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CHAPTER 3

THREE DIMENSIONAL MUSCULOSKELETAL MODELLING OF THE SEATED BICEPS CURL RESISTANCE TRAINING 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 is effective in assessing the safety and efficacy of exercising on a seated biceps curl 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 biceps curl machine was set at fifty percent of the functional strength one repetition maximum (1RM) for each anthropometric case, two repetitions were performed. Results indicated that the LifeModeler™ default model was not adequate to solve the forward dynamics simulations and therefore adjustments had to be made to the default model to successfully complete the forward dynamics simulations. The software was able to indicate the anthropometric differences with regards to the biceps curl machine's engineered adjustability as the 5th percentile female was accommodated poorly on the machine. However, the poor positioning of the small female did not appear to put her at increased risk for injury in comparison to the other two anthropometric cases. High recorded lumbar spine anterior/posterior (A/P) shear forces for the three anthropometric cases and maximum muscle tensions for the female and 50th percentile male indicate that the seated biceps curl exercise may pose a risk for injuries. To conclude, it appears as if 3D musculoskeletal modelling can be used to evaluate resistance training equipment such as the seated biceps curl machine however the limitations as indicated by this study must be taken into consideration especially when using the default Lifemodeler^{TM} model.

Keywords: *Resistance training equipment, seated biceps curl, biomechanics, anthropometric, modelling, Lifemodeler^{TM}, inverse dynamics, forward dynamics*

Introduction

Resistance training refers to a method of conditioning designed to overload the musculoskeletal system, leading to accelerated enhancement of muscle strength (Fleck and Kraemer, 1997).Traditionally, resistance training was used primarily by adult athletes to enhance sport performance and increase muscle size. Today, it is recognized as a method of enhancing the health and fitness of men and women of all ages and abilities (Howley, 2007). The popularity of resistance training is clearly evidenced by the extensive growth of fitness centres and sales of resistance exercise equipment for home use (Vaughn, 1989; Lou *et al*., 2007).

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.

Mathematical and computer modelling is suitable for a wide variety of applications such as the design, production and alteration of medical equipment (prostheses, orthopaedic and orthodontic devices) as well as sports and training equipment (Alexander, 2003; Kazlauskiené, 2006). With the capability to simulate 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. In addition, computer simulation models permit the study of the complex interactions between biomechanical variables (Kenny *et al*., 2005). Thus the primary aim of this study was to determine the efficacy of a default 3D musculoskeletal model in evaluating resistance training equipment design.

In a series of articles 3D musculoskeletal modelling, with a focus on biomechanical and anthropometric variables, will be used to evaluate four commonly used pieces of resistance training equipment in order to assess the suitability of this method for exercise equipment design evaluation. This study presents the musculoskeletal modelling of three anthropometric cases while exercising on a commercially available seated biceps curl resistance training machine.

The biceps curl exercise is a commonly used, predominantly single joint open-kinetic-exercise used to isolate the biceps muscles. The Biceps brachii, from which the exercise derives its name, Brachialis and Brachioradialis muscles contribute most to this action, with assistance from the Pronator teres and wrist flexor group (Durall, 2004; Reiser *et al*., 2007). There are many variations of the traditional biceps curl exercise using dumbbells, barbells and machines. Incline dumbbell curls and dumbbell preacher curls are two variations of the standard dumbbell biceps curl generally applied to optimize Biceps brachii contribution during elbow flexion by fixing the shoulder angle at a specific position. These different protocols may impose different demands to the neuromuscular system, resulting in different solutions for the load sharing between elbow flexors (Oliveira *et al*., 2009). The biceps curl exercise regardless of variation can be divided into two phases: (1) lifting phase to flexed position and (2) lowering phase to extended position (Floyd, 2009).

Methods

Equipment

A 3D musculoskeletal full body model was created using LifeModeler™ software and incorporated into a multibody dynamics model of the seated biceps curl 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 *et al*., 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 consisted 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. 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).

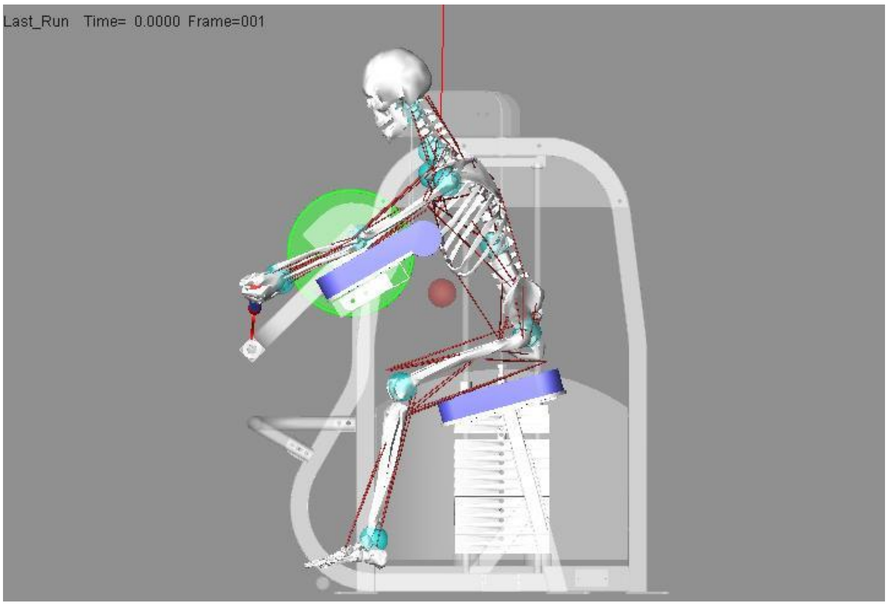

Figure 1. 3D musculoskeletal modelling of the biceps curl resistance training machine and 50th percentile male musculoskeletal model using LifeModeler^{TM} and MSC ADAMS software.

Musculoskeletal full body human and seated biceps curl 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 seated biceps curl resistance training machine was obtained from a South African exercise equipment manufacturing company. 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 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 seated biceps curl 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 arms at the left and right humerus, as well as the upper torso at the sternum to the preacher curl "platform" and spherical joints were used to connect the hands to the handle bars of the biceps curl machine. Bushings were also used in order to secure the lower torso to the seat of the exercise 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.

Table I. Starting exercise posture for the 3 anthropometric cases on the biceps curl resistance training machine. Results are presented for the sagittal, transverse and frontal planes (degrees). Note that F = flexion, E = extension, ER = external rotation and AB = abduction.

Joint	5^{th} percentile female	50^{th} percentile male	95^{th} percentile male
Scapula	0.0; 0.0; 0.0	0.0; 0.0; 0.0	0.0; 0.0; 0.0
Shoulder	85.5(F); 0.0; 0.0	85.5(F); 0.0; 0.0	85.5(F); 0.0; 0.0
Elbow	15.0(F); 0.0; 0.0	8.0(F); 0.0; 0.0	8.0(F); 0.0; 0.0
Wrist	0.0; 60.0(ER); 0.0	0.0; 60.0(ER); 0.0	0.0; 60.0(ER); 0.0
Hip	45.0(F); 0.0; 8.0(AB)	62.0(F); 0.0; 8.0(AB)	62.0(F); 0.0; 8.0(AB)
Knee	70.0(F); 0.0; 0.0	50.0(F); 0.0; 0.0	50.0(F); 0.0; 0.0
Ankle	13.0(E); 0.0; 0.0	13.0(E); 0.0; 0.0	13.0(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	30.0(F); 0.0; 0.0	30.0(F); 0.0; 0.0	32.0(F); 0.0; 0.0

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 biceps curl 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 1.41 seconds and the eccentric phase slightly longer at 2.84 seconds to mimic conventional resistance training technique in which the downward phase is more deliberate to prohibit the use of momentum. The 1.41 second concentric phase included a STEP function 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.

After the inverse dynamics simulation was performed, the rotational motion was removed from the rotational joint of the lever arm of the seated biceps curl 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). 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.

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 seated biceps curl resistance training machine. Key aspects included start and end exercise posture as well as maintaining correct exercise technique throughout the exercise during the simulation. Start and end exercise posture evaluation entailed positioning of the axilla on the top of the preacher curl "platform" as to support the back of the upper arms, alignment of the elbow joint with the axis of rotation of the machine, hip flexion between 80 – 90 degrees and a knee angle of approximately 90 – 100 degrees. The feet are supposed to be positioned flat on the ground. Correct technique was assessed in terms of limited compensatory movements and performing the biceps curl through the full range of motion as determined by the inverse dynamics.

The kinematic and kinetic data from the simulations were analysed specifically in terms of peak muscular force production of the prime movers of the seated biceps curl. Thus for the purpose of this study, efficacy of the equipment was assessed by evaluating whether the equipment exercised the muscles it was designed for, does the biceps curl machine exercise the prime flexors of the elbow joint? 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 (since no loading limits were found for the elbow joint) found in the literature as well as the peak muscular forces for the prime flexors of the elbow. Injury risk to both these structures are real especially when lifting excessive masses and or during execution of exercise with poor postures.

The statistical analyses of the results were completed using the STATISTICA© software package (Statsoft). Due to the nature of this study basic descriptive statistics were performed and a Pearson's product moment correlation coefficient was used to determine relationships between appropriate variables. Statistical significant differences were indicated by a p-value of less than 0.05.

Results

Table II presents the body mass and stature of the three anthropometric cases based on BMI data obtained from RSA-MIL-STD 127 Vol 1 (2004). Table III presents the external resistance the models had to overcome during the forward dynamics simulations, 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	12
50th percentile male	22
95th percentile male	35

The LifeModeler™ default model was not adequate to solve the forward dynamics simulations for any of the anthropometric cases. In order to solve this problem the following adjustments were made to the default model: 1) an increase in the 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.

Muscle force production (N), contraction (shortening and lengthening) (mm) and joint torque (Nm) for the right side are reported on. Theoretically, the results of the left and right side should be similar however this could have been slightly influenced by the alignment of the seat and preacher curl “platform”.

Force production of the Biceps brachii short head (BBS) and Biceps brachii long head (BBL) and the Brachialis (B) muscles are presented in Table IV. The peak force production is the highest for the BBL in comparison to the BBS in all the anthropometric cases. The peak B muscle force production was less than either the BBS or BBL for all the anthropometric cases except for the 95th percentile male whose peak B muscle force production was equal to his BBS muscle force production. The 5th percentile female exerted the highest force for all muscles followed by the 50th percentile male and lastly the 95th percentile male.

Table IV. Right Biceps brachii and Brachialis muscles force production (N) results for the 3 anthropometric cases.

Musculoskeletal model	Muscle	Mean (N)	Min.	Max.
5th percentile female	Biceps brachii short head (BBS)	255.5	-8.8	268.9
	Biceps brachii long head (BBL)	235.5	-9.5	329.5
	Brachialis (B)	152.9	-6.7	215.1
50th percentile male	Biceps brachii short head (BBS)	209.7	48.9	221.5
	Biceps brachii long head (BBL)	225.3	135	267.7
	Brachialis (B)	166.5	52.9	172.6
95th percentile male	Biceps brachii short head (BBS)	205.8	2.9	172.3
	Biceps brachii long head (BBL)	60.4	-4.1	215.9
	Brachialis (B)	149.9	0.2	172.3

Absolute muscle contraction results for the BBS, BBL and B muscles are presented in Table V. The mean muscle length is greatest for the BBS in comparison with the BBL for all the anthropometric cases. Furthermore, the maximum, minimum and mean muscle lengths are smaller for the B muscle in comparison to both the heads of the BB muscle for all three anthropometric cases. The mean muscle contraction length for all the muscles is greatest for the 95th percentile male and smallest for the 5th percentile female.

Table V. Right Biceps brachii and Brachialis absolute muscles contraction (mm) results for the 3 anthropometric cases.

Musculoskeletal model	Muscle	Mean (mm)	Min.	Max.
5th percentile female	Biceps brachii short head (BBS)	239.2	228.9	253.9
	Biceps brachii long head (BBL)	217.0	206.8	235.9
	Brachialis (B)	105.6	103.5	112.0
50th percentile male	Biceps brachii short head (BBS)	300.6	281.1	315.8
	Biceps brachii long head (BBL)	274.8	253.8	294.3
	Brachialis (B)	131.9	122.6	142.5
95th percentile male	Biceps brachii short head (BBS)	330.5	307.5	349.5
	Biceps brachii long head (BBL)	303.9	280.7	325.3
	Brachialis (B)	143.3	129.8	156.5

Due to the involvement of wrist and elbow joints in the biceps curl exercise, torque for these joints is presented in Table VI. The mean wrist torque is lower than the mean elbow torque for all three the anthropometric cases. Furthermore, the torque values for both joints are lowest for the 5th percentile female and highest for the 95th percentile male.

Table VI. Right wrist and elbow joint 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	0.8	2.8	3.9
	Elbow	3.7	-28.3	11.6
50th percentile male	Wrist	1.9	-4.2	3.7
	Elbow	8.1	5.4	17.7
95th percentile male	Wrist	3.4	0.2	6.2
	Elbow	12.6	1.8	25.3

The length (contraction) of the BBL muscle was shortest at approximately 1.6 s and 5.6 s (Figure 2). The correlation between sagittal elbow joint angle and joint torque was statistically significant ($p < 0.05$) for all three anthropometric cases (5^{th} percentile female: r = -0.87, 50^{th} percentile male: r = -0.87, 95^{th} percentile male: r = -0.98). Therefore as the muscles shortened and the elbow joint angle decreased the joint torque increased.

Maximum elbow joint torque production was produced at approximately 1.6 s and 5.6 in the three anthropometric cases (Figure 3 and 4).

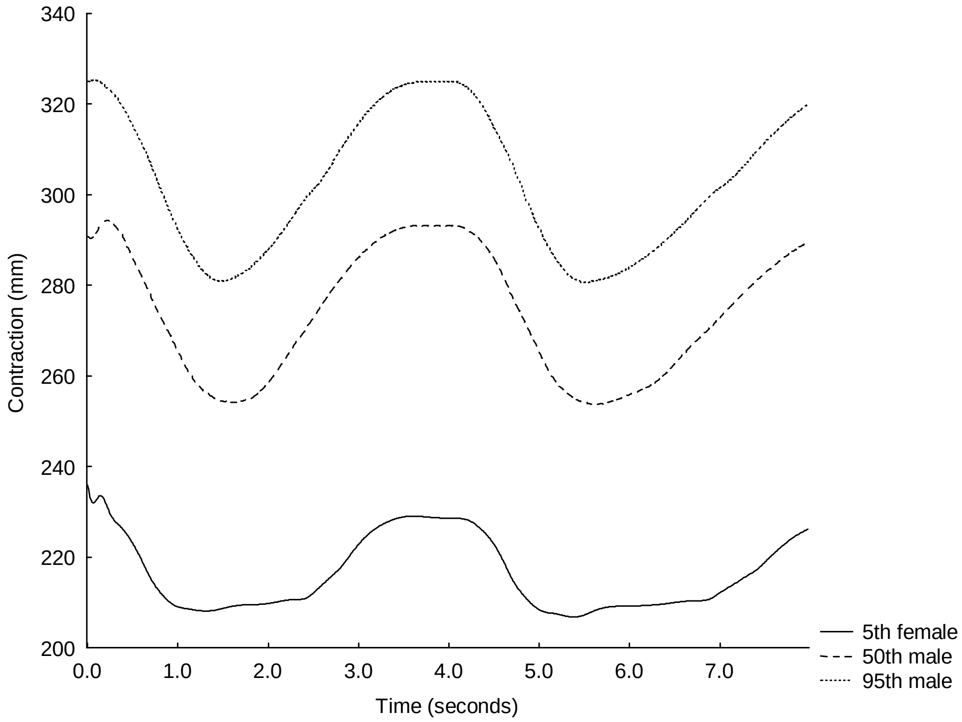

Figure 2. Long head of the Biceps brachii contraction (mm) for the 3 anthropometric cases (2 repetitions).

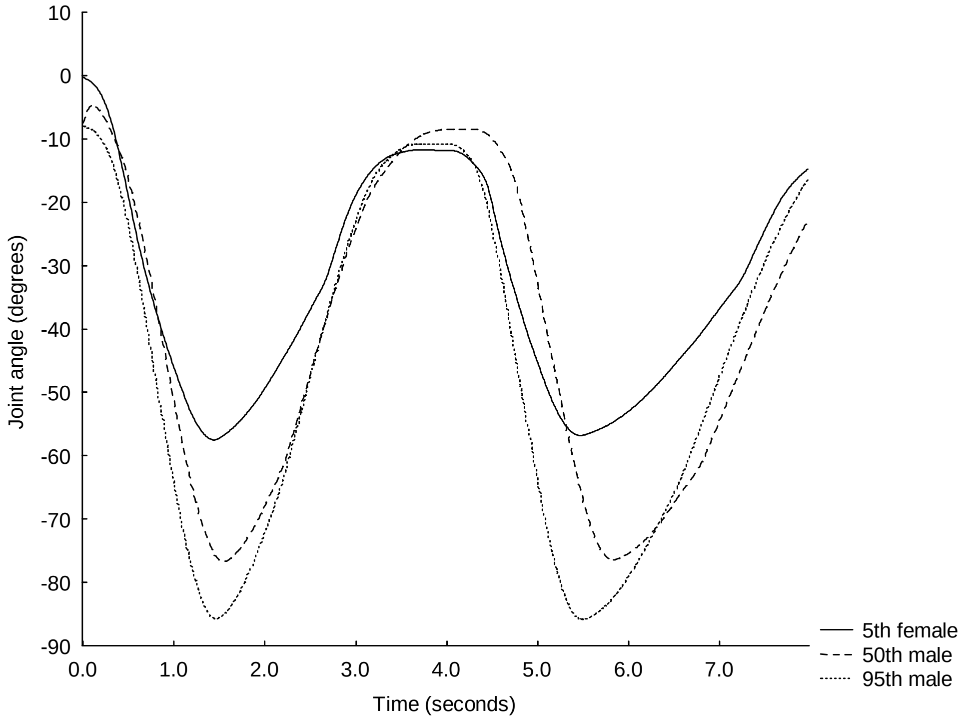

Figure 3. Sagittal elbow joint angle (°) for the 3 anthropometric cases (2 repetitions). Note: negative joint angle indicates elbow flexion.

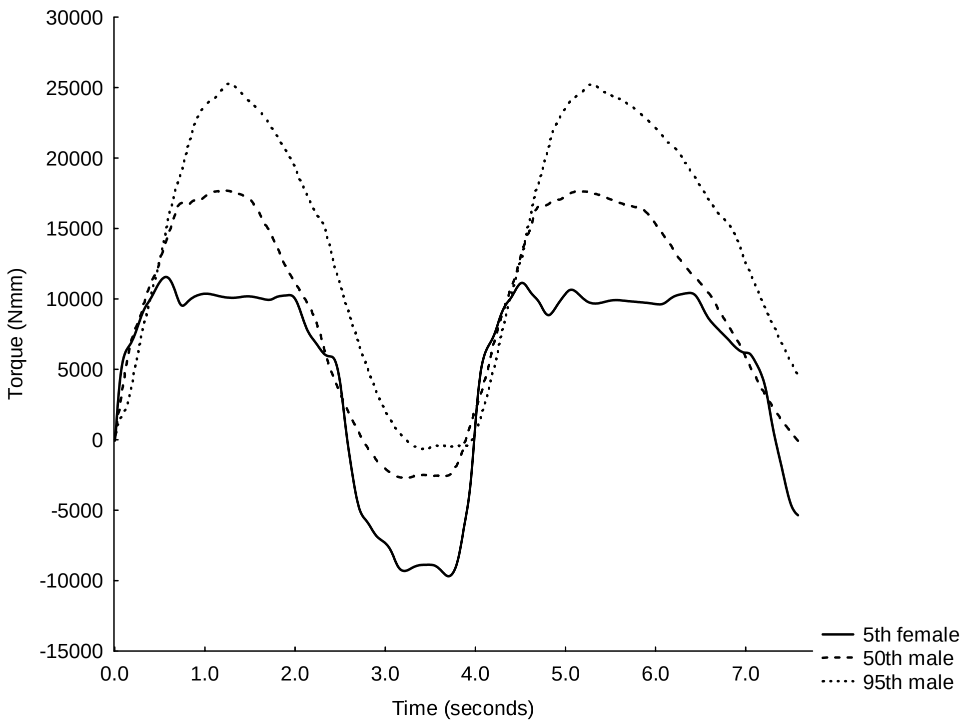

Figure 4. Elbow joint torque (Nmm) for the 3 anthropometric cases (2 repetitions).

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 Tables VII and VIII respectively. Peak thoracic spine joint compression forces were greatest for the 95th percentile male, followed by the 50th percentile male and were lowest in the 5th percentile female. There was a similar trend for the peak lumbar spine joint compression forces except that the 50th percentile male's compression force was slightly higher than the 95th percentile males. In all anthropometric cases the peak lumbar spine joint compression forces were greater than the peak thoracic spine joint compression forces.

Table VII. 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	Thoracic spine	827.9	-508.2	1774.1
	Lumbar spine	1175.8	-608.0	2337.2
50th percentile male	Thoracic spine	996.3	-402.0	2123.3
	Lumbar spine	1559.7	-85.4	2920.5
95th percentile male	Thoracic spine	971.2	-226.0	2133.2
	Lumbar spine	1466.0	-160.7	2821.7

Peak A/P lumbar joint shear forces were greater than peak A/P thoracic joint shear forces for the three anthropometric cases. The 5th percentile female recorded the lowest peak A/P lumbar and thoracic joint shear forces, followed by the 50th percentile male and the 95th percentile male recorded the highest peak shear forces.

Table VIII. 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	Spinal joint	Mean (N)	Min.	Max.
5th percentile female	Thoracic spine	-358.6	-736.8	209.5
	Lumbar spine	461.3	-906.0	179.1
50th percentile male	Thoracic spine	-402.0	-901.0	223.0
	Lumbar spine	-544.9	-1109.0	117.2
95th percentile male	Thoracic spine	-440.7	-974.3	138.7
	Lumbar spine	-569.8	-1180.7	96.0

Discussion

The first conclusion that can be drawn from this study is that the LifeModeler™ default model was not adequate to solve the forward dynamics simulations for any of the anthropometric cases. In order to solve this the following adjustments were made to the default model: 1) an increase in the pCSA of the three default elbow flexor muscles, 2) manipulate the muscle origins and insertions and 3) decrease the joint stiffness in the forwards dynamics simulations.

Muscle tension depends on several factors including neural activation, pCSA, muscle architecture and muscle length (Durall, 2004). The pCSA of the BBL, BBS and B muscles had to be increased for all three anthropometric cases by 50% (Table IX). Due to differences in measurement methodology anatomical cross-sectional areas (aCSA) are smaller than pCSAs (Akagi *et al.,* 2009). Despite this difference, the adjustments resulted in significantly larger pCSAs when compared to elbow flexor muscle anatomical cross-sectional area (aCSA) measurements by magnetic resonance imaging (MRI) in young males and females of 182 mm^2 and 103 mm^2 respectively (Akagi *et al.,* 2009). These adjustments however were necessary in order to solve the forward dynamics simulations. It is interesting to note that the pCSA area for the 50th percentile male was larger than that of the 95th percentile male for both muscle groups. The apparent reasoning for this discrepancy according to the manufacturers of the software has to do with the proportionality of the volume differences between the two cases. The 95th percentile male is 146mm taller but the increase in body mass was only 6kg therefore there was approximately a 9% increase in height with only a 9% increase in volume. To keep proportionality, volume should increase three times more than stature. Thus, 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.

Table IX. Physiological cross-sectional area (pCSA) after adjustments (mm^2) for the 3 anthropometric cases.

Musculoskeletal model	Biceps brachii short head	Biceps brachii long head	Brachialis
5^{th} percentile female	147.2	180.5	116
50^{th} percentile male	178.7	218.8	139.7
95^{th} percentile male	177.6	217.4	138.9

The muscle origin and insertion points of the muscles also had to manipulated in order to increase the moment arm and therefore allow greater torque to be produced around the elbow joint. Considering that the literature suggests considerable individual variation in muscle origin and insertion locations (El-Naggar, 2001; Ramesh *et al.,* 2007) the adjustments were deemed anatomically reasonable. For instance Ramesh (2007) found that the sternocleidomastoid muscle varies much in the extent of the origin from the clavicle. In some cases the clavicular head may be as narrow as the sternal, in others it may be as much as 7.5cm in breadth. Due to this variability in human anatomical structure an individual whose tendons are inserted on the bone further from the joint centre should be able to lift heavier weights because of the longer moment arm (Beachle and Earle 2008). Moment arms for muscles are generally quite small, usually in the order of several centimetres, and change with joint angle. The moment arm of the BB muscle is smallest at the extremes of the elbow joint range of motion and largest within the midrange. Because moment arm profiles of all flexor muscles are not identical, not all muscles will contribute similarly to the exercise (Reiser *et al*., 2007). The origin of the BBS muscle was relocated 50mm superiorly and 10mm medially from the default position. While the origin of the BBL muscle was relocated 10mm superiorly and medially from the default position. Insertions of both the heads were moved 20mm distally from the default position. It should be noted that this influenced the contraction results of the muscles and therefore the BBS muscle mean length was longer than the BBL muscle.

Lastly, the joint stiffness was reduced during the forward dynamics simulation only. Joint stiffness during inverse dynamics (default model) simulations is artificially increased solely for the purpose of ensuring high quality kinematics. One could argue that this is a plausible adjustment as in reality healthy joints experience minimal joint stiffness and therefore the joint stiffness was decreased to finite levels through various iterations until acceptable kinematics was achieved. Even after the adjustments the 5^{th} percentile female and the 50^{th} percentile male BBL muscle reached their maximum force production as can be seen in Figure 5. A possible reason for this could be that the biceps curl machine design does not accommodate the anthropometric dimensions of the 5^{th} percentile female and the 50^{th} percentile male as well as that of the 95^{th} percentile male. A discrepancy with regards to the alignment of the elbow joint with the axis of rotation of the lever arm could result in a disproportionately higher relative muscle force production required to overcome the external resistance. This could result in the muscles reaching maximal force production for extended periods of time which is undesirable in terms of muscular injury risk.

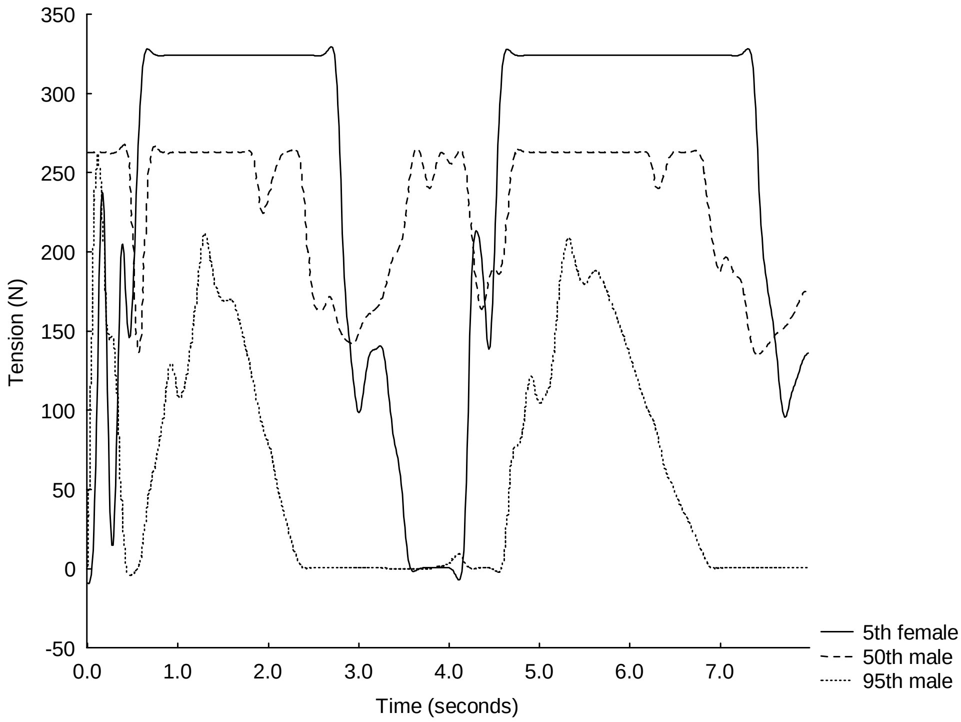

Figure 5. Biceps brachii long head force production (N) for the 3 anthropometric cases (2 repetitions).

The second conclusion of this study was that the software was able to sufficiently indicate anthropometric differences with regards to the biceps curl 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 biceps curl machine except for the 5th percentile female (Figure 6). The small female's feet could not reach the ground and her elbow joint could not be aligned properly with the axis of rotation of the machine despite maximum adjustments to the seat. The commercially available machine does not allow for manual adjustability of the preacher curl "platform". However the "platform" had to be adjusted within the modelling environment so that the small female could reach the handle bars of the biceps curl machine. These adjustments to the preacher curl "platform" would not be possible in reality and therefore should be an important design consideration for the manufacturer.

As a result the exercise postures of the 5th percentile female were negatively affected as opposed to the 50th and 95th percentile males. This deficiency in the adjustability of the equipment once again highlights the problem that not all equipment is equally fitted to all individuals and anthropometry differences should be taken into consideration when designing exercise equipment (Hamilton *et al*., 2009). Furthermore, if an individual is not accommodated appropriately on a piece of equipment exercise technique and posture can be negatively influenced. It was also noted when positioning the musculoskeletal models that the preacher curl “platform” was not parallel with the seat of the biceps curl machine. The fact that the misalignment of the seat and preacher curl “platform” was noted also alludes to the suitability of the modelling for determining such factors.

Lastly, with regards to the biomechanical evaluation in terms of exercise efficacy and injury risk the following could be deduced from the study. The Lifemodeler™ default model consisted only of the BB and B muscles. However, other muscles also play an important role in elbow flexion such as the Brachioradialis muscle (Table X). 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. In addition, comparisons should be made between variations in technique as well as different exercises for the same muscle groups or different manufacturer’s equipment for the same exercise in order to make an informed evaluation of the piece of equipment. The study did however show that the force production was greater for the BBS and BBL in comparison with the B muscle. Furthermore, 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, even although the resistance used for all three cases was proportionally calculated to correlate the anthropometric dimensions.

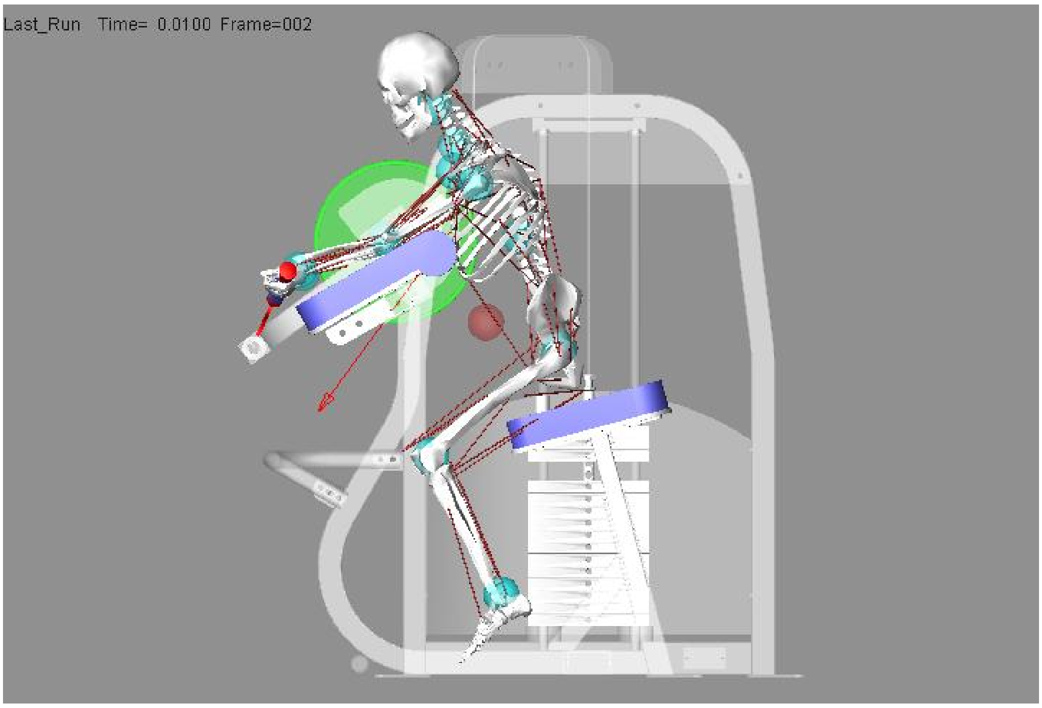

Figure 6. 5th percentile female's positioning on the seated biceps curl machine.

Table X. Biceps curl movement analysis (Floyd, 2009). Note: *primary elbow flexor muscle omitted from the default LifeModeler model.

Joint	Lifting phase to flexed position		Lowering phase to extended position	
	Action	**Agonists**	**Action**	**Agonists**
Wrist and hand	Flexion	Wrist and hand flexors (isometric contraction)	Flexion	Wrist and hand flexors (isometric contraction)
		Flexor carpi radialis		Flexor carpi radialis
		Flexor carpi ulnaris		Flexor carpi ulnaris
		Palmaris longus		Palmaris longus
		Flexor digitorum profundus		Flexor digitorum profundus
		Flexor digitorum superficialis		Flexor digitorum superficialis
		Flexor pollicus longus		Flexor pollicis longus
Elbow	Flexion	Elbow flexors (Concentric contraction)	Extension	Elbow flexors (eccentric contraction)
		Biceps brachii		Biceps brachii
		Brachialis		Brachialis
		*Brachioradialis		*Brachioradialis
		Pronator teres		Pronator teres

The joint torque values obtained for the wrist and elbow appear to be plausible when comparing the values to peak values obtained by means of isokinetic testing bearing in mind that the values obtained in this study were not from maximal tests. For example wrist flexion/extension values of 13.8 Nm and 12.7 Nm respectively at 60 degrees per second in non-disabled subjects (Van Swearigen, 1983) and elbow flexion/extension values of 36 Nm for both elbow flexion and extension at 60 degrees per second in female college basketball players (Berg *et al.,* 1985). Joint torque values for the elbow joints were higher than that of the wrist joints and the joint torques produced were also appropriate to the size of the anthropometric cases since torque can be quantitatively defined as the magnitude of a force multiplied by the length of its moment arm (Beachle and Earle, 2008). The results also indicate that as the sagittal elbow joint angle decreased the elbow joint torque increased. Muscle can produce maximum tension at or near their resting length because the greatest numbers of actin and myosin bonds are formed when the muscles are at this length. The resting position of the biceps brachii would theoretically occur when the elbow is bent roughly 75 degrees because the total arc of movement at the elbow is roughly 150 degrees, Thus, at 75 degrees of elbow flexion, the biceps brachii is midway between fully elongated and fully shortened (Durall, 2004). Interestingly in this study, the maximum joint elbow torques were reached at joint angles between approximately 55 degrees (5^{th} percentile female) and 85 degrees (95^{th} percentile male) (Figure 3). This corresponds favourably with the literature's proposal of 75 degrees. The maximum elbow torque production for all three anthropometric cases was at approximately 1.6 s and 5.6 s (Figure 4) which appears to correspond with the shortest BBL contraction (Figure 2). Although these results may appear contradictory it must be noted that the peak joint torques were reached with the muscles close to their shortest length during the exercise period which was indeed very close to the natural resting lengths for BB muscles and not necessarily equal to the shortest anatomical length of the muscle during the full range of motion of the joint.

While the differences in absolute muscle contractions (Table V) are to be expected Figure 7 indicates that relative muscle contraction as a percentage of starting muscle length was similar for the males but slightly less for the female. It could indicate that her range of motion might have been less during the forward dynamics simulation when compared to that of the two male models.

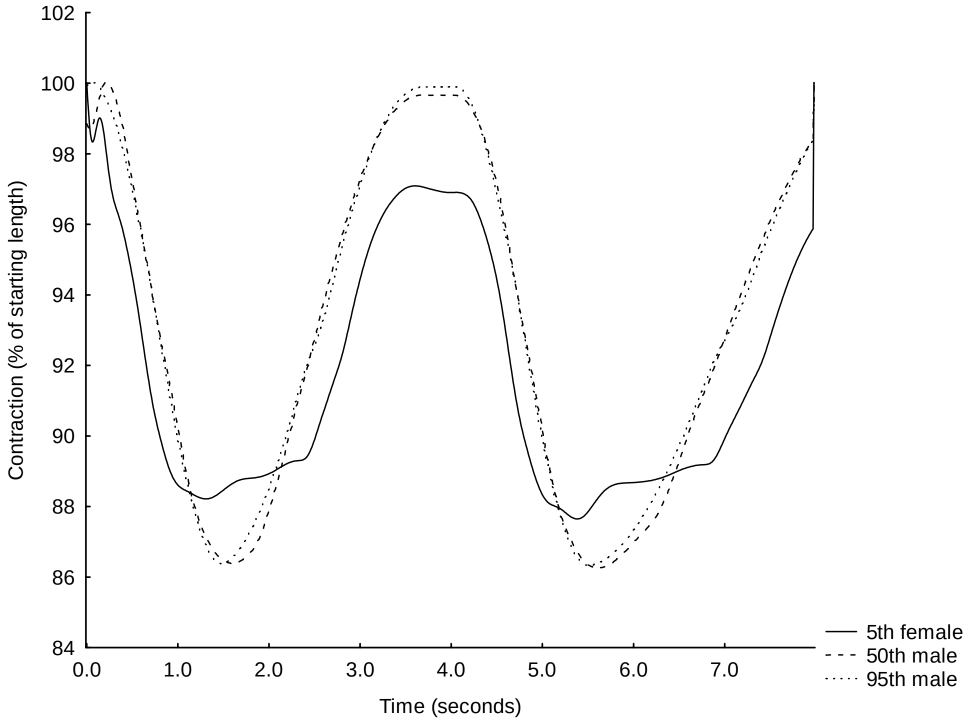

Figure 7. Biceps brachii long head contraction (mm) as a percentage of starting length for the 3 anthropometric cases (2 repetitions).

There are three load types: compression, tension, and shear. Tensile loads tend to pull the ends of a body apart, compressive loads tend to push the ends together, and shear loads tend to produce horizontal, or parallel, sliding of one layer over another (Whiting and Zernicke, 2008). In terms of risk assessment of musculoskeletal injury it was important to evaluate the compression and A/P shear forces of the thoracic and lumbar spine as the back is a common area for injury during exercise. In addition there is research that exists regarding the maximum recommended limits when performing various tasks thus making comparisons between recorded values and recommended limits possible. 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 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. All three anthropometric cases were below the recommended failure limit of 3.4 kN however were above 600 N and therefore could still be putting them at risk for injury.

The thoracic 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). However this was not the case for the lumbar spine joint A/P shear forces, the male anthropometric cases were both above 1000 N and the 5th percentile female was slightly below. Furthermore, the lumbar spine joint A/P shear forces were greater than the thoracic spine joint A/P shear forces for all the anthropometric cases.

It is important to note that although the compression (thoracic and lumbar spine) and thoracic spine joint A/P shear forces recorded were 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' estimated 1RM and therefore if exercisers use a resistance closer to their maximum the loading values may exceed the acceptable limits. The modelling also 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. Core musculature also plays an important role in protecting exercisers especially the back during training which is also not taken into account. The core can be defined as the lumbo-pelvic-hip complex. The core is where the centre of gravity is located and where all movement begins (Prentice, 2010a). The core operates as an integrated functional unit whereby the entire kinetic chain works synergistically to produce force, reduce force, and dynamically stabilize against abnormal force. In an efficient state, each structural component distributes weight, absorbs force, and transfers ground reaction forces (Prentice, 2010b).

While limited data exists on safe muscle tension values, due to large individual variability, the results of the muscle tensions for the 5th percentile female and 50th percentile male indicate that one of the prime movers of the elbow were strained above its maximum calculated capacity for extended periods during the exercise. This should be deemed to be a high risk for muscular injury during the exercise.

Conclusion

The 3D musculoskeletal modelling was able to indicate interesting design elements and flaws as well as biomechanical and anthropometrical limitations of the evaluated seated biceps curl resistance training machine. It has therefore once again been demonstrated that the anthropometric dimensions of the end-user must be taken into account when designing exercise equipment. It must be noted however, contrary to what was expected poor positioning of the small female did not appear to put her at increased risk for injury in comparison to the other two anthropometric cases who were adequately accommodated by the biceps curl resistance training machine. High recorded lumbar spine A/P shear forces for the three anthropometric cases and maximum muscle tensions for the female and 50th percentile male indicate that the seated biceps curl exercise may pose a risk for injuries. 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 lacking adequate bio-fidelity. Mathematical and computer modelling is continually being improved and thus the limitations will hopefully be addressed thus making the process of 3D musculoskeletal modelling more user-friendly and effective in evaluating various pieces of equipment and thus ensuring the safety and efficacy of the exercise for the end-user. Unfortunately, currently it is still a fairly time consuming procedure requiring a process of many iterations in order to perform the modelling and provide plausible results. However an important benefit of 3D musculoskeletal modelling that should not be forgotten is fact that it is a relatively inexpensive manner of evaluating resistance training equipment design and can be performed without putting the subject at risk of injury.

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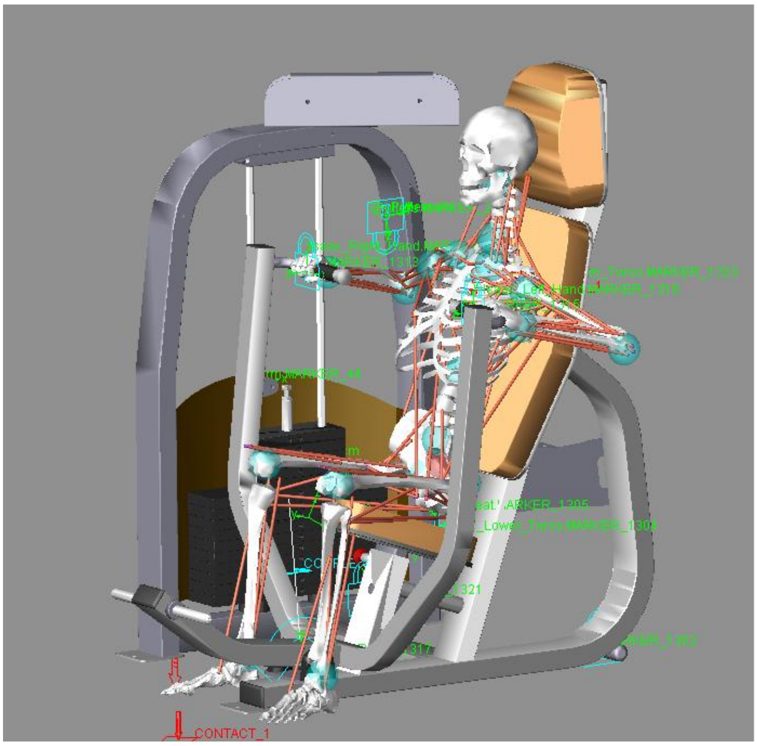
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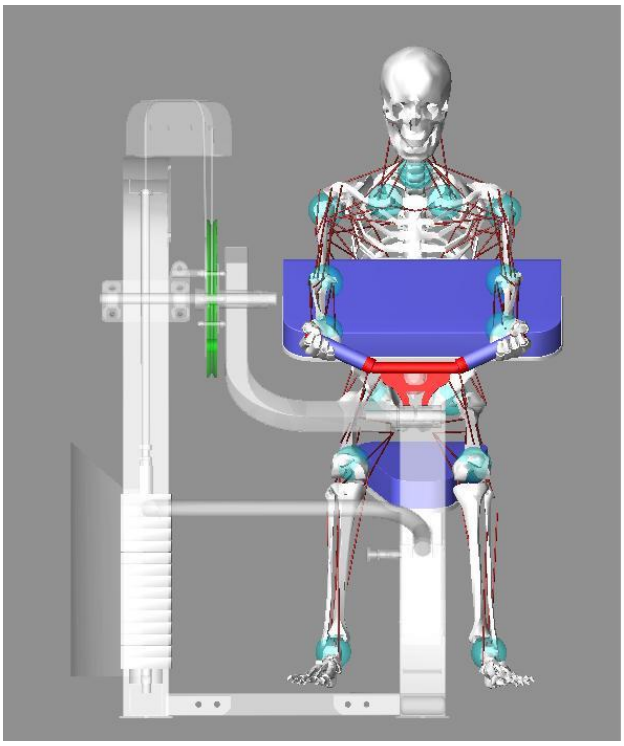

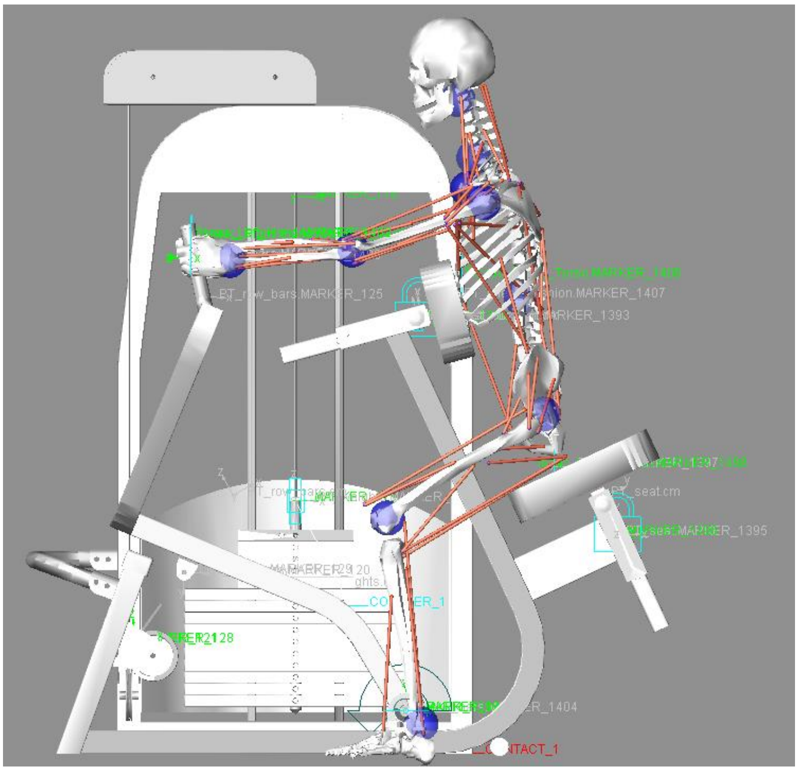
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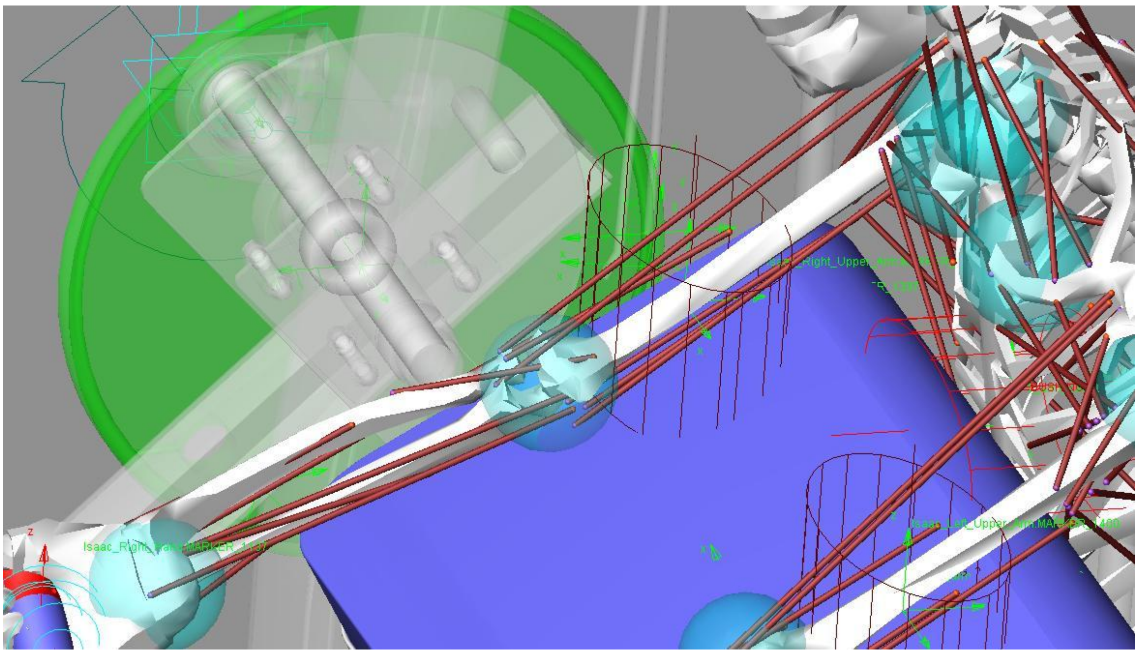

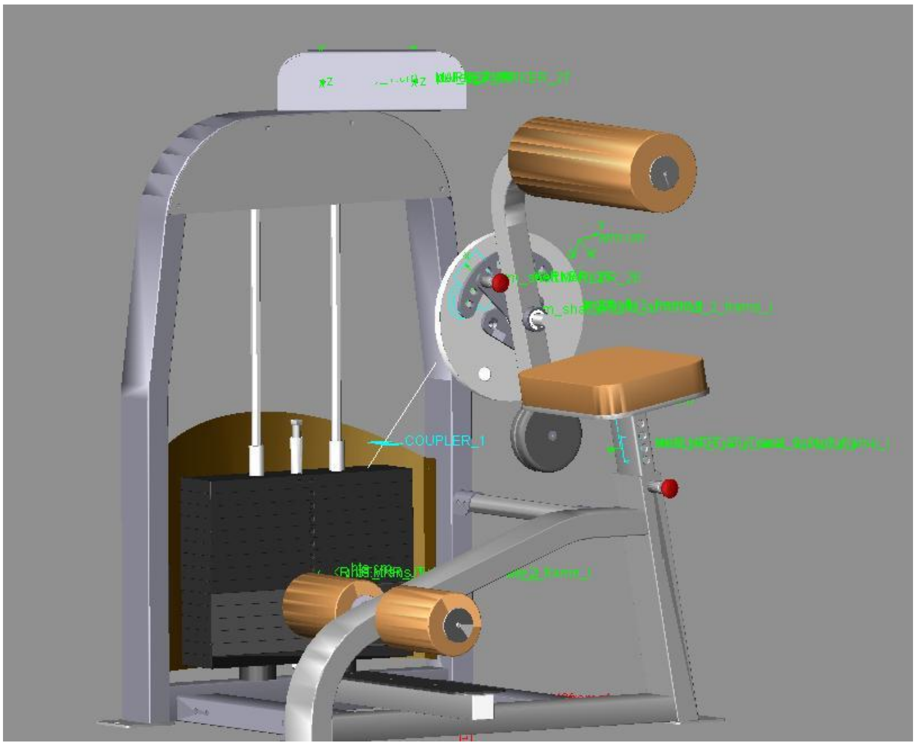
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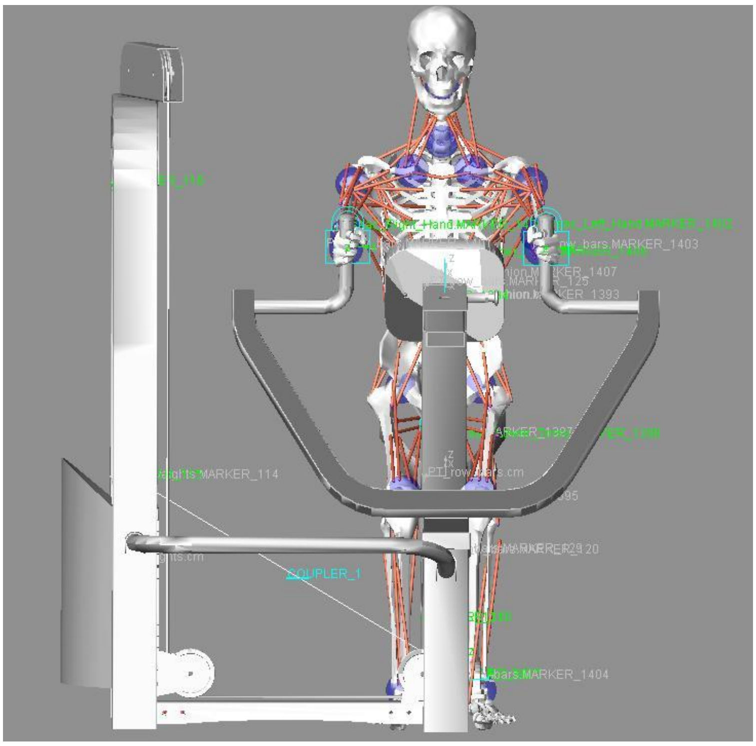
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