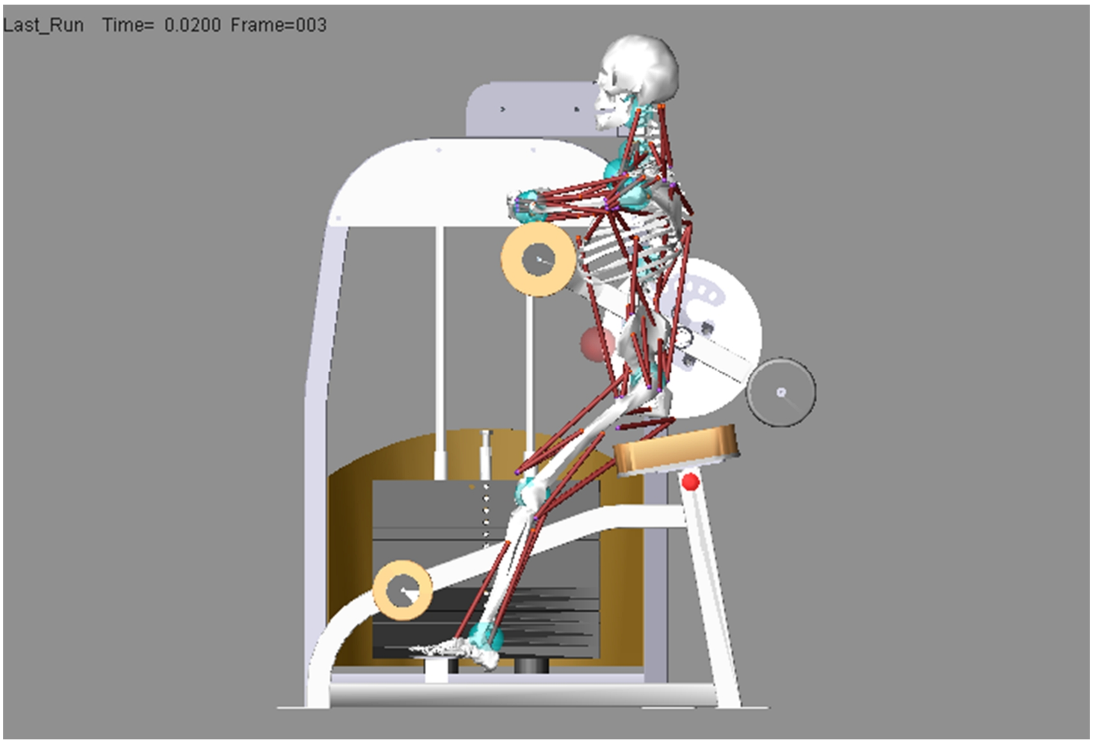

Figure 5. 5th percentile female's positioning on the abdominal crunch resistance training machine

Conclusion

It can be concluded that the default model of the Lifemodeler™ software was successful in evaluating the abdominal crunch resistance training exercise. No adjustments had to be made to the default model in order to solve the forwards dynamics simulations. The most significant value of the abdominal crunch resistance training machine 3D musculoskeletal modelling was in demonstrating the unacceptable thoracic and lumbar spine joint compression and A/P forces which could place the exerciser at high risk for a back injury. Therefore, caution should be used when prescribing the exercise for the training of the abdominal muscles especially if the individual has a predisposing back problem or injury. In addition, individuals of small anthropometric dimensions such as some females and children cannot be accommodated suitably on the machine which unfavourably influences exercise posture and technique which can further place the exerciser at increased risk for injury and decrease the efficacy of the exercise. Therefore, design adjustments to the abdominal crunch resistance training machine such as adapting the foot rest and abdominal crunch pad/ cushion length should be considered by the manufacturer. The models in this study were not validated and therefore this can be considered a limitation of the study, it is recommended that motion capture data should be used to determine individual movement patterns of each anthropometric case during such a validation.

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