CHANGES IN LUMBAR JOINT MOMENTS USING A FEMALE SPECIFIC TORSO MODEL DURING RUNNING

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The purpose of this study was to quantify the peak lumbar joint flexor / extensor moments following changes in torso and breast mass during running using an innovative computer musculoskeletal model. Kinematic and kinetic data were collected for a female participant running at 2.6 m/s. An MRI scan of the breasts was used to calculate breast mass and centre of mass location relative to the torso. An OpenSim whole body model was customised with two point-mass segments added to the torso to represent the breasts. Key findings have shown that changes in breast mass can cause peak lumbar flexor / extensors moments to be over or underestimated by up to ~18%. These results suggest that including the mass of the breasts in female specific models, during dynamic activities such as running, is an important aspect that must be considered for future work.

KEYWORDS: breast, OpenSim, mass, musculoskeletal

INTRODUCTION: Musculoskeletal models have been used to investigate changes in muscular demand associated with a variety of human motion (Baskar & Nadaradjane, 2016; Kar & Quesada, 2013; Scarton et al., 2017). One aspect of the modelling process involves the requirement for a participant-specific model, based on an individual's anthropometric measurements, strength parameters and motion used to execute a task (Wilson et al., 2006). Anthropometric measurements (such as segment lengths, circumferences) are utilised to estimate the inertia characteristics of geometric shapes used to represent body segments (Mills et al., 2009; Pain & Challis, 2006; Yeadon et al., 1990). Many generic models using commercial (e.g. Visual 3D) or open source (e.g. OpenSim) software facilitate some customisation of the torso segment but remain limited to a scaled male shaped torso segment and therefore fail to account for the effects of breast mass (located on the anterior torso) on the calculated loads.

Breast volumes can range from 150 to 2000 ml (McGhee & Steele, 2011) and assuming a breast mass density of 945 kg/m^3(Sanchez et al. 2016), breast mass can range from 0.14kg to 1.89kg. Changes in breast mass, following breast augmentation or reduction surgery, mastectomy or weight gain, has been shown to alter female posture (Nicoletti et al., 2015), and change lumbar erector spinae muscular activity (Schinkel-Ivy et al., 2016). These findings confirm that breast mass alters musculoskeletal demand and illustrates that female whole-body musculoskeletal models require improved torso segment design to enable important research within the female population in a variety of applications. Neglecting this important aspect may cause a misrepresentation of the estimates of lumbar joint moments in females. The aim of this study was to vary the torso and breast mass within the OpenSim model to determine the effect upon lumbar flexor / extensor moments during running. The purpose was to improve the prediction of the muscular demand of models, using female participants and aid the subsequent understanding of potential injury mechanisms in a variety of sports and activities.

METHODS: Following institutional ethical approval, one female participant (height: 1.64 m; mass: 65 kg; bra size: 34D) was recruited for this study and provided written informed consent. The participant was asked to conduct a gentle warm up then 53 reflective markers were attached to the body and breasts at key landmarks (Figure 1b). A series of anthropometric measurements (segment lengths, circumferences and landmark separation distances) were manually recorded. The participant was asked to stand for a 5s static trial, then asked to run over ground, bare breasted at a self-selected speed (2.6 m/s) whilst synchronised kinematic and kinetic data were collected. Three force platforms (Kistler, 9281CA; 1000Hz) and a 16 camera motion capture system (Qualisys, Sweden; 300Hz) collected synchronised kinematic and kinetic data for one gait cycle. Magnetic Resonance Imaging (MRI) scans of the torso and breasts (Figure 1a) were acquired with a breast coil on a Philips Ingenia 1.5 T (Philips Healthcare, Best, NL) using the dual-echo mDixon sequence (software version 5.1.7.2) (Eggers et al., 2011) on the same day. An acquisition matrix of 300 x 300 was used with in-plane resolution of 1.5 x 1.5 mm^2 and a slice thickness of 3 mm. Breast mass and centre of mass (CoM) were calculated by segmenting the breast from the torso and identifying the tissue as either fatty ($900 kg \cdot m^{-3}$) or glandular ($1057 kg \cdot m^{-3}$). For each slice, the area of glandular and fatty tissue was measured and multiplied by the slice thickness; this volume was then used to calculate breast mass. The 3D reconstruction of the breast was used to calculate centre of mass location relative to the sternal notch.

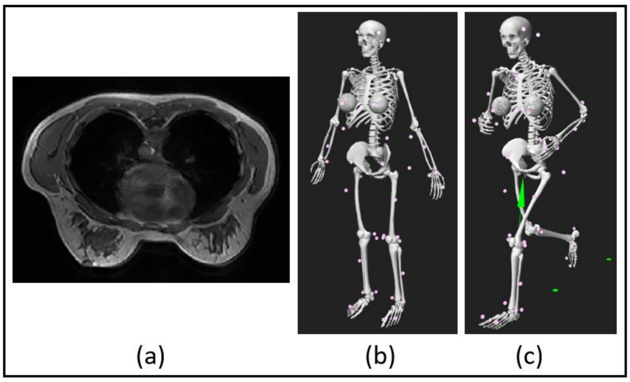

Figure 1. Customisation of Full-Body Lumbar Spine (FBLS) model (Raabe & Chaudhari, 2016) in OpenSim. (a) MRI slice of torso and breast, (b) static pose, (c) dynamic running trial.

The FBLS model (Raabe & Chaudhari, 2016) comprised of 21 segments and 30 degrees-of-freedom, the five lumbar vertebrae were modelled as individual bodies, and coupled constraints were implemented to describe the net motion of the spine. This base model was customised to include point-mass segments (breasts) attached to the torso at the location calculated from the MRI scan and static trial kinematic data (Figure 1b). This approach was used to provide a first approximation of the effect of altered torso model geometry on resulting lumbar joint moments. The customised OpenSim model was scaled to the participant using both the static trial and anthropometric measurements with a root mean square error of <1.2cm and maximum error of <2cm between the experimental and model marker locations. The Inverse Kinematics tool (weighted least squares between experimental and model marker locations) was used to calculate the joint time histories. The joint time histories were combined with the ground reaction forces (GRF) to run inverse dynamics analyses (Figure 1c). Firstly, the proportions of breast to torso mass were changed from 0 to 2 kg (0kg being no breast mass and 100% torso mass; to ~3% breast mass and ~97% torso mass), whilst the combined breast and torso mass remained constant. Secondly, torso mass remained constant and breast mass was increased (simulating breast augmentation) from 0kg to 0.8kg (Nicoletti et al., 2015). Inverse dynamics were run using the model for five changes in torso to breast mass proportions and a further four run for each of the additional breast mass (augmented) conditions. Peak lumbar spine flexor / extensor moments were output and compared to the changes in torso / breast mass conditions.

RESULTS: The results suggest that increasing the breast mass by 2 kg (and decreasing torso mass proportionally) causes a decrease in peak lumbar flexor moments (~18%) and an increase in peak lumbar extensor moment (~6%) when compared to a ‘male’ torso with no breast mass. Adding breast mass (simulated breast augmentation) also results in a decrease in peak lumbar flexor moments (~8%) and an increase in peak lumbar extensor moments (~7%) during running (Table 1).

Table 1. Peak flexor or extensor moments following torso / breast mass changes during running.

		Peak flexor and extensor moment (Nm/kg)							
Torso	Breast	Lumbar 5 / 4		Lumbar 4 / 3		Lumbar 3 / 2		Lumbar 2 / 1	
Mass (kg)	Mass (kg)	Flex	Ext	Flex	Ext	Flex	Ext	Flex	Ext
16.75	0.00$^{\Delta}$	0.27	0.91	0.28	0.96	0.28	1.00	0.23	0.83
15.75	0.50$^{\Delta}$	0.25	0.90	0.26	0.95	0.26	0.98	0.22	0.82
14.75	1.00$^{\Delta}$	0.24	0.92	0.25	0.97	0.25	1.01	0.21	0.84
13.75	1.50$^{\Delta}$	0.23	0.93	0.24	0.99	0.25	1.03	0.21	0.86
12.75	2.00$^{\Delta}$	0.22	0.95	0.23	1.01	0.24	1.05	0.21	0.88
15.62	1.13+0.00*	0.25	0.90	0.26	0.95	0.26	0.99	0.22	0.82
15.62	1.13+0.20*	0.24	0.91	0.25	0.96	0.26	1.00	0.22	0.84
15.62	1.13+0.40*	0.24	0.93	0.25	0.98	0.25	1.02	0.21	0.85
15.62	1.13+0.60*	0.24	0.94	0.25	0.99	0.25	1.04	0.21	0.86
15.62	1.13+0.80*	0.23	0.96	0.24	1.01	0.25	1.05	0.21	0.88

*participant left and right breast mass + additional breast mass per breast (simulated augmentation)
$^{\Delta}$ per breast

DISCUSSION: The aim of this study was to vary the torso and breast mass within a female specific musculoskeletal model to determine the effect on lumbar flexor / extensor moments. The results provide a first approximation of the effect of altered torso / breast geometry on joint moments during running. Key findings show that increasing breast mass by 2 kg (and decreasing torso mass proportionally) causes a decrease in peak lumbar flexor moments (~18%) and an increase in peak lumbar extensor moment (~6%). Importantly, the results also show that if a researcher uses female participants and scales a 'male' torso model, this can under or overestimate the lumbar joint moments during running. The example data (Table 1) demonstrated that for a female participant (bra size 34D), lumbar flexor moments are overestimated by ~7% and lumbar extensor moments are underestimated by ~1% during running at 2.6 m/s. These findings illustrate that torso CoM position is influenced by the redistribution of torso mass (inclusion of breast mass) and for a given GRF vector, lumbar joint moments will change. This is an important consideration when investigating changes in muscular demand between genders in areas such as the effects of load carriage on joint work during running (Liew et al., 2016).

Breast augmentation surgery has also been shown to not alter posture (Mazzocchi et al., 2012) due to increases in self-esteem, however increased muscular activity (Schinkel-Ivy et al., 2016) is required to maintain posture. The results (Table 1) also support these findings and show that an 800cc implant can decrease the peak lumbar flexor moment by ~8% and increase the peak lumbar extensor moments by ~7% during running. These findings confirm that breast mass does impact upon musculoskeletal demand and could have implications for pre and post-surgical advice.

Whilst the musculoskeletal model has not been directly evaluated using muscle activity data (Raabe & Chaudhari, 2016), the magnitude of the lumbar flexor / extensor moments during running are comparable to those published (Raabe & Chaudhari, 2016). This provides increased confidence in the musculoskeletal model results and illustrates that female whole body musculoskeletal models require improved torso segment design to enable important research within the female population in a variety of applications. Furthermore, neglecting this important aspect may cause a misrepresentation of the estimates of muscular demand in females by up to ~18%.

It is important to highlight that the 'female' musculoskeletal model results illustrate a first approximation of how changes in torso geometry can effect torso joint moments during running. It is noted the breast centre of mass location was not altered following increases in breast mass and that up to 15 cm of breast motion occurs during running (Scurr et al., 2011), therefore the dynamics of a moving breast may further alter the musculoskeletal loading. It is reasonable to assume that the simplification of the female breast in this model and no representation of the dynamics of the female breast anatomy may result in greater differences between 'male' and 'female' torso musculoskeletal models. Therefore, it is recommend that future musculoskeletal models, using female participants, consider the possible effect that breast mass and motion may have on the subsequent calculation of musculoskeletal loading.

CONCLUSION: Key findings have shown that changes in torso / breast mass are sufficient to alter the positon of the torso CoM and can cause peak lumbar flexor / extensors moments to be over or underestimated by up to ~18%. These results suggest that including the mass of the breasts in female specific models, during dynamic activities such as running, is an important aspect that must be considered for future work.

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