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

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This study aimed to investigate the effect of breast mass and motion on lumbar flexor / extensor moments during running. Kinematic and kinetic data were collected for a female participant running at 2.6 m/s. An MRI scan was used to calculate breast mass and centre of mass location. An OpenSim model was customised with two point-mass segments added to the torso to represent the breasts. Three model variations were constructed (combined breast and torso mass; separate breast and torso mass without breast motion; separate breast and torso mass with breast motion). Findings show that neglecting breast motion causes peak lumbar extensor moments to be underestimated by ~3.4%, compared with a combined breast and torso mass model. These results highlight the importance of including breast motion in female specific MSK models, during activities such as running.

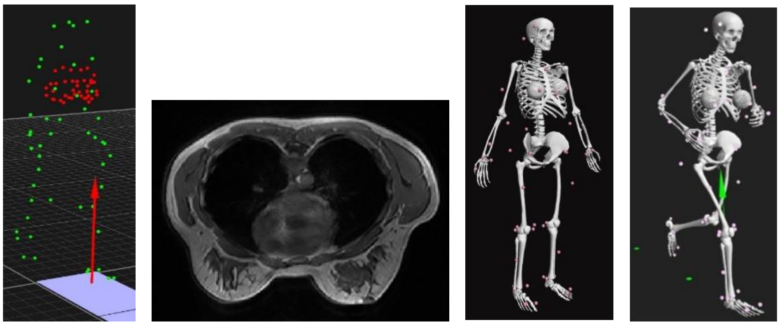
KEYWORDS: breast, OpenSim, mass, musculoskeletal.

INTRODUCTION: Musculoskeletal (MSK) modelling and simulation methods are often applied using generic models, with subject-specific models (Modenese et al 2016; Suwarganda et al., 2019) rarely used due to the associated experimental (3D imaging data) and modelling (segmentation and registration) challenges related to their creation. In both cases, these models are often created and scaled without including anatomical differences between male and female athletes, that cannot just be accounted for with scaling alone. This approach has been used to investigate how changes in body weight loads effect spinal loading within the female torso during a range of activities (Bruno et al., 2017). However, this study raised concerns over the suitability of the representation of the actual loading experienced by female participants as the most obvious difference in male and female torsos is the soft tissue in the breasts of females. Whilst some attempts have been made to produce female components of a model such as the spine or neck musculature (Zheng, 2011; Roos et al., 2020), soft tissue inclusion of any kind in MSK models is still limited as not only is there the issue of incorporating it in the mechanical structure of the model, there is a larger issue that the *in vivo* soft tissue mechanics are not well known in the first place.

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, and move from 2 to 15 cm during running, which has been linked with both pain and discomfort (Scurr et al., 2011). Additional experimental research has found muscular activity increases during running (Milligan et al., 2014) with reduced breast support, confirming that breast mass and movement, impacts upon torso musculoskeletal loading. Hence, neglecting this important aspect may cause a misrepresentation of the estimates of lumbar joint moments in females. The aim of this study was to investigate the effect of breast mass and breast motion upon lumbar flexor / extensor moments during running.

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 conducted a gentle warm up, then 53 reflective markers were attached to the body and 42 on the breasts at key landmarks (Figure 1a) using hypoallergenic double-sided tape. 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. On the same day, Magnetic Resonance Imaging (MRI) scans of the torso and breasts (Figure 1b) were acquired 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). 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 ($900kg\cdot m^{-3}$) or glandular ($1057kg\cdot m^{-3}$). 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 markers was used to calculate centre of mass location relative to the torso (Jones et al., 2020).



(a) (b) (c) (d)

Figure 1. Customisation of Full-Body Lumbar Spine (FBLS) model (Raabe & Chaudhari, 2016) in OpenSim. (a) motion capture markers on the body (green) and breasts (red) (b) MRI slice of torso and breasts, (c) static pose, (d) dynamic running trial.

The FBLS model (Raabe & Chaudhari, 2016) comprised of 21 segments and 30 degrees-of-freedom (DOF), the five lumbar vertebrae were modelled as individual bodies, and coupled constraints were implemented to describe the net spine motion. 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 1c). The customised OpenSim model was scaled using both the static trial and anthropometric measurements with a root mean square error of <1.2 cm and maximum error of <2.0 cm between experimental and model marker locations. The Inverse Kinematics tool (weighted least squares between experimental and model marker locations) was used to calculate joint time histories.

Three variations on the model's construction were developed: Firstly, the torso and breasts were constructed by adding each breast mass (left breast = 0.59 kg and right breast 0.54 kg) to the mass of the torso (15.62 kg) (model 1). Secondly, each breast was positioned separately on the torso (Figure 1c) and the appropriate breast mass applied (each breast was rigidly attached to the torso, hence no breast motion) (model 2). Thirdly, the torso and breasts were constructed the same as model 2 but each breast could move independently via a 3 DOF joint (model 3). The whole body joint time histories were combined with the ground reaction forces (GRF) to run inverse dynamics analyses (Figure 1d) using the 3 model constructions and peak lumbar spine flexor / extensor moments were output and compared.

RESULTS: The results show that lumbar extensor moments increased by up to ~2.5% when separating the torso and breast mass (model 2) from the combined torso and breast mass (model 1). A further increase in lumbar extensor moments of ~1.4% occurred when the dynamic motion of the breast mass was also included within the model construction (model 3) (Table 1). There was minimal change in flexor moments between model constructions. The inferior motion of the breasts near mid-stance coincides with a peak in Superior-Inferior (S-I) breast force and peak lumbar extensor moments (Figure 2).

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

Model Construction	Peak flexor and extensor moment (Nm/kg)							
	Lumbar 5 / 4		Lumbar 4 / 3		Lumbar 3 / 2		Lumbar 2 / 1	
	Flex	Ext	Flex	Ext	Flex	Ext	Flex	Ext
Model 1 torso & breast mass combined	0.26	0.88	0.27	0.93	0.27	0.96	0.22	0.80
Model 2 separate torso & breast mass–static	0.25	0.90	0.26	0.95	0.26	0.99	0.22	0.82
Model 3 separate torso & breast mass–dynamic	0.26	0.91	0.27	0.96	0.27	1.00	0.23	0.84

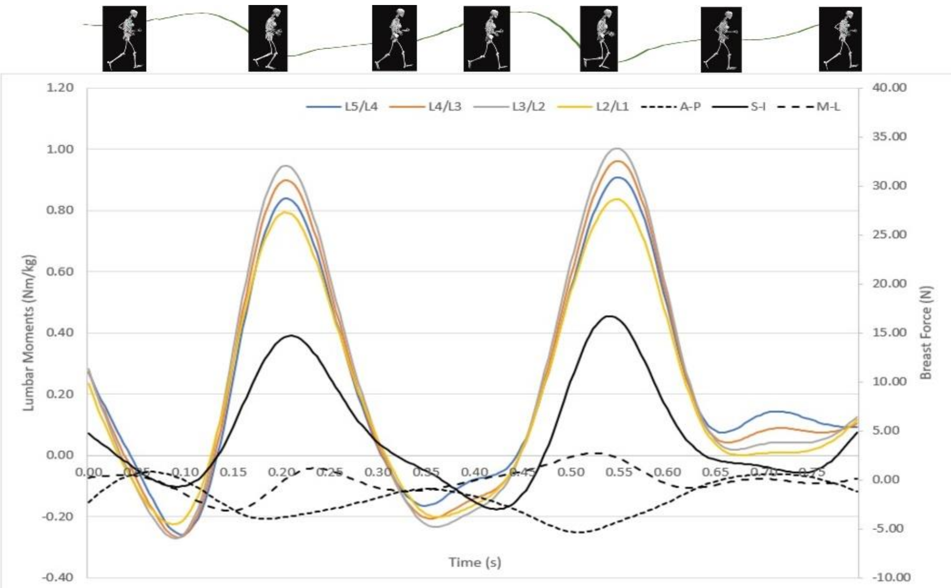


Figure 2. Model 3 lumbar moments (+ = extensor, - = flexor) and right breast force (Anterior-Posterior (A-P); Superior-Inferior (S-I); Medio-lateral (M-L)) time histories during a gait cycle. (NB: green line within gait cycle model pictures shows the relative superior-inferior breast motion over the gait cycle).

DISCUSSION: The aim of this study was to investigate the effect of breast mass and motion upon lumbar flexor / extensor moments during running. Key findings show that if the mass of the breasts are modelled separate to the torso, but constrained with no motion, lumbar extensor moments increase by ~2.5% when compared to the breast mass combined with the torso mass. Importantly a further ~1.4% increase in lumbar extensor moments occurred when the dynamic motion of the breasts were incorporated into the model.

The example data (Table 1) suggests that for a female participant (bra size 34D) 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. Additionally, the inclusion of dynamic breast motion further increases the lumbar extensor moments required to maintain the joint kinematics associated with this jogging trial. 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). Throughout the gait cycle, breast motion caused changes in breast force and hence the lumbar moments required to maintain posture whilst running. Figure 2 shows that during initial foot contact to mid stance the breast moves in an inferior direction, relative to the torso, whilst also inducing increases in S-I breast force. This coincides with a peak in the extensor moments required to maintain an upright position during running. The participant in this study had a combined breast mass of 1.13 kg. However, women with greater breast mass, such as 1.9 kg per breast (Brown et al., 2012) or who have increased mass due to breast augmentation surgery will likely experience greater musculoskeletal demand and require greater lumbar extensor moments to maintain posture.

Additional empirical evidence suggests increased muscular activity (Schinkel-Ivy et al., 2016) is required to maintain posture, with increases in breast mass and the current findings also builds upon previous work by Mills & Jones (2020) showing how increases in breast mass effect lumbar joint moments. The magnitude of the lumbar flexor / extensor moments during running are comparable to those published (Raabe & Chaudhari, 2016). This similarity 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. Although a limitation that the model is based upon one participant, with a bra size of 34D, and the magnitude of lumbar loading may vary depending upon breast mass and motion, the underpinning mechanical response does show the need to consider the breast in female models. Therefore, it is recommended 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 incorporating the breast mass and their motion during running influences peak lumbar extensors moments, for the breast size modelled in this study. These results suggest that including the mass and motion 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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