THREE DIMENSIONAL MUSCULOSKELETAL MODELLING OF THE SEATED BICEPS CURL RESISTANCE TRAINING EXERCISE

Keywords: *Forward dynamics, inverse dynamics, Lifemodeler[TM] , resistance training equipment, seated biceps curl*

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Abstract

The aim of this study was to evaluate the benefits and limitations of using three dimensional (3D) musculoskeletal modelling (LifeModeler™) in assessing the safety and efficacy of exercising on a seated biceps curl resistance training machine. Three anthropometric cases were studied, representing a 5th percentile female, 50th percentile and 95th percentile male. Results indicated that the LifeModeler™ default model was not adequate to solve the forward dynamics simulations. Therefore adjustments had to be made to the default model to successfully complete the forward dynamics simulations. The software was able to sufficiently highlight the shortcomings of the biceps curl machine's engineered adjustability in relation to the anthropometric dimensions of the studied cases, as the 5th percentile female could not be accommodated suitably on the machine. High lumbar spine anterior/posterior shear forces for all anthropometric cases and maximum muscle tensions for the female and 50th percentile male indicate that the seated biceps curl exercise may pose risks for injuries. To conclude, it appears that 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.

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Introduction

Design of exercise equipment is a complicated task and warrants consideration of a series of biomechanical and ergonomics factors. Furthermore, increased loading is inevitable on certain parts of the body due to the repetitive nature of exercises. Improvement in equipment design could reduce this hazard and offset such a negative effect on the body (Dabnichki, 1998).

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). Capable of simulating musculoskeletal human models interacting with mechanical systems, three dimensional (3D) musculoskeletal modelling may be able to answer many questions concerning the effects of the resistance training equipment on the body. In addition, computer simulation models permit the study of the complex interactions between biomechanical variables (Kenny *et al*., 2005).

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, 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 curl and dumbbell preacher curl are two variations of the standard

dumbbell biceps curl generally applied to optimize the 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). Regardless of variation, the biceps curl exercise can be divided into two phases: (1) lifting phase to flexed position and (2) lowering phase to extended position (Floyd, 2009).

Currently, there does not appear to be any regulation of exercise equipment design or production in either South Africa or internationally. Therefore, evaluation methods are needed to ensure the equipment's efficacy as well as the safety of the end-user. Thus the primary aim of this study was to evaluate the benefits and limitations of using three dimensional (3D) musculoskeletal modelling (LifeModeler™) in assessing the safety and efficacy of exercising on a seated biceps curl resistance training machine.

Methods

Software

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™ (San Clemente, USA) software runs as a plug-in on the MSC ADAMS software. LifeModeler[TM] 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). The default model, as generated through the software, was evaluated. This model consists of 19 segments including a base set of

joints for each body region. Every bone in the human body is included. Furthermore, the default model has a full body set of 118 muscles attached to the bones at anatomical landmarks, which includes most of the major muscle groups in the body. The muscles were created with trainable passive elements (Biomechanics research group, 2006).

Musculoskeletal full body human and seated biceps curl computer aided design (CAD) models

Models for the three anthropometric cases were created. The human models were created using the GeBOD anthropometry database (default LifeModeler™ database) but were based on body mass index (BMI) data obtained from RSA-MIL-STD 127 Vol 1 (2004)(Table I). This data is representative of the South African National Defence Force (SANDF) which is kept current by a yearly sampling plan and can be considered an accurate representation of the broader South Africa population. A process described by Bredenkamp (2007) was followed to characterize the body forms of SANDF males and females found in RSA-MIL-STD-127 Vol 1.   This process identified variances in body form as identified by principal component analysis. Two principle components (PC's) for the SANDF males and females were included in the modelling process and presented the positive boundary case (being tall and thin) and the negative boundary case (being short and heavy).   Positive and negative boundary cases represent the boundary conditions to be accommodated in design (Gordon and Brantley, 1997). A "small" female, an "average" male, and a "large" male were the three anthropometric cases chosen for this study.   They are traditionally known as a 5th percentile female, 50th percentile male and a 95th percentile male based on the BMI. Thus, for the purpose of building these biomechanical models, a correlation between BMI and functional body

strength was assumed. Similar assumptions have previously been made in biomechanics full body model simulations (Rasmussen *et al*., 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 test whether the exercise machine could accommodate the full spectrum 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 (Figure 2). The model in a Parasolid file format was imported into the ADAMS simulation software.

The Adams software was used to create two design variables in order to adjust the external resistance (as selected by the amount of weights when using a selectorised resistance training machine) and to specify the radius of the cam over which the cable of an actual exercise machine would run in order to lift the selected resistance. This was possible since this machine employed a circular cam system. A special contact force (solid to solid) was created between the weights being lifted and the remainder of the weight stack during the simulation. A coupler joint was created linking the revolute joint (driver joint) of the lever arm attached to the handle bars with the translational joint (coupled joint) of the weight stack (Figure 2). The design variable created for the radius of the cam was then referenced as part of the function of the coupler joint in calculating the external resistance, taking into account the resistance selected as well as the radius of the cam on the machine. The design variable created for the mass of the weights was

then adjusted according to the pre-determined resistance for each anthropometric case, explained in the next section.

The external resistance applied in the models was based on data obtained from RSA-MIL_STD 127 Vol 5 (2001). This database consists of a range of human functional strength measurement variables for SANDF males and females. 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 one repetition maximum (1RM) functional strength for each anthropometric case was used, which can be considered a manageable resistance to perform an exercise with appropriate form and technique for two repetitions (Table I).

Simulation

Extreme care was taken with the positioning of the 3D musculoskeletal model onto the CAD model of the seated biceps curl machine to ensure proper technique, posture and positioning according to best exercise principles. The engineered adjustability of the exercise machine was used in order to ensure correct positioning for each of the anthropometric cases. Bushings 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 (Figure 2). 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 they allow for limited translational and rotational motion. Also, the amount of motion can be controlled by changing stiffness and damping characteristics in all three orthogonal directions.

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 that resulted from the prescribed machine movement. The movement replicated two repetitions of the exercise separated by a slight pause between repetitions.

After the inverse dynamics simulation was performed, the rotational motion was removed from the rotational joint of the lever arm of the 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. During the forward dynamics

simulation the model is driven by the internal forces (muscle length changes resulting in joint angles 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^{TM} default model in order to solve the models with plausible kinematics during the forward dynamics simulations. The changes included: 1) increased the physiological cross-sectional area (pCSA) of the three default elbow flexor muscles, 2) manipulated the muscle origins and insertion locations and 3) decreased the joint stiffness in the forwards dynamics simulations. The details of these changes are covered in the Discussion. 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 full spectrum 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 technique throughout the exercise during simulations. Start and end exercise posture evaluation entailed positioning of the axilla on the top of the preacher curl "platfform" 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 should 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.

Peak muscular force production of the prime movers was analysed to determine exercise efficacy of the seated biceps curl. For the purpose of this study, efficacy of the equipment was assessed by evaluating whether the equipment exercised the muscles it was designed for, i.e. 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. Risk to both these structures are very real, especially while lifting excessive masses and/or exercising with poor postures.

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 Gassemieh, 2007). However, criteria for determining whether a particular task or exercise is "safe" based on tissue-level stresses or joint loading are available for only a small number of tissues and loading regimes (e.g. lower back motion segments in compression) (Wagner *et al.*, 2007). Therefore for this study anterior/posterior (A/P) shear forces and joint compression forces were used as safety criteria.

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

Results

Muscle force production (N), muscle length (mm) and joint torque (N-m) are reported for the right side. Theoretically, the results of the left and right side should be similar although 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 II. 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.

Muscle length results for the BBS, BBL and B muscles are presented in Table II. 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 length for all the muscles is greatest for

the 95th percentile male and smallest for the 5th percentile female. The length of the BBL muscle was shortest at approximately 1.6 s and 5.6 s (Figure 3).

Due to the involvement of wrist and elbow joints in the biceps curl exercise, torque for these joints (right side) are presented in Table III. 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. Maximum elbow joint torque production was produced at approximately 1.6 s and 5.6 in the three anthropometric cases (Figure 4 and 5).

Thoracic (T12/L1 intervertebral joint) and lumbar (L5/S1 intervertebral joint) spine compression and A/P shear forces are presented in Tables III and IV 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.

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.

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 as the default model was not capable of generating large enough joint torques to perform the biceps curl. In order to solve this problem the following adjustments were made to the default model: 1) increased the physiological cross-sectional area (pCSA) of the three default elbow flexor muscles, 2) manipulated the muscle origins and insertion locations and 3) decreased the joint stiffness in the forward dynamics simulations. These adjustments were based on running various iterations in order to produce kinematics during the forward dynamics simulation that best matched that of the inverse dynamics simulation (when the model was driven by the external motion agents). Once accurate kinematics was obtained in the forward dynamics run, the kinetic results were evaluated in order to assess if they were reasonable.

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 V). 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, approximately a 9% increase in height with only a 9% increase in volume. To keep proportionality, volume should increase three times more than stature. An additional point to consider is the

significant variance in muscular strength between subjects of similar anthropometry due to differences in conditioning levels.

The muscle origin and insertion points of the BBS and BBL muscles also had to be manipulated in order to increase the moment arm, allowing greater torque to be produced around the elbow joint (Table VI). There is considerable variability in human anatomical structure, including the points at which tendons are attached to bone. An individual whose tendons are inserted on the bone farther 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 on the order of several centimeters, 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 (Table VI). 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. It should be noted that these adjustments resulted in a longer muscle mean length of the BBS than the BBL muscle. While anatomically unusual, these adjustments were required to solve the models during the forward dynamics simulations.

Lastly, the joint stiffness was reduced in the forward dynamics simulation only. Joint stiffness during inverse dynamics (default model) simulations was 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 therefore the joint stiffness was decreased by finite levels through various iterations until acceptable kinematics was achieved. Even after the adjustments the 5th percentile female and the 50th percentile male BBL muscle reached their maximum force production as can be seen in Figure 6. A possible reason for this could be that the biceps curl machine design does not accommodate the anthropometric dimensions of the 5th percentile female and the 50th percentile male as well as that of the 95th 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.

The second conclusion of this study is that the software is able to sufficiently highlight the anthropometric differences with regards to the biceps curl machine's engineered or manufactured adjustability. The adjustability of the biceps curl machine accommodated all the anthropometric cases except for the 5th percentile female (Figure 7). 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 posture of the 5th percentile female was negatively affected. This deficiency in the adjustability of the equipment once again highlights the problem that not all equipment can be 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 perpendicular with the seat of the biceps curl machine. The fact that the misalignment of the seat and preacher curl "platform" was obvious during the modelling process indicates that this method is effective in highlighting such design flaws.

Lastly, with regards to the biomechanical evaluation in terms of exercise efficacy and injury risk, the following could be deduced from the study. The force production was greater for the BBS and BBL in comparison with the B muscle. This result is to be expected as the BB is the prime mover of the biceps curl exercise and is also a larger muscle than the B. It appears as if the relevant muscle groups are being exercised during the seated biceps curl exercise**,** but in order to successfully evaluate the efficacy of the exercise in more detail**,** all the relevant muscles should be present on the model and it would also be useful to compare similar exercises in terms of peak muscle force production of prime movers. Furthermore, the 5th percentile female's force production for all studied muscles was the greatest in comparison with the other anthropometric cases.

A possible explanation for these results is that anatomical differences could result in greater force production required by the small female in order to overcome the external resistance. A shorter lever arm (even though the resistance used for each case was proportional to the anthropometric dimensions) as well as poor accommodation resulting in poor alignment could result in unexpectedly high muscle force produced by the female model.

The joint torque values obtained for the wrist and elbow appear to be plausible, as they fall well below peak values obtained by means of isokinetic testing. For example wrist flexion/extension values of 13.8 N-m and 12.7 N-m respectively at 60 degrees per second in non-disabled subjects (Van Swearigen, 1983) and elbow flexion/extension values of 36 N-m for both elbow flexion and extension at 60 degrees per second in female college basketball players (Berg *et al.,* 1985).

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 BB 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 BB is midway between fully elongated and fully shortened (Durall, 2004). In this study, the maximum joint elbow torques were reached at joint angles between approximately 55 degrees (5th percentile female) and 85 degrees (95th percentile male)(Figure 4). 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 3). Although these results may appear contradictory, it must be noted that the shortest muscle length reached during the exercise period was indeed very close to the natural resting length of the BB muscles, to be distinguished from the shortest anatomical length of the muscle during the full range of motion of the joint.

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). For 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 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 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. A model with 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, and no further recommendations other than " time of exposure needs to be minimised" and "a preferred system requires optimal ergonomic position with reduced back bending posture" are made. All three anthropometric cases were below the recommended failure limit of 3.4 kN but 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.

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 so if exercisers use a resistance closer to their maximum the loading values may exceed the acceptable limits. The modelling also does not consider varying training status or muscular strength and endurance of individuals which could affect the individual's risk for injury. Core musculature, which plays an important role in reducing the joint loading on the spine, 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 was strained above its maximum 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 highlight some 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. High recorded lumbar spine A/P shear forces for the three anthropometric cases indicate that the seated biceps curl exercise may pose a risk for low back injuries. Extended periods of maximal muscle tension in both the 5th percentile female and 50th percentile male indicate that the seated biceps curl exercise may pose a risk for elbow flexor injuries. However, the unfavourable positioning of the small female did not appear to put her at increased risk for injury in comparison to the other two anthropometric cases. The Lifemodeler^{TM} 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. 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.

3D musculoskeletal modelling can certainly be used to evaluate resistance training equipment design, but the limitations discussed in 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, making the process of 3D musculoskeletal modelling more user-friendly and effective in evaluating various types 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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Table I. Anthropometric and strength data of population groups studied (RSA-MIL-STD-127, Vol 1, 2004 and Vol 5, 2001).

User group	Body mass (kg)	Stature (mm)	User population group exercise resistance (50% 1RM) kg
5th percentile female	49.5	1500	12
50th percentile male	65.0	1720	22
95th percentile male	85.0	1840	35

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

Musculoskeletal model	Muscle	Max (N)	Mean (mm)	Min (mm)	Max (mm)
5th percentile female	Biceps brachii short head	268.9	239.2	228.9	253.9
	Biceps brachii long head	329.5	217.0	206.8	235.9
	Brachialis	215.1	105.6	103.5	112.0
50th percentile male	Biceps brachii short head	221.5	300.6	281.1	315.8
	Biceps brachii long head	267.7	274.8	253.8	294.3
	Brachialis	172.6	131.9	122.6	142.5
95th percentile male	Biceps brachii short head	172.3	330.5	307.5	349.5
	Biceps brachii long head	215.9	303.9	280.7	325.3
	Brachialis	172.3	143.3	129.8	156.5

Table III. Right wrist and elbow joint torque (N-m) (sagittal plane) and thoracic and lumber spine joint compression force (N) results for the 3 anthropometric cases.

Musculoskeletal model	Joint	Mean (N-m)	Min (N-m)	Max (N-m)	Max (N)
5th percentile female	Wrist	0.8	2.8	3.9	
	Elbow	3.7	-28.3	11.6	
	Thoracic spine				1774.1
	Lumbar spine				2337.2
50th percentile male	Wrist	1.9	-4.2	3.7	
	Elbow	8.1	5.4	17.7	
	Thoracic spine				2123.3
	Lumbar spine				2920.5
95th percentile male	Wrist	3.4	0.2	6.2	
	Elbow	12.6	1.8	25.3	
	Thoracic spine				2133.2
	Lumbar spine				2821.7

Table IV. Thoracic and lumbar spine joint anterior/posterior (A/P) shear forces (N) for the 3 anthropometric cases.

Musculoskeletal model	Spinal joint	Max (N)
5th percentile female	Thoracic spine	736.8
	Lumbar spine	906.0
50th percentile male	Thoracic spine	901.0
	Lumbar spine	1109.0
95th percentile male	Thoracic spine	974.3
	Lumbar spine	1180.7

Table V. 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
5th percentile female	147.2	180.5	116.0
50th percentile male	178.7	218.8	139.7
95th percentile male	177.6	217.4	138.9

Table VI. Right Biceps brachii muscles lengths (mm) before and after the origin and insertion adaptations for the 3 anthropometric cases.

Musculoskeletal model	Muscle	Default model	After adaptation	Change
5th percentile female	Biceps brachii short head	209.4	249.7	40.3
	Biceps brachii long head	224.4	229.6	5.2
50th percentile male	Biceps brachii short head	282.7	315.1	32.4
	Biceps brachii long head	281.9	292.6	10.7
95th percentile male	Biceps brachii short head	314.4	338.5	24.1
	Biceps brachii long head	315.7	339.8	24.1

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

Figure 2. A top view (top left), side view from the right (top right), side view from the left (bottom right) and front view (bottom left) of the exercise machine. Descriptions for the labelled parts are as follows: A = preacher curl "platform" (non-adjustable), B = adjustable seat, C = handle bars, D = circular cam, E = translational joint created in order to simulate the lifting of the weights, F = coupler joint created in order to link the translational movement of the weights and the rotational movement of the lever arm of the apparatus, G = centre of rotation for the lever arm, which should also serve as a guideline for placement of the elbow joint centre.

Figure 3. Long head of the Biceps brachii muscle length (mm) for the 3 anthropometric cases.

Figure 4. Sagittal elbow joint angle (°) for the 3 anthropometric cases.

Figure 5. Elbow joint torque (Nmm) for the 3 anthropometric cases.

Figure 6. Biceps brachii long head force production (N) for the 3 anthropometric cases.

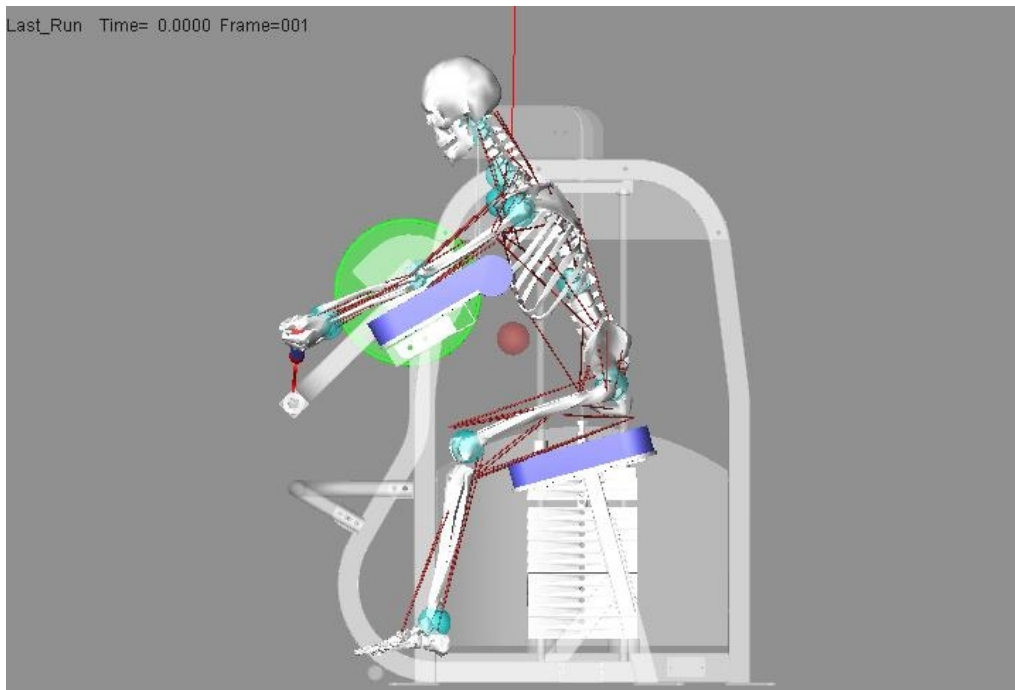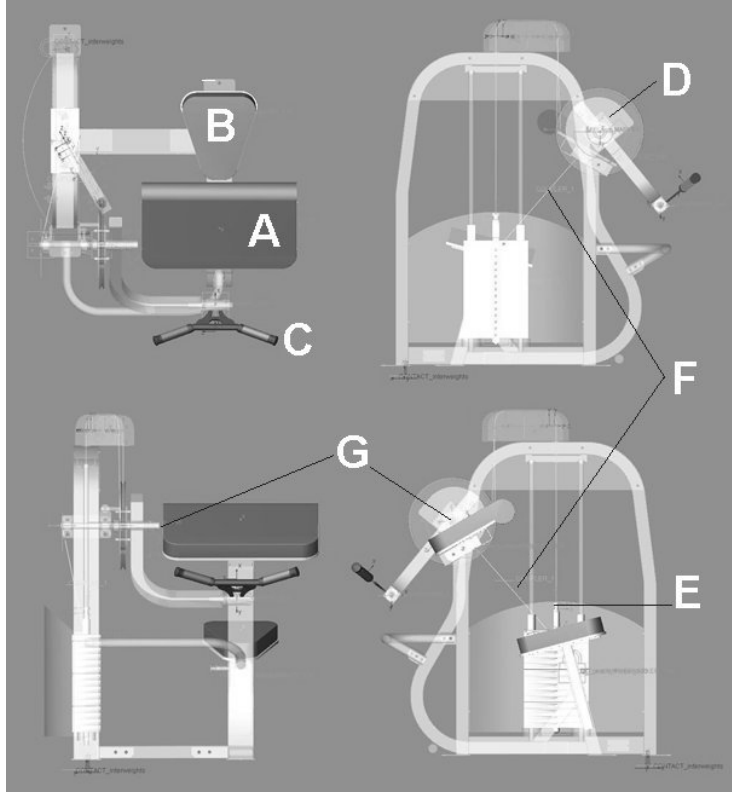Figure 7. 5th percentile female's positioning on the seated biceps curl machine
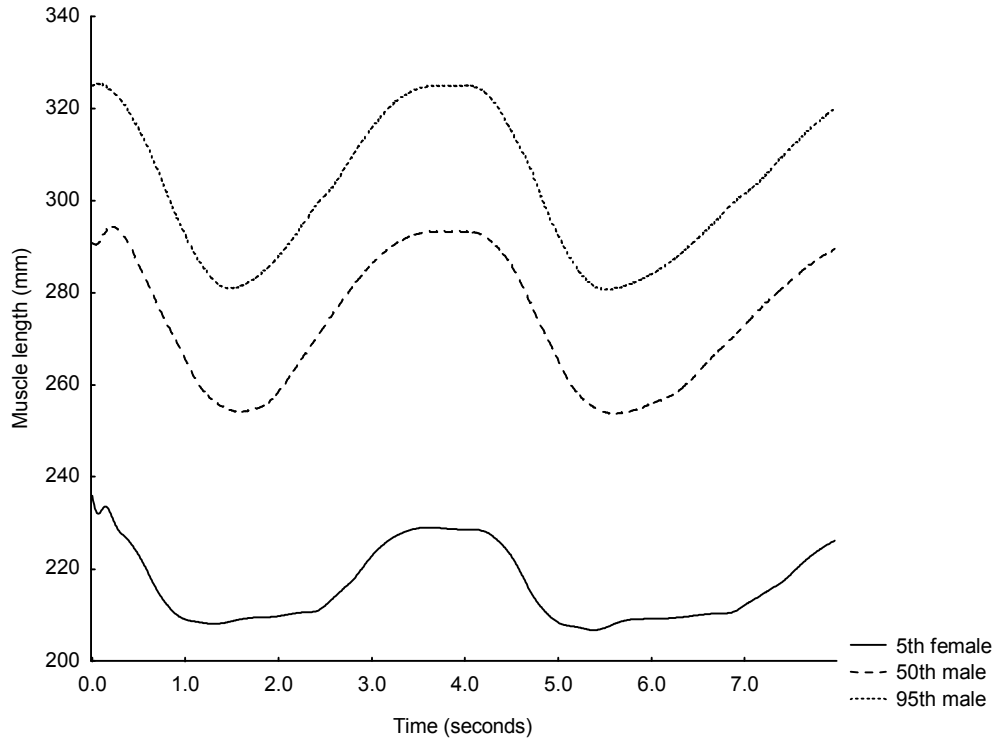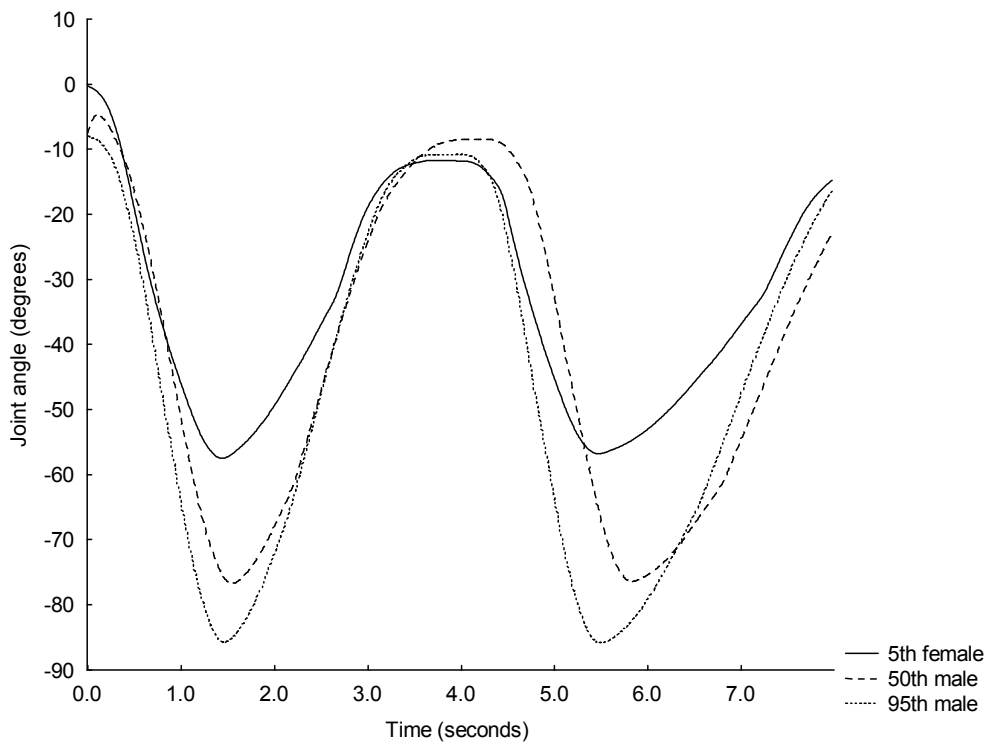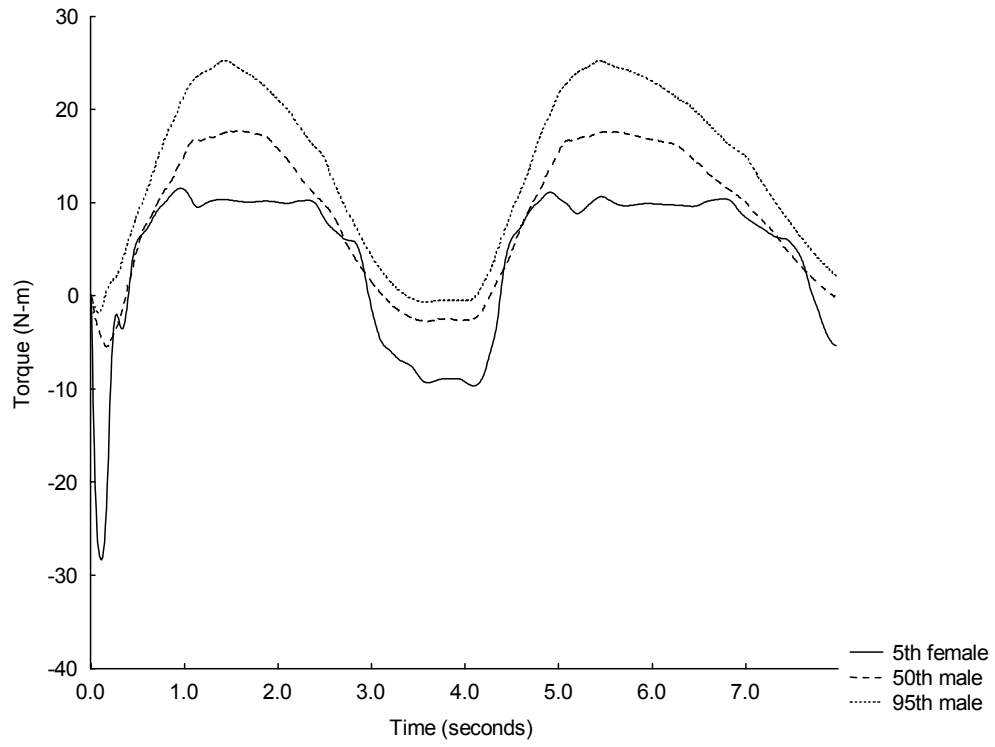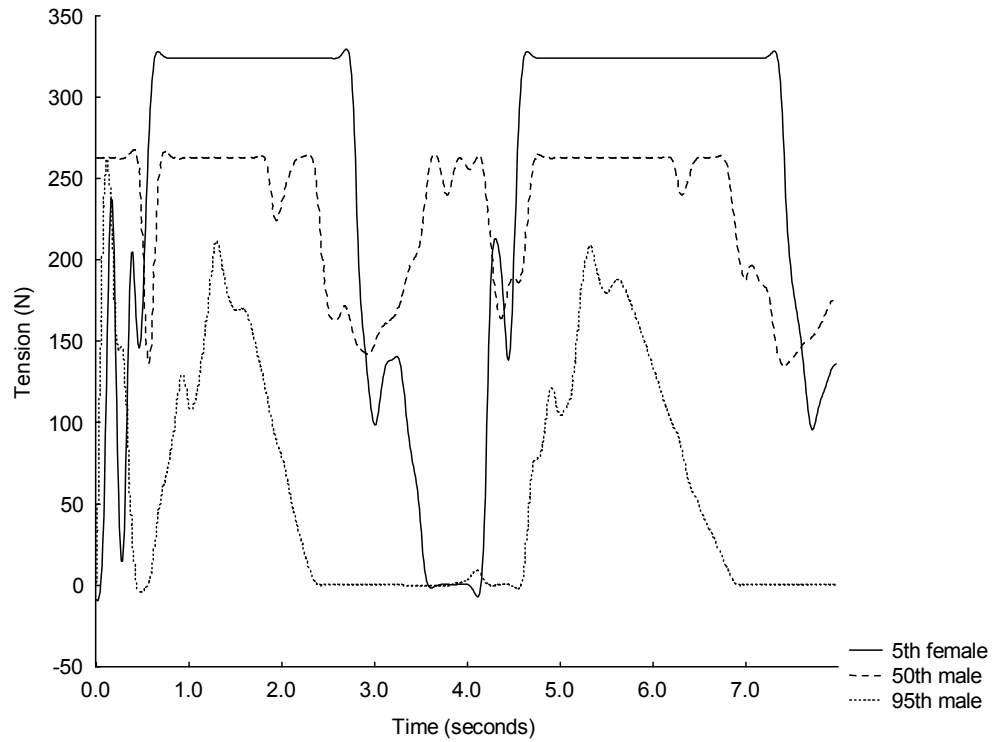

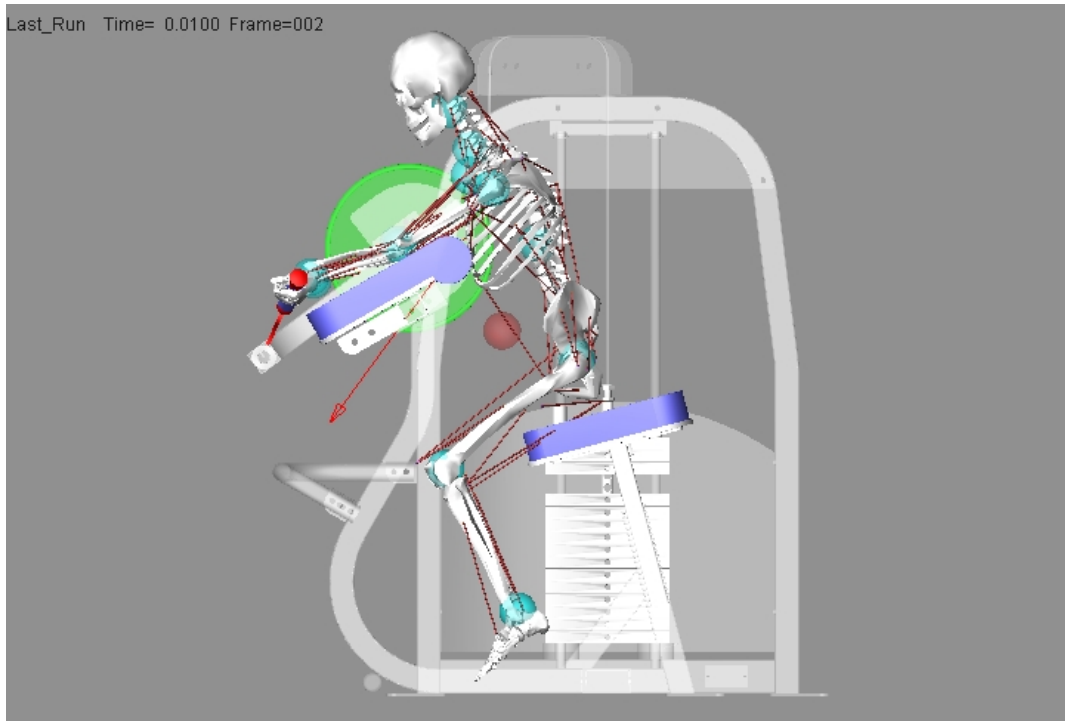Last_Run Time= 0.0000 Frame=001

Figure 1.

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Figure 6.

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Figure 7.