CHAPTER 6

THREE DIMENSIONAL MUSCULOSKELETAL MODELLING OF THE CHEST PRESS RESISTANCE EXERCISE FOCUSING ON THE BIOMECHANICAL AND ANTHROPOMETRIC CONSIDERATIONS OF THE END-USER

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Abstract

The aim of this study was to evaluate whether three dimensional (3D) musculoskeletal modelling could be effective in assessing the safety and efficacy of exercising on a chest press resistance training machine. The focus of the evaluation was on biomechanical and anthropometric considerations of the end-user. Three anthropometric cases were created; these represented a 5th percentile female as well as a 50th and 95th percentile male based on body mass index (BMI). Resistance on the chest press machine was set at fifty percent of the functional strength one repetition maximum (1RM) for each anthropometric case, two repetitions were performed. The results indicate that adjustments had to be made to the default model in order to solve the forward dynamics simulations using recorded joint angulations during the inverse dynamics simulations. As a result no muscle (force and contraction) results could be obtained which negatively impacted the value of the modelling in evaluating the chest press exercise. The anthropometric dimensions of the end-users appeared to be adequately accommodated by the chest press's engineered or manufactured adjustability. Although pushing activities can pose a potential risk for spine injuries it did not appear as if the exercise put undue strain on the spinal structures when exercised with correct positioning and technique at an

appropriate resistance. However, the wrist joint and cervical spine appear to be vulnerable areas during the chest press exercise due to the relatively high wrist torque values in comparison to other joints as well as the relatively high cervical compression loads recorded. To conclude, although important design elements and flaws were highlighted by the 3D modelling in this series, mathematical and computer modelling does have its limitations especially when the default model is used.

Keywords: *Resistance training equipment, chest press, biomechanics, anthropometric, modelling, Lifemodeler^{TM}, inverse dynamics, forward dynamics*

Introduction

This article constitutes the fourth and last article in a series. The series consists of three dimensional (3D) musculoskeletal modelling with a focus on biomechanical and anthropometric variables of four commonly used pieces of resistance training equipment. Design of exercise equipment is a complicated task and warrants consideration of a series of biomechanical and ergonomics factors. Furthermore, there is inevitably increased loading on certain parts of the body due to the repetitive nature of exercises. Improvement in equipment design could reduce these hazards and offset such a negative effect on the body (Dabnichki, 1998). Currently, there is no regulation of exercise equipment design and production in South Africa. Therefore, a need exists to subject such pieces of equipment to evaluation methods of which the goal is to ensure the equipment's efficacy as well as the safety of the end-user.

Resistance training has emerged as an essential part of the individual's programme to improve performance, fitness, and even health. Although resistance training machines are expensive, they have several advantages over free weights: They are safer and more versatile, they save time, and they eliminate equipment theft. Using a machine also makes it much easier to change resistance as you move from one exercise to another. On the negative side, a machine restricts you to a set series of lifts and movements and you don't learn to balance the load as well (Sharkey and Gaskill, 2007). This article covers the evaluation of the open-kinetic-chain chest press resistance training exercise. Some of the most popular exercises in resistance training are those that work the chest musculature (Pectoralis major and Pectoralis minor). When developed properly, these muscles contribute a great deal to an attractive upper body and to added success in many recreational and athletic activities. Many chest exercises provide an added benefit because they also work muscles of the shoulder (Anterior deltoids) and upper arm (Triceps brachii) (Beachle and Groves, 1992).

Methods

Equipment

A 3D musculoskeletal full body model was created using LifeModeler™ software and incorporated into a multibody dynamics model of the chest press resistance exercise machine modelled in MSC ADAMS (Figure 1). The LifeModeler™ 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 (Schillings *et al*., 1996; Rietdyk and Patla., 1999; Hofmann *et al*., 2006; Agnesina *et al*., 2006; De Jongh, 2007; Olesen *et al*., 2009). It was decided to evaluate a default model as generated through the software. This model consists of 19 segments including a base set of joints for each body region. Specifically, 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. Furthermore, the default model has 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. Closed loop simple muscles were modelled. 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. Because of this approach, an inverse dynamics simulation using passive recording muscles is required prior to simulation with closed loop muscles. Simple muscles fire with no constraints except for the physiological cross-sectional area (pCSA), which designates the maximum force a muscle can exert. The graphs of simple muscle activation curves will generally peak at a flat force ceiling value (Biomechanics research group, 2006).

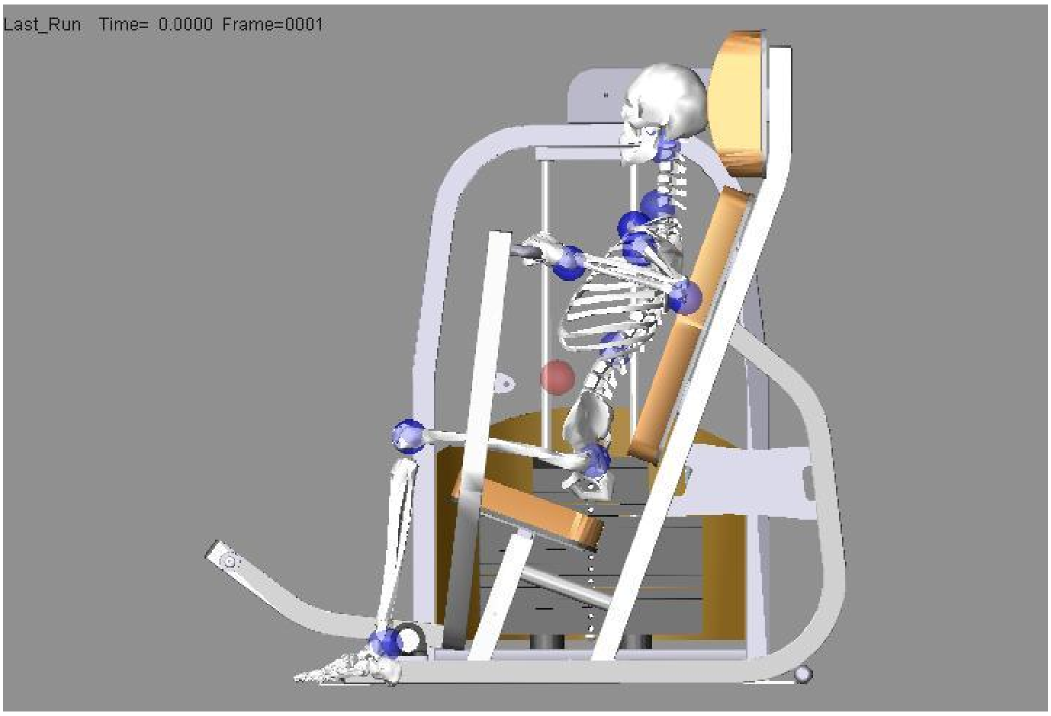

Figure 1. 3D musculoskeletal modelling of the chest press resistance training machine and 95th percentile male musculoskeletal model using LifeModeler™ and MSC ADAMS software.

Musculoskeletal full body human and chest press computer aided design (CAD) models

Three anthropometric cases were created for each piece of equipment. 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). This standard is a representative anthropometry standard of the South African National Defence Force (SANDF) which is kept current by a yearly sampling plan and is an accurate representation of the broader South Africa population. Bredenkamp (2007) described a process to characterize the body forms of SANDF males and females. Body form variances described by two principle components (PC's) for the SANDF males and two PC's for SANDF females were included in the modelling process. Positive and negative boundary cases of each PC, representing the boundary conditions to be accommodated in design (Gordon and Brantley, 1997), identified the total range of four male and four female models. It was decided to use the cases representing the smallest female as well as an average and large male for the three anthropometric cases for this study. These cases could be seen as what are traditionally known as a 5th percentile female, 50th percentile male and a 95th percentile male based on the BMI of each of these cases. 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.,* 2007). A study by Annegarn *et al.* (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 ensure that the equipment can accommodate an acceptable sample of the South African end-user population. A CAD model of the chest press resistance training machine was obtained from a South African exercise equipment manufacturing company. The model in a Parasolid file format was imported into the LifeModeler™ 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 however, this would not be possible with exercise machines employing non-circular cam systems, in order to attain better mechanical advantage for the end-user. 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) of the lever arm attached to the handle bars with the translational joint of the weight stack. The design variable created for the radius of the cam was referenced as the scale of the coupled joint (translational joint at weights). The design variable created for the mass of the weights was then adjusted according to the pre-determined resistance for each anthropometric case.

The external resistance applied in the models was based on data obtained from RSA-MIL-STD 127 Vol 5 (2001). This database consists of a range of human functional strength measurement variables for SANDF males and females. Due to its representivity this standard may be considered an accurate representation of the functional body strength of the South Africa population (RSA-MIL-STD-127, 2001). Furthermore, functional strength data was used from activities that most closely resembled the movements of the exercise as well as the muscle groups used during such movement. Fifty percent of the functional strength one repetition maximum (1RM) for each anthropometric case was used as this can be considered a manageable resistance to perform an exercise with appropriate form and technique for two repetitions.

Simulation

Extreme care was taken with the positioning of the musculoskeletal model onto the chest press machine to ensure technique, posture and positioning was correct according to best exercise principles (Table I). Furthermore, total manufacturer adjustability of the exercise machine was used in order to ensure correct positioning for each of the anthropometric cases. The following steps were performed in order to ensure realistic kinematics during the inverse dynamics simulations: 1) Positioning of the human model on the exercise equipment, 2) Adjustment of the posture to allow for the human machine interface to be created, 3) Creating the constraints between the human and machine, 4) Prescribing the motion of the repetitions, 5) Evaluation of the resultant kinematics, 6) Adjustment of joint positions until inverse dynamics resulted in a realistic exercise movement. Bushing elements were used to secure the lower torso to the seat as well as the neck to the back rest and spherical joints were used to connect the hands to the handle bars of the chest press machine. Bushing elements were preferred to fixed joint elements because it allows 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. The original joints created in the biomechanical model had default joint parameters (Stiffness (K) =1E4, Dampening (C) =1000). Joints with such high joint stiffness are created to ensure a relatively “rigid” model that provides a stable and smooth motion when manipulated by motion splines. This is especially important during the movement of the model into the initial posture as well as to ensure smooth model motion during inverse dynamics. After the muscle lengths had been recorded in the inverse dynamics, the joint stiffness was changed to near zero, to represent actual stiffness in human joints.

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 handle bars of the chest press 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 1.66 seconds and the eccentric phase longer at 3.33 seconds to mimic conventional resistance training technique in which the downward phase is more deliberate to prohibit the use of momentum. The 1.66 second concentric phase included a STEP function approximation over 0.5 seconds to ensure a gradual start to the movement. The joints forces of the model were recorded during the inverse dynamics simulation in order to calculate the changes in joint torques to result in the required machine movement.

After the inverse dynamics simulation was performed, the rotational motion was removed from the rotational joint of the lever arm of the chest press machine. The 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). It is important to note that changes had to be made to the LifeModeler™ default model in order to solve the models with plausible kinematics during the forward dynamics simulations. Considering the research problem the detail of these changes will be discussed under the discussions section. All results presented are derived from the forward dynamics simulations after these changes to the default model were made.

Table I. Exercise starting posture for the 3 anthropometric cases on the chest press machine. Results are presented for the sagittal, transverse and frontal planes (degrees). Note that F = flexion, E = extension, IR = internal rotation, AB = abduction and AD = adduction.

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	20.0(E); 20.0(IR); 70.0(AB)	20.0(E); 20.0(IR); 74.0(AB)	20.0(E); 20.0(IR); 70.0(AB)
Elbow	125.0(F); 0.0; -5.0(AD)	125.0(F); 0.0; -5.0(AD)	125.0(F); 0.0; -5.0(AD)
Wrist	0.0; 10.0(IR); 0.0	0.0; 10.0(IR); 0.0	0.0; 10.0(IR); 0.0
Hip	68.0(F); 0.0; 10.0(AB)	85.0(F); 0.0; 10.0(AB)	93.0(F); 0.0; 10.0(AB)
Knee	55.0(F); 0.0; 0.0	85.0(F); 0.0; 0.0	85.0(F); 0.0; 0.0
Ankle	15.0(E); 0.0; 0.0	7.0(E); 0.0; 0.0	7.0(E); 0.0; 0.0
Upper neck	0.0; 0.0; 0.0	0.0; 0.0; 0.0	5.0(F); 0.0; 0.0
Lower neck	10.0(F); 0.0; 0.0	10.0(F); 0.0; 0.0	15.0(F); 0.0; 0.0
Thoracic	0.0; 0.0; 0.0	0.0; 0.0; 0.0	0.0; 0.0; 0.0
Lumbar	20.0(E); 0.0; 0.0	20.0(E); 0.0; 0.0	20.0(E); 0.0; 0.0

Data analysis

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 chest press resistance training machine. Key aspects included start and end exercise posture as well as maintaining correct technique throughout the exercise during the simulations.

The kinematic and kinetic data from the simulations were analysed specifically in terms of exercise efficacy and peak muscular and joint force production of the prime movers of the chest press. 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. The dynamic mode of analysis includes all the aspects of motion in the calculation of joint forces and internal stresses, including the effects introduced by changing velocity and acceleration components (Wagner *et al.*, 2007). Different joint loading criteria were derived using biomechanical research taking into consideration the posture and anthropometry (Cooper and Ghassemieh, 2007). However, criteria for determining whether a particular task or exercise is "safe" based on tissue-level stresses are available for only a small number of tissues and loading regimes (e.g. lower back motion segments in compression) (Wagner *et al.*, 2007).

Due to the nature of this study only basic descriptive statistics were performed by means of the STATISTICA© software package (Statsoft).

Results

Three anthropometric cases created for each piece of equipment based on BMI data obtained from RSA-MIL-STD 127 Vol 1 (2004) were used for the study and results were assessed (Table II). Table III represents the external resistance applied in the models, fifty percent of the functional strength 1RM for each anthropometric case was used for two repetitions.

Table II. Anthropometric details of population groups studied (RSA-MIL-STD, Vol 1, 2004).

User population group	Body mass (kg)	Stature (mm)
5th percentile female	49.5	1500
50th percentile male	65.0	1720
95th percentile male	85.0	1840

Table III. User population strength data (RSA-MIL-STD, Vol 5, 2001).

User population group	User population group exercise resistance (50% 1RM) kg
5th percentile female	7
50th percentile male	19
95th percentile male	35

Due to the involvement of wrist, elbow and shoulder joints in the chest press exercise, torque values for these joints are presented in Table IV. Values for the right side of the body are reported on as theoretically the values of the left and right side should be similar. Peak wrist torque values in comparison with the other joints were the highest for all the cases studied except the 95th percentile male. Peak shoulder torque values in comparison with the other joints were the lowest in the 5th percentile female and 50th percentile male. The lowest peak joint torque for the 95th percentile male was for the wrist. The 5th percentile female recorded the highest peak joint torque values for the wrist and elbow and the 95th percentile male for the shoulder (Figure 2).

Table IV. Right wrist, elbow and shoulder torque (Nm) results in the sagittal plane for the 3 anthropometric cases. Note that the torque values presented in the Figures are in Nmm due to the default units of the modelling software.

Musculoskeletal model	Joint	Mean (Nm)	Min.	Max.
5th percentile female	Wrist	1.3	-1.3	6.5
	Elbow	4.2	0.0	6.1
	Shoulder	0.7	0.0	1.1
50th percentile male	Wrist	0.8	-0.2	3.3
	Elbow	1.0	-0.7	2.3
	Shoulder	1.0	0.0	1.2
95th percentile male	Wrist	3.1	-6.8	2.7
	Elbow	2.0	-0.2	2.9
	Shoulder	1.8	-0.2	3.0

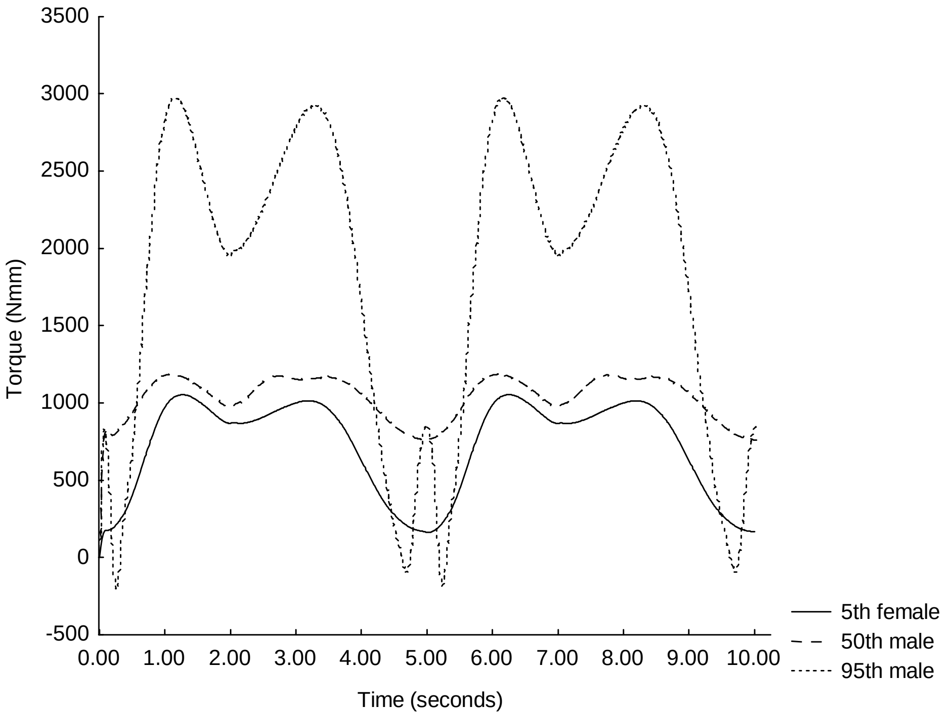

Figure 2. Sagittal right elbow joint torque (Nmm) for the 3 anthropometric cases (2 repetitions).

The chest press exercise is a multi-joint exercise thus movement in the sagittal plane of the shoulder, elbow and wrist are reported on (Table V). The least movement occurred at the wrist joint, followed by the shoulder joint with the most movement occurring at the elbow joint for the three anthropometric cases (Figure 3).

Table V. Sagittal right wrist, elbow and shoulder joint angle (°) for the 3 anthropometric cases.

Musculoskeletal model	Joint	Mean (degrees)	Min.	Max.
	Wrist	4.4	-0.4	8.6
5^{th} percentile female	Elbow	-105.0	-125.0	-76.7
	Shoulder	16.4	13.7	20.0
	Wrist	2.8	-3.5	7.1
50^{th} percentile male	Elbow	-103.4	-125.0	-73.1
	Shoulder	17.6	15.7	20.0
	Wrist	3.0	-3.6	7.9
95^{th} percentile male	Elbow	-102.3	-125.1	-68.6
	Shoulder	16.6	13.9	20.0

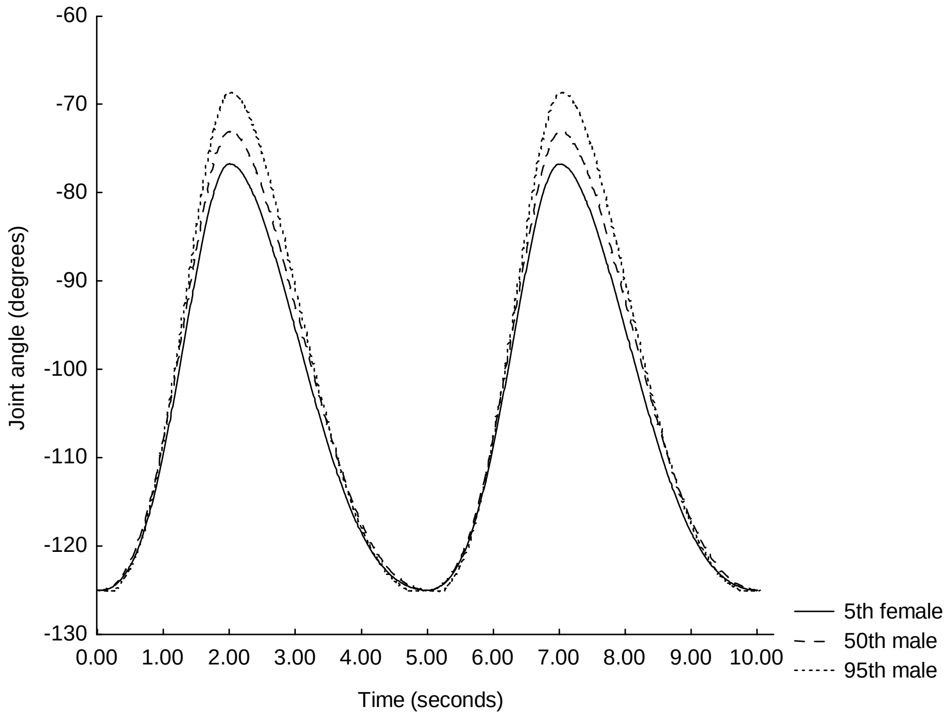

Figure 3. Sagittal right elbow angle (°) for the 3 anthropometric cases (2 repetitions). Note: negative joint angle indicates elbow flexion.

Results for cervical (C7/T1), thoracic (T12/L1 intervertebral joint) and lumbar (L5/S1 intervertebral joint) spine compression and anterior/posterior (A/P) shear forces are presented in Tables VI and VII respectively. Peak thoracic and lumbar spine joint compression forces were greatest for the 50th percentile male. While peak cervical spine joint compression was the highest for the 95th percentile male (Figure 4). In all the anthropometric cases the peak thoracic compression forces were the lowest, followed by the peak lumbar compression forces and the highest peak compression forces were recorded in the cervical spine.

Table VI. Cervical, thoracic and lumbar spine joint compression forces (N) for the 3 anthropometric cases. Note: positive values indicate forces in a superior direction and negative values indicate forces in an inferior direction.

Musculoskeletal model	Spinal joint	Mean (N)	Min.	Max.
5th percentile female	Cervical spine	-486.3	-590.2	-372.0
	Thoracic spine	100.3	79.4	149.1
	Lumbar spine	145.0	124.1	193.8
50th percentile male	Cervical spine	-467.1	-538.0	-329.0
	Thoracic spine	140.0	113.7	168.1
	Lumbar spine	200.0	173.2	227.6
95th percentile male	Cervical spine	852.5	1248	474.0
	Thoracic spine	-32.7	-97.1	162.8
	Lumbar spine	28.2	-36.1	223.9

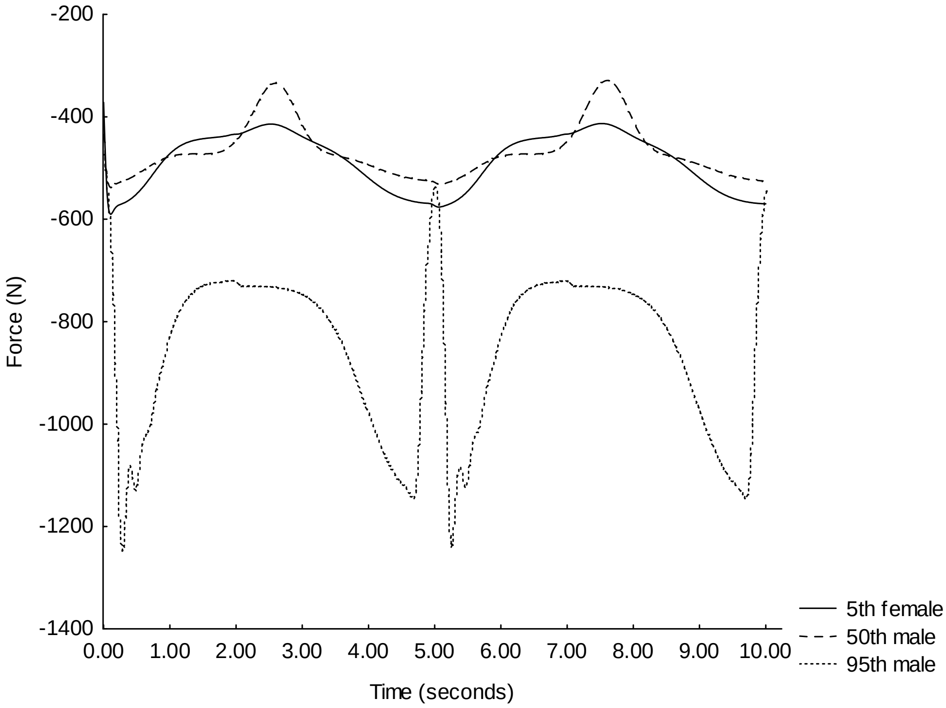

Figure 4. Cervical compression forces (N) for the 3 anthropometric cases (2 repetitions).

In all the joints except the cervical spine, the 95th percentile male had the highest peak A/P shear forces, followed by the 5th percentile female and lastly the 50th percentile male recorded the lowest A/P shear forces. The 5th percentile female recorded the highest cervical spine A/P shear forces and the 95th percentile male the lowest. The cervical peak A/P shear forces were the highest in comparison with the thoracic and lumbar spine joints for the 5th percentile female and 50th percentile male (Figure 5).

Table VII. Cervical, thoracic and lumbar spine joint anterior/posterior (A/P) shear forces (N) for the 3 anthropometric cases. Note: positive values indicate forces in a posterior direction and negative values indicate forces in an anterior direction.

Musculoskeletal model	Joint	Mean (N)	Min.	Max.
5th percentile female	Cervical spine	-486.4	-590.2	-372.0
	Thoracic spine	-311.7	-385.4	-232.3
	Lumbar spine	-266.9	-340.6	-187.6
50th percentile male	Cervical spine	-467.1	-538.0	-328.9
	Thoracic spine	-280.8	-336.5	-148.1
	Lumbar spine	-221.3	-277.0	-88.6
95th percentile male	Cervical spine	-66.0	-155.1	-18.3
	Thoracic spine	267.6	-400.8	128.2
	Lumbar spine	267.6	401.0	128.2

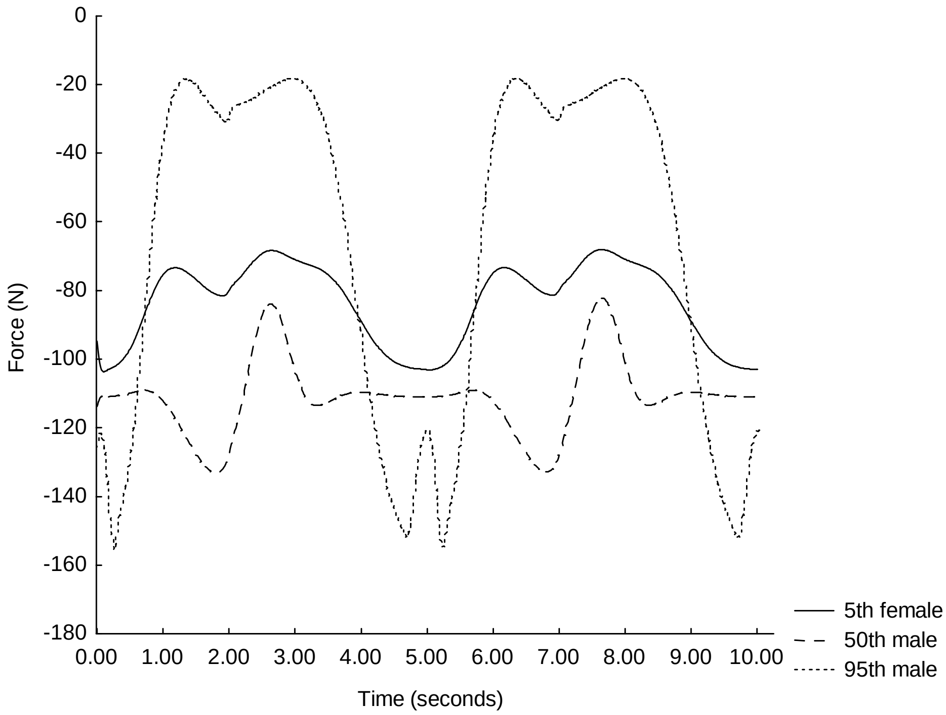

Figure 5. Cervical anterior/posterior (A/P) shear forces (N) for the 3 anthropometric cases (2 repetitions).

Results for wrist and elbow joint (right side) A/P shear forces are presented in Table VIII. Peak wrist and elbow joint A/P shear forces were lowest for the 50th percentile male and highest for the 95th percentile male (Figure 6). Peak wrist A/P shear forces were slightly lower than elbow shear forces for all the anthropometric cases.

Table VIII. Right wrist and elbow joint anterior/posterior (A/P) shear forces (N) for the 3 anthropometric cases. Note: positive values indicate forces in a posterior direction and negative values indicate forces in an anterior direction.

Musculoskeletal model	Joint	Mean (N)	Min.	Max.
5th percentile female	Wrist	-26.3	-31.7	-7.1
	Elbow	-26.3	-32.3	-3.5
50th percentile male	Wrist	-4.3	-14.4	6.8
	Elbow	-4.3	-15.0	7.4
95th percentile male	Wrist	-100.9	-118.2	-9.2
	Elbow	-100.8	-120.7	-7.4

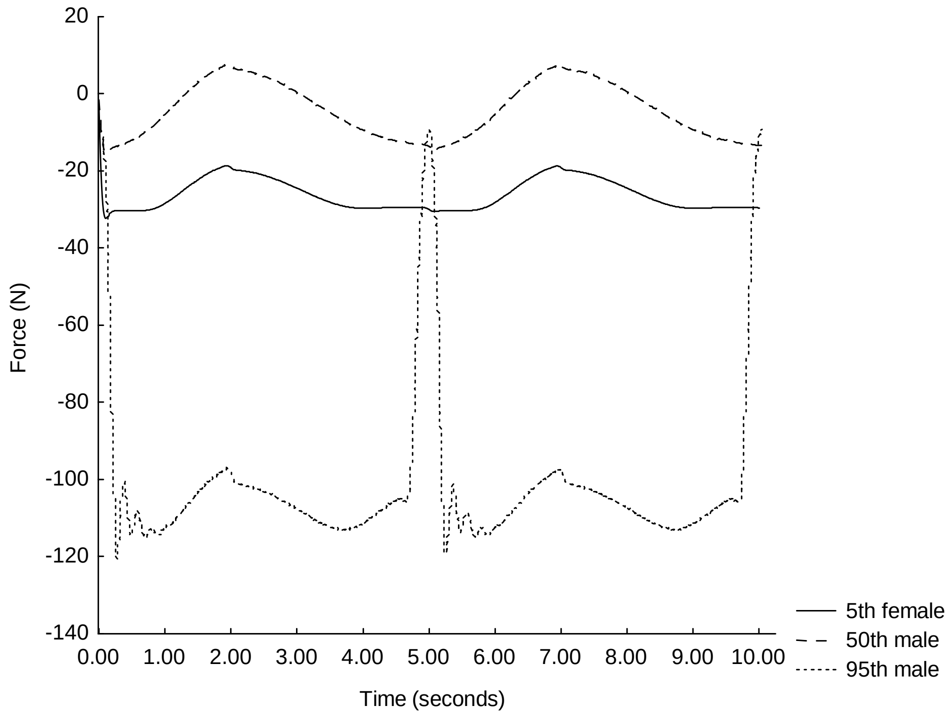

Figure 6. Elbow anterior/posterior (A/P) shear forces (N) for the 3 anthropometric cases (2 repetitions).

Discussion

It can be concluded from this study that the LifeModeler™ default model was not adequate to solve the forward dynamics simulations for any of the anthropometric cases. In the previous article (3) in this series which covered the evaluation of the seated row exercise, the same problem was encountered and the forward dynamics had to be solved by recording joint angulations in the inverse dynamics simulation. The same solution was made use of in this study. Possible reasons for this could include the degrees of freedom involved in a multi joint exercise involving highly mobile joints such as the shoulder. Furthermore it could be that additional musculature is required to provide more stability in the shoulder joint during the forward dynamics simulations. In this study in order to solve this problem the joint angulations recordings in the inverse dynamics simulations were used to solve the forward dynamics simulations. This option creates a trained PID-servo type controller on the joint axis. The joint is commanded to track an angular history spline with a user-specified gain on the error between the actual angle and the commanded error. A user-specified derivative gain is specified to control the derivative of the error. Therefore, results from muscle forces (N) and contraction (shortening and lengthening) (mm) could not be analysed. Ideally these parameters should be analysed when evaluating an exercise. It appears that more complex, multi-joint or compound exercises that require too many degrees of freedom such as the chest press and seated row exercise pose a problem for default model and therefore models with more detailed musculature may be required to solve the forward dynamics simulations sufficiently. Important musculature required for the performance of the chest exercises that are not included in the LifeModelerTM default model are the Serratus anterior, Coracobrachialis and Anconeous. It was not however within the scope of this study to produce anatomical detailed models but rather to evaluate the default model of the software.

Secondly, resistance training machines often can isolate muscle groups or joints while minimising extraneous body movements. Achieving this benefit requires that each individual be properly fitted, which may be a problem for certain populations. Children and small adults might not be able to adjust to the dimensions of the machine (National Strength and Conditioning Association, 1985). This study did not highlight any obvious anthropometric differences with regards to the chest press machine's engineered or manufactured adjustability. All three anthropometric cases appeared to be positioned adequately on the chest press machine. This was not the case in the modelling performed on the seated biceps curl and abdominal crunch machines, which demonstrated the inability of the machines to adjust appropriately to individuals with small anthropometric dimensions such as some females and children. As a result the exercise technique was negatively influenced and injury risk was increased for these exercises.

Lastly, with regards to the biomechanical evaluation in terms of exercise efficacy and injury risk the following could be deduced from the study. Due to the fact that the forwards dynamics simulations for this study were solved by recording the joint angulations changes during the inverse dynamics simulations and not muscle length changes. Results for the muscle forces and contractions were not obtained the therefore could not be analysed. This negatively influenced the value of modelling with regards to evaluating the chest press exercise as muscle forces and contractions provide important information regarding the efficacy and injury risk of the exercise.

Joint torque values obtained for the wrist, elbow and shoulder appear to be plausible when comparing the values to peak values obtained by means of isokinetic testing at 60 degrees per second. For example, wrist flexion/extension values of 13.8 Nm and 12.7 Nm respectively in non-disabled subjects (Van Swearigen, 1983). Elbow flexion/extension values of 36 Nm for both elbow flexion and extension in female college basketball players (Berg *et al.,* 1985) and shoulder flexion/extension values of 77 Nm and 53 Nm for males and 38 Nm and 24 Nm for females respectively in a group of non-disabled (Nicholas *et al*., 1989).

Joint torque values for the three joints evaluated were much lower than values obtained during peak isokinetic testing however it is important to bear in mind that the values obtained in this study were not from maximal testing such as the isokinetic testing. Interestingly, the peak wrist joint torque was the highest recorded value for all the joints in the anthropometric cases except the 95th percentile male which indicate the important role the wrist plays in the chest press or similar pushing movements. In contrast, the elbow produced the highest torque in the seated row exercise which is a pulling movement. Therefore, the results imply that proper alignment of the wrists during the chest press exercise may be important in an exercise such as the chest press because of the higher torque values produced by this joint.

The elbow joint range of motion was the greatest in comparison with the shoulder and wrist for the three anthropometric cases studied. This was to be expected as most of the movement that occurs in a chest press exercise is as a result of elbow extension produced by the elbow extensors, Triceps brachii and Anconeus muscles (Floyd, 2009).

Pushing and pulling as opposed to lifting activities might also be associated with significant risk to the low back (National Institute for Occupational Health, 1997, Hoozemans *et al.,* 1998). The chest press exercise can be considered a pushing activity. It must be kept in mind that the sited research is primarily referring to occupational tasks however important similarities and conclusions can be drawn with exercises that use similar actions to occupational tasks and activities that require pulling. Furthermore 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.

In 2009 a study by Knapik and Marras (2009) found that there was greater compressive loading at all spine levels when performing pulling compared with pushing activities. Therefore, one would expect the bench press (pushing action) exercise to be possibly safer than an exercise such as the seated row (pulling action) with respect to spine compression forces.

Previous research from the American National Institute for Occupational Safety and Health (NIOSH) (1997) 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 and Ciriello, 1991; Cooper and Ghassemieh, 2007, Knapik and Marras, 2009). British standards (BS EN 1005-3, 2002) recommend 600 N as the cut-off point for carrying masses, no further recommendations except " time of exposure needs to be minimised" and "a preferred system requires optimal ergonomic position with reduced back bending posture" are made. Therefore, all three anthropometric cases were well below the recommended failure limit of 3.4 kN. However, the cervical spine compression forces of the 50th percentile male and 5th percentile female were near the recommended cut-off of 600 N and the 95th percentile male exceeded the cut-off. It should be noted that the significantly higher forces recorded for the 95th male are considered to be an artefact of the constraint that was used in order to secure the head to the head-rest of the equipment. While it is not clear why the use of this constraint did not produce similar results in the other two models it may be that slight differences in positioning of the constraint could be the cause of the different results. The use of the constraint was however deemed necessary since the kinematics during the forward dynamics simulation was not acceptable without it. Without the constraint there was an unnatural movement in the chest region of the models. Considering the results of the 5th percentile female and the 50th percentile male it should still be noted that the chest press exercise appear to put the user at risk for injury in the cervical region.

Historically, spine compression in the lower lumbar spine has been the variable of interest for risk to the low back during work and exercise training. However, during horizontal force application (pushing of the chest press exercise), it is expected that shear forces within the spine increase dramatically due to the application of force in the hands and the reaction of the trunk musculature. Thus shear forces may represent the critical measure of risk (Knapik and Marras, 2009). According to Knapik and Marras (2009), in general, pushing activities impose greater potentially risky A/P shear forces upon the spine than pulling. Pushing imposed up to 23% greater A/P shear forces compared to pulling. Increases in shear forces were a result of the increased flexor muscle coactivity required for the activity. During extension (as in lifting a weight), the large erector spinae muscles can provide much of the power required for the lift. However during pushing, the flexor muscles that have a much more limited physiological cross-sectional area (pCSA) must generate internal force. In order to generate the required force, much greater co-activations of the muscle flexors are necessary. Since many of the oblique flexor muscles have a large horizontal muscle fibre orientation, these muscles produce significant shear forces (Knapik and Marras, 2009).

Although the peak spine A/P shear forces recorded were in general greater than the peak compression forces in this study, the cervical, thoracic and lumbar spine joint A/P shear forces for the three anthropometric cases are below the most commonly cited spine tolerance of 1000 N for shear force as stipulated by McGill (1996). It is important to note however that although the spine compression and A/P shear forces recorded were well within the acceptable limits the modelling does not take into account the repetitive nature and accumulative effect of exercise. Furthermore, the resistance used was only 50% of each of the anthropometric cases' estimated1RM and therefore if exercises use a resistance closer to their maximum the loading values may exceed the acceptable limits.

Handle height appears to affect the mechanical load at the low back and shoulder considerably and it is recommended that carts be designed and adjustable so that it is possible to push or pull at shoulder height (Hoozemans *et* *al.,* 2004). The same principle can be applied to the chest press machine and the handle bars should be at approximately shoulder height, which was the case for the three anthropometric models and thus this could have assisted in reducing the spine loads, especially the A/P shear spine forces. Unfortunately, after conducting a literature search it became clear that information regarding A/P shear forces of the shoulder, elbow and wrist joints is scarce. However, the following information regarding handle height may be applicable in terms of reducing A/P shear forces on these joints during the chest press exercise. Handle height and the magnitude of force were found to be significantly related to the net moment at the shoulder. Net moments at the shoulder are kept low during pushing and pulling activities by keeping the wrist, elbow, and shoulder close to the line of action of the exerted force or by directing the exerted force such that the shoulder joint remains close to the line of action of the exerted force (Hoozemans, *et al.,* 1998). Thus, alignment of the shoulder, elbow and wrist joints should be considered when designing the handle bars of chest press resistance training exercise machines which could assist in reducing shoulder strain during the exercise, especially when a heavy resistance is used.

Conclusion

Adjustments had to be made to the default model in order to solve the forward dynamics simulations using recorded joint angulations during the inverse dynamics simulations. As a result no muscle (force and contraction) results could be obtained which negatively impacts the value of the modelling in evaluating an exercise. The anthropometric dimensions of the end-users appeared to be adequately accommodated by the chest press's engineered or manufactured adjustability. Although pushing activities can pose a potential risk for spine injuries it did not appear as if the exercise put undue strain on the spinal structures when exercised with correct positioning and technique at an appropriate resistance. However, the wrist joint and cervical spine appear to be vulnerable areas during the execution of the chest press exercise due to the relatively high wrist joint torques produced in comparison to other joints as well as the reasonably high cervical compression loads recorded for the three anthropometric cases.

Finally to conclude this series, 3D musculoskeletal modelling was able to highlight some important design elements and flaws as well as biomechanical and anthropometrical limitations of the four pieces of resistance training machines evaluated in this series. It was demonstrated that anthropometric dimensions of the end-user must be taken into account by the designer and manufacturers of exercise equipment. Failure to do this can place the exerciser at risk for injury and reduce the benefits from using the exercise. Mathematical and computer modeling does however have its limitations especially when the default model is used. 3D musculoskeletal modeling is certainly the way of the future and with the developments and improvements that are continually being made will probably form a major role in the design of most types of equipment.

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CHAPTER 7

SUMMARY, RECOMMENDATIONS AND GENERAL CONCLUSIONS

7.1 SUMMARY

The motivation for this study originated from a concern for the quality and apparent lack of scientific data that supports exercise equipment design and specification. Currently, there is no standard biomechanical evaluation protocol for exercise equipment and more specifically resistance training equipment. Therefore, a need exists to develop and implement a basic biomechanical evaluation protocol for exercise equipment. As a result the safety of the exerciser will be maximised and the efficacy of the exercise will also be enhanced.

Therefore, the goal of this study was to evaluate whether a three dimensional musculoskeletal modelling (3D) protocol is effective in assessing the safety and efficacy of resistance training equipment. The focus of the evaluations was on the biomechanical and anthropometric considerations of the end-user.

The study aimed to achieve the follow objectives:

- To develop an evaluation protocol through computer modelling for resistance training equipment. The protocol will include:
    - anthropometry evaluation,
    - biomechanical evaluation;
- To implement the evaluation protocol on four pieces of resistance training equipment;
- Identify potential risk for musculoskeletal injury;
- Make recommendations on how the equipment could be improved with regards to design in order to maximise safety and exercise efficacy; and

- Make recommendations regarding limitations of the evaluation protocol. Evaluate if the protocol is sensitive enough to highlight injury risk and limitations in equipment design.

The hypothesis of the study was:
3D musculoskeletal modelling focusing on biomechanical and anthropometric considerations of the end-user is effective in evaluating the overall design of resistance training equipment.

The main findings of this research effort, in relation to the objects presented are:

7.1.1 Develop an evaluation protocol through computer modelling for resistance training equipment focusing on biomechanical and anthropometric considerations of the end-user.

An evaluation protocol through computer modelling was established.

The process followed included the following steps for each piece of equipment:

- Gather anthropometric data and corresponding functional strength data;
- Import the body model;
- Create the soft tissues;
- Merging the CAD model of the resistance training machine with the body model;
- Positioning of the body model on the resistance training machine;
- Adding the applicable constraints to the model;
- Adding motion agents to the model;
- Running the equilibrium simulation;
- Running the inverse-dynamics simulation;
- Preparing the model for dynamic simulation;
- Running the parametric analysis;
- Completing a literature search on the relevant resistance training exercise as well as relevant literature on safe loading limits;

- Interrogating the results; and
- Concluding research findings.

Slight variations in the modelling procedure were necessary in order to complete the protocol successfully for each piece of resistance training equipment which is discussed under the implementation of the evaluation protocol.

7.1.2 To implement the evaluation protocol on four pieces of resistance training equipment.

The 3D musculoskeletal modelling protocol was applied to four pieces of resistance training equipment, namely the:

- Seated biceps curl;
- Abdominal crunch;
- Seated row; and
- Chest press

Each piece of equipment presented unique challenges. In three of the four studies (seated biceps curl, seated row and chest press resistance training exercises) the default model of the modelling software was not adequate to solve the forward dynamics simulations and thus adjustments had to be made to the default model in order to complete the modelling process. In order to solve this problem for the seated biceps curl resistance training exercise the following adjustments were made to the default model: 1) an increase in 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. For the seated row and chest press resistance training exercises the joint angulations were recorded during the inverse dynamics simulations and the resulting joint torques were used to drive the model in the forward dynamics simulations. Unfortunately, as a result for these two exercises no muscle force or contraction results were obtained which impacted negatively on the value of the results received for the analysis of the exercises. Additional challenges were encountered using the default model in the modelling process which is discussed under the limitations of the evaluation protocol.

7.1.3 Identify potential risk for musculoskeletal injury.

The modelling process by means of the LifeModeler^{TM} software was able to identify some potential risk for musculoskeletal injury (Table 7.1). The abdominal crunch resistance training exercise demonstrated the most significant potential for risk for injury when performing the exercise. Unacceptable thoracic and lumbar spine joint compression as well as anterior/posterior A/P shear forces was recorded during the simulations and thus this exercise appears to 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. High lumbar A/P shear forces were also recorded for the seated biceps curl resistance training exercise which also alluded to potential excessive strain to the low back. Furthermore, the wrist joint and cervical spine were identified as vulnerable areas when exercising on the chest press machine due to the results obtained during the chest press simulations. No substantial risk was identified for the seated row resistance training exercise when appropriate positioning, good exercise technique and a suitable resistance is used by the exerciser. Therefore, the modelling process does appear to be able to identify some potential risk for injury however to gain considerable value from the information obtained from the modelling process regarding injury risk it is necessary to have knowledge of safe loading limits to make an informative comparison. Such information is only available for a small number of tissues and loading regimes (e.g. lower back motion segments in compression). Another point of reflection is the fact that most of the available literature on tissue-level stresses is from research conducted on occupational activities.

In addition, the modelling does not take into consideration varying training status or muscular strength and endurance of individuals which could either increase or decrease the individuals risk for injury depending on which side of the continuum they find themselves. The repetitive nature of exercise is also an essential element that should be considered in order to suitably evaluate the safety of an exercise.

Table 7.1: Potential risk for musculoskeletal injury while performing exercise on the 4 resistance training machines.

Resistance training machine	Injury risk areas	Recorded maximal values	Safe loading limits
Seated biceps curl	Lumbar spine (A/P shear)	5^{th} percentile female: 906.0 N	1000 N
		50^{th} percentile male: 1109.0 N	1000 N
		95^{th} percentile male: 1180.7 N	1000 N
	Extended periods of maximum muscle tension (Biceps brachii long head)	5^{th} percentile female: 329.5 N	-
		50^{th} percentile male: 267.7 N	-
Abdominal crunch	Thoracic spine (compression)	5^{th} percentile female: 11043.0 N	3400 N
		50^{th} percentile male: 4206.4 N	3400 N
	Lumbar spine (compression)	95^{th} percentile male: 4673.9 N	3400 N
		5^{th} percentile female: 12580.2 N	3400 N
	Thoracic spine (A/P shear)	50^{th} percentile male: 3388.6 N	3400 N
		95^{th} percentile male: 3664.2 N	3400 N
		5^{th} percentile female: 5827.9 N	1000 N
		50^{th} percentile male: 3201.3 N	1000 N
		95^{th} percentile male: 3067.0 N	1000 N
	Lumbar spine (A/P shear)	5^{th} percentile male: 5122.3 N	1000 N
		50^{th} percentile male: 559.9 N	1000 N
		95^{th} percentile male: 436.8 N	1000 N
Seated row	No vulnerable areas identified	-	-
Chest press	Cervical spine (compression)	5^{th} percentile female: 590.2 N	3400 N
		50^{th} percentile male: 538.0 N	3400 N
		95^{th} percentile male: 1248.0 N	3400 N
	Wrist (torque)	5^{th} percentile male: 6.5 Nm	13.8 Nm
		50^{th} percentile male: 3.3 Nm	13.8 Nm
		95^{th} percentile male: 2.7 Nm	13.8 Nm

7.1.4 Make recommendations on how the equipment could be improved with regards to design in order to maximise safety and exercise efficacy.

It was once again demonstrated from this research effort that the anthropometric dimensions of the end-user must be taken into account when designing exercise equipment. Two of the resistance exercise machines evaluated could not accommodate the anthropometric dimensions of the small (5th percentile) female namely; the seated biceps curl and the abdominal crunch resistance training machines. This discrepancy between her anthropometric dimensions and the machines engineered or manufactured adjustability appeared to place her at significant risk for a spine injury, specifically on the abdominal crunch resistance training machine. Further, as a result of anthropometric discrepancy it seemed as if the exercise (abdominal crunch resistance training exercise) was not successful at isolating her abdominal muscles thereby reducing the effectiveness of the exercise.

Therefore, results of this study indicate that the manufacturer of the resistance training exercise equipment used for this study has managed to accommodate the average as well as the very large end-user but not individuals with small anthropometric dimensions such as small adults or children. If these individuals exercise on the equipment they will not be able to adjust the equipment for optimal exercise posture and movement and therefore may not get the full benefits of the exercise or worse injure themselves.

Small alterations such as making an adjustable preacher curl “platform” for the seated biceps curl machine or adapting the foot rest for the abdominal crunch machine may contribute significantly to improving the overall design of the resistance exercise machines and therefore the safety and the efficacy of the end-user. Therefore, it appears as if the 3D musculoskeletal modelling protocol has the potential to make some recommendations regarding improvements in the design of the exercise training equipment.

Table 7.2: Recommended resistance training equipment design alterations.

Resistance training exercise	Problems identified regarding equipment design	Recommended alterations to equipment design
Seated biceps curl	Preacher curl "platform" not parallel with the seat	Alignment of preacher curl "platform" adjusted to ensure it is parallel with the seat
	Non-adjustable preacher curl "platform"	Manufacture an adjustable (height) preacher curl "platform"
	No foot rest	An adjustable foot rest
	Seat - limited adjustable range	Increase range of adjustments for seat
Abdominal crunch	Non-adjustable foot rest (height)	Manufacture an adjustable (height) foot rest
	Non-adjustable crunch pad/cushion (height)	Manufacture adjustable (height) crunch pad/cushion)
	Seat - limited adjustable range	Increase range of adjustments for seat
Seated row	No problems identified	Adjustable foot rest may be beneficial
Chest press	No problems identified	Adjustable foot rest may be beneficial

7.1.5 Make recommendations regarding limitations of the evaluation protocol. Evaluate if the protocol is sensitive enough to highlight injury risk and limitations in equipment design.

The 3D musculoskeletal modelling was able to highlight some interesting design elements and flaws as well as biomechanical and anthropometrical limitations of the evaluated resistance training machines. Thus, 3D musculoskeletal modelling can certainly be used to evaluate resistance training equipment design however the limitations as indicated by this study must be taken into consideration especially when using default models.

The following problems were encountered with the default model in the modelling process:

- The primary limitation of the default model of the software is that it lacks adequate bio-fidelity.
- The modelling can be a fairly time consuming process requiring a process of many iterations to be able to provide plausible results;
- Caution should be employed when using the default model to not assume that a matching anthropometry will result in reliable muscle strength capabilities;

this is further complicated by the significant variance in muscular strength between subjects of similar anthropometry due to differences in conditioning levels;

- To truly evaluate exercise efficacy all the important muscles that play a role in the movement should be present. It is possible to add muscles to the default model and then assess their relative contribution to the produced force (as a percentage of their maximal force generating capacity), however this can be time consuming and was not within the scope of this study;
- It is important to bear in mind when evaluating spine loads 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. Thus, individualised vertebra and corresponding joints might produce different results; and
- It appears that more complex, multi-joint or compound exercises such as the chest press or seated row resistance training exercise pose a problem for the default model and therefore models with more detailed musculature may be required to solve the forward dynamics simulations sufficiently.

Therefore, the modelling process is a reasonably time intensive method even when adjustments do not have to be made to the default model as it is a process that requires many iterations to get the ideal musculoskeletal model completed and positioned appropriately on the resistance training equipment and when additional adaptations have to made to the model it does not always prove a practical solution for the evaluation of the equipment. Furthermore, adequate training and in-depth knowledge of the software as well as biomechanical and functional anatomy expertise is essential not only to perform the modelling but to analyse the results. However, it is important to bear in mind that despite the challenges involved with the modelling process in comparison to other methods it is still a relatively simple, inexpensive and safe means for evaluating resistance training equipment design.

7.2 RECOMMENDATIONS FOR RESEARCH AND PRACTICE

Further the following recommendations for future research:

- Unless significant improvements are made to the default model of the Lifemodeler™ software. The application of the software using the default model is limited in terms of evaluating resistance training equipment. Further the iterations that need to be made during the modelling process to ensure valid and reliable results is time consuming and thus a more user-friendly and adequate software would also lend the process more practical for the evaluation of exercise equipment. A more detailed model than the default model may solve some these issues;
- Expertise with regards to the software as well as functional anatomy, biomechanics and exercise technique is crucial to make the evaluation process valuable;
- Ideally, norms and data on safe loading limits and injury risk should be established for a larger variety of tissue structures and loading regimes. Further, if more data was available on safe loading during exercises rather than only occupational activities would also increase the usefulness of the information obtained from the modelling process; and
- Evaluations and comparisons should be made with resistance training equipment from different manufacturers as well as exercises that train similar muscle groups. Valuable information can be gained regarding the safety and exercise efficacy between the exercises or pieces of equipment and recommendations can be made regarding exercise prescription.

7.3 GENERAL CONCLUSIONS

Accurate assessment of the risk of injuries and performance efficacy during exercise training and occupational activities as well as subsequent design of exercise equipment, effective prevention and treatment programmes depend, amongst others, on an accurate estimation of biomechanical and anthropometric considerations of the end-user. Such knowledge can be acquired primarily by 3D musculoskeletal modelling as experimental attempts remain invasive, costly and limited. Unfortunately, currently 3D musculoskeletal modelling is still a fairly time consuming process requiring a process of many iterations in order to perform the modeling and providing plausible results. However, it is continually being improved and thus the limitations will hopefully be addressed thereby making the process of 3D musculoskeletal modelling more user-friendly and effective in evaluating various pieces of exercise equipment and thus ensuring the safety and efficacy of the exercise for the end-user. To conclude, it would appear as if the research hypothesis was proven correct. 3D musculoskeletal modelling is certainly the way of the future and will eventually probably play a major role in the design of most exercise equipment.