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Modelling the female torso and breast during physical activity: Implications on spinal loading

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ABSTRACT

Methods of modelling the female torso during physical activity often neglect the position and movement of the breast. This novel investigation compares three female torso modelling approaches that differ in complexity (integrated breast, fixed breast, dynamic breast) to determine the effect on spinal joint moments during running and jumping. The commonly used integrated breast model distributed breast mass within the torso, the fixed breast model attached the mass of the breasts to fixed positions on the anterior of the torso, and a new dynamic breast model enabled relative motion between the breasts and anterior torso. Key findings demonstrated minimal differences in lumbar spine moments (<0.05 Nm/kg; 4%) between integrated breast and fixed breast models but greater differences, up to 0.86 Nm/kg (68%) during running and 0.89 Nm/kg (82%) during jumping, when breast motion was included. Thoracic spine moments revealed similar patterns with minimal differences (<0.05 Nm/kg; 11%) between integrated breast and fixed breast models and greater differences, up to 0.48 Nm/kg (92%) during running and 0.63 Nm/kg (66%) during jumping, with the dynamic breast model. Future female musculoskeletal models should consider including breast mass and motion to avoid mis-representing spinal loading in females during running and jumping.

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KEYWORDS
Biomechanics; moments; musculoskeletal; soft tissue; spine

1. Introduction

Musculoskeletal modelling has been extensively used for non-invasive estimation of internal loads acting on the human body during various forms of physical activity (Akhavanfar et al., 2022; Cazzola et al., 2017; Mills et al., 2009). Recent practice is to scale the segment geometry of a gender specific rigid body skeleton model to experimental participants (Banks et al., 2023). Models of the female torso include detailed geometry of the underlying skeleton (Bruno et al., 2017) but do not include any separate soft tissue segments, such as the female breasts, and instead any breast mass is included within the scaling of the torso segment. Breast mass and motion may be an important consideration given that previous research has demonstrated that the inclusion of soft tissue movement within musculoskeletal models reduces predictions of internal loading and improves alignment of external forces with experimental data (Gittoes et al., 2006; Pain & Challis, 2006).

The soft tissue of the female breasts has a combined mass of 0.92 kg for a (UK average) bra size of 34D (Turner & Dujon, 2005). Whilst simplified fixed breast geometry and mass have been included in a model investigating the effect of pregnancy on joint moments (Haddox et al., 2020), the breasts actually move in three-dimensions relative to the torso by up to 15.2 cm during running (Scurr et al., 2011) and 18.7 cm during jumping (Bridgman et al., 2010) when unsupported, thus inducing a force on the torso. Whilst it is possible to adapt torso inertial properties to account for the anterior position of the breast mass (Haddox et al., 2020), breast motion relative to the torso and the associated forces and moments that are applied cannot be represented using this approach. Therefore, without a female specific torso model with sufficient details to modify the mass and motion of the breasts relative to the torso, current research may be mis-representing the internal loading for females during common physical activities such as running and jumping.

Back pain is widespread in the general population, especially among individuals engaged in physical activity (Trompeter et al., 2017). Many women report back pain that is attributed to a larger breast size (Coltman et al., 2018), and larger breasts combined with breast motion during physical activity have the potential to induce spinal joint moments that will require additional muscular effort to maintain torso kinematics. Earlier research has shown that altering the mass or the anterior–posterior position of the load carried, relative to the torso, changes the forces acting within the spine during physical activity (Rose et al., 2013). Furthermore, a change of 0.05 Nm/kg in lumbar spine moments has also been reported as the difference between participants with and without back pain (Hasegawa et al., 2018). Although several studies have investigated spinal loading during physical activity (Alvim et al., 2019; Banks et al., 2023; Bruno et al., 2017), the soft tissue mass and motion of the female breasts relative to the torso have been neglected in musculoskeletal models to date.

The general challenge in modelling human soft tissue using a multi-body system lies in accurately capturing the complex behavioural properties of soft tissues while considering their

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interactions with other body segments as well as external forces. Open source software, such as OpenSim, can be used for the development of musculoskeletal models and the simulation of human movement (Delp et al., 2007). A full-body musculoskeletal model with multiple bodies, rotational degrees of freedom within the lumbar and thoracic spine and female skeletal geometry (Bruno et al., 2017), would allow for model customisation and development of dynamic breast segments. Therefore, the aim of this study was to investigate the effect of three female torso and breast modelling approaches upon spinal loading during physical activity.

2. Materials and methods

2.1. Data collection

Following an institutional ethical approval, one healthy female (74.6 kg, 1.79 m, 26 years, UK bra size 34DD) provided written informed consent to participate in this study. Fifty-seven reflective markers were attached to key anatomical locations (Figure 1(a)) using hypoallergenic adhesive tape, with a handheld ultrasound machine (Sonosite Edge, USA) used to guide marker placement on the spine. One additional marker was placed on the nipple of each breast over the bra. The marker set was required in the measurement-based scaling process to ensure accurate representation of the participant's torso and spine motion and to ensure appropriate force application to the model at the feet. A 19-camera motion capture system (Qualisys, Sweden), sampling at 250 Hz, was synchronised with three force platforms (9281E, Kistler, Switzerland) sampling at 1000 Hz. For gross breast motion, represented by a single marker, sampling rates between 200–250 Hz are common place (Mills et al., 2016), compared with higher sampling rates that are used to investigate localised intersegmental deformation (Mills et al., 2011). The force platforms were embedded in series within the lab floor in the direction of running and central to the capture volume (Figure 1(b)). The left and right breast boundaries were identified using the folding line method (Lee et al., 2004) and the most superior, inferior, medial and lateral positions on the boundary marked using a surgical marker pen. The participant wore a sports bra (Triumph, Triaction) that represented a typical bra a female might wear when doing physical activity and fitted using the best-fit criteria to minimise any relative motion between breast and bra (White & Scurr, 2012). The participant was asked to stand statically for 15 s, whilst a hand-held 3D surface scanner (Artec Eva, Luxembourg) was used to record the torso and breast geometry for subsequent calculation of each breast volume, mass and centre of mass (COM) location. Following a gentle warm up, the participant was asked to stand statically for 5 s, whilst data were collected and then asked to perform a running trial at a self-selected speed (3.15 m/s), followed by a maximal effort standing vertical jump (with hands on hips, jump height 27 cm). Kinematic and kinetic data were collected for one complete running gait cycle and from initiation of countermovement to recovery of balance for the standing vertical jump. One gait cycle and jump trial from one participant were used, similar to previous modelling research (Masters & Challis, 2022; Mills et al., 2009; Pain & Challis, 2006), as the purpose of this paper was to understand how model construction effects loading on the spine rather than variability of an individual's performance. In addition, constraints within the laboratory environment meant that the participant was likely not running at a steady velocity and still accelerating; given the same input kinematics were used to drive all models, the spinal loading can still be compared directly.

2.2. Data processing

Artec Studio 17 Professional (Artec3D, Luxembourg) was used to process the torso/breast scans in high-definition with a 3D resolution of 0.5 mm. Using the marked breast boundary each breast was extracted from the torso. The posterior of the breast (simulated chest wall) was filled using the "fix holes" function and volume was obtained using the "measurements" function within the software. Each breast mass was calculated using the

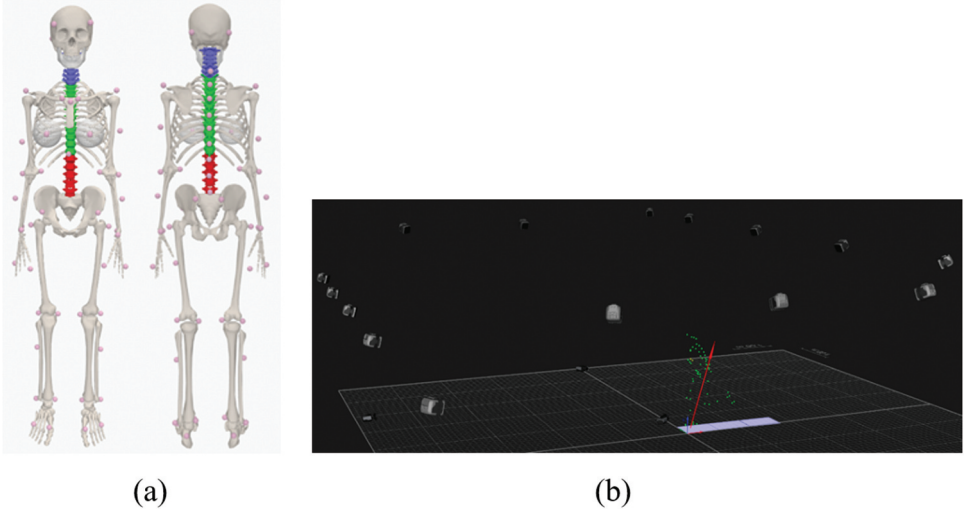

Figure 1. (a) Customised motion capture marker placement locations. (b) Laboratory experimental setup.

breast volume and a breast density of 945 kg/m^3 (Sanchez et al., 2017); the mass of each breast was 0.747 kg (right) and 0.754 kg (left). The breast centre of mass position was calculated using the static 3D scanned breast data, and assuming each breast was a geometric solid hemisphere. Theoretically, the centre of mass of a hemisphere lies on the line passing from the apex of the hemisphere (assumed to be at the nipple) to the centre of the base of the hemisphere. The centre of mass was calculated using $Y = 3R/8$; where "Y" was the distance to the centre of mass and "R" was the radius of the hemisphere. The radius was calculated from the nipple marker (breast apex) to the inferior breast boundary, along the chest wall. The centre of mass was at a distance Y anteriorly from the hemisphere base towards the breast apex. A virtual breast centre of mass marker was created in all static and dynamic trials within the Qualisys Track Manager software. All kinematic and kinetic data were processed and exported via a customised MatLab script for importing into OpenSim (Simtk.org). The standard OpenSim workflow (Akhavanfar et al., 2022) was followed to ensure the generic female geometry torso model (Bruno et al., 2017) and was scaled to the participant based on marker data in the static trial. The breasts were not included in the scaling process, and their mass was included within the torso segments (L5-T1). Following successful completion of the scaling process, for both the fixed and dynamic breast models, the mass of each breast was subtracted from the calculated torso mass (16.49 kg) and added to the relevant breast segment, maintaining whole body mass. The Inverse Kinematics Tool in OpenSim was used to find the values for the generalised coordinates (joint angles and positions) in the model that best matched the experimental kinematics using a weighted least squares problem, whose solution aimed to minimise both marker and coordinate errors. The resulting kinematics, filtered with a Butterworth 2nd order low pass filter with a cut-off frequency of 5 Hz (Racz & Kiss, 2021), were then implemented, in conjunction with the recorded ground reaction force data, within the Inverse Dynamics tool to calculate the lumbar and thoracic spinal joint moments during the dynamic trials, as described in section 2.3.

2.3. Musculoskeletal models

A female geometry full body model (Burkhart et al., 2020) with a fully articulated torso consisting of a thoracolumbar spine (T1 through L5) (Figure 1), with 3 rotational degrees-of-freedom at each inter-vertebral joint, and ribcage (24 individual ribs and a sternum) was selected for customisation for this study. The ribcage was assumed to be rigid, and the male geometry version of the model (Bruno et al., 2015) was previously validated for estimations of spinal loading and trunk muscle tension against previously collected in-vivo measurement of intradiscal pressure, vertebral compression from telemeterised implants and trunk muscle EMG. Three torso and breast variations of the scaled female musculoskeletal model were developed for this study. Firstly, in Model 1 (integrated breast model) the breast mass was included within the mass of the torso and the inertia properties of the torso maintained (Figure 2(a)). Secondly, Model 2 (fixed breast model) was adapted from the integrated breast model to include fixed breasts whose mass was separate to that of the torso (Figure 2(b)). Each breast segment was represented as a point mass located at the centre of mass of the breast when in a static standing position. Each breast segment was attached via a weld joint (0 degrees of freedom), which fixed the breast segments to the torso so they were unable to move relative to each other. The inertia properties of the torso remained unaltered, except the mass of the scaled torso was reduced by 1.501 kg (the total breast mass). Finally, Model 3 (dynamic breast model) was adapted from the fixed breast model to include dynamic breasts whose mass could move separate to that of the torso (Figure 2(c)). Similar to Model 2, the torso construction and mass remained the same. Each breast segment was represented by a point mass and attached via a 3 degree of freedom sliding joint to the torso. This design permitted translation of the breast segment relative to the torso in three planes and the force at the joint to be output. The initial position of each breast segment was defined by the centre of mass position of the breast during standing.

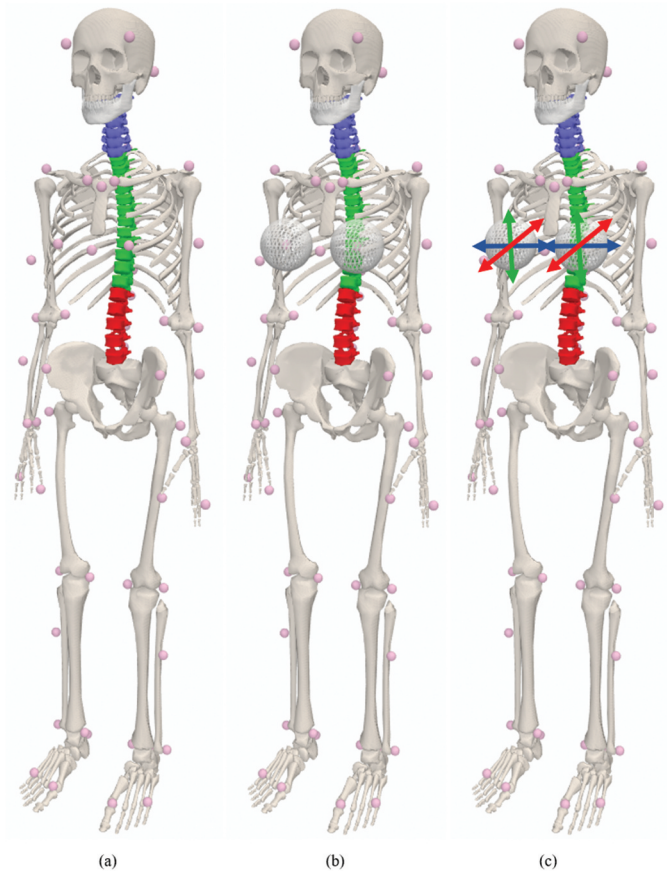

Figure 2. Graphical representation of the three musculoskeletal models: (a) integrated breast model, (b) fixed breast model, and (c) dynamic breast model.

2.4. Data analysis

The flexion/extension, lateral bend, and axial-rotation angle of the torso body (defined by markers on the sternal notch, xiphoid process, 7th cervical vertebra and 8th thoracic vertebra) (Wu et al., 2005) were recorded in the global coordinate system, for each activity (running and jumping). The

breast centre of mass position and breast joint force (the force applied by the breasts on the torso at the sliding joint) was calculated in the local coordinate system of the torso body in three directions (anterior-posterior; medial-lateral; superior-inferior) within Model 3 (dynamic breast) to understand that effect of breast motion on spinal joint moments. All spinal joint moment data were normalised to the participant's whole-body mass (74.6 kg). For each activity, net joint moments acting at each lumbar spine joint (L1–L5) and each thoracic joint (T1–T12) were averaged at each instant in time to create a single net lumbar and thoracic moment time history in each direction (flexion/extension; left/right bending; axial clockwise/anticlockwise rotation) (Raabe & Chaudhari, 2016). Each time history was also normalised to 101 data points (i.e., 0–100% of the gait cycle or jump at 1% increments). Key events within the gait cycle, such as right and left foot contact and take off and landing within the jumping trial, were determined from the vertical ground reaction force data and highlighted on the time histories. The minimum and maximum spinal moment in each direction was reported as a peak moment for both the running gait cycle or vertical jump. Additionally, the net lumbar and thoracic joint moment over the running gait cycle and vertical jump was calculated. Finally, as this study sought to understand the extent to which the proposed modelling decisions affect the absolute values of estimated spinal joint moments, only reporting the absolute differences between the estimated spinal moments results could has been misleading because the interpretation of the differences relates to the magnitude of predicted spinal moments. Given there is no "correct" value by which to scale the difference between the modelling approaches, the relative differences in the predicted maximum, minimum and net spinal moments with respect to the integrated breast model (Model 1) were also calculated for each activity (Akhavanfar et al., 2022) (Equation 1 – example for maximum lumbar moments) and presented as a percentage.

$$\text{Relative Difference}_{\text{model i}} = \frac{max\ Lumbar_{\text{model i}} max\ Lumbar_{\text{model 1}}}{max\ Lumbar_{\text{model 1}}} \times 100 \quad (1)$$

3. Results

3.1. Running

During the running gait cycle, the torso angle ranged from −13.6° to −24.6° of flexion; −7.6° left bend and 10.1° right bend; 16.8° anticlockwise to −25.5° clockwise rotation, when viewed from above (Figure 3(a)). The participant did not achieve steady-state running velocity, hence the misalignment of the start and end of the trajectories as shown in Figure 3(a). The largest breast joint force was observed in the superior/inferior direction (left breast: 23.5 N, right breast: 21.3 N), which was greatest when the breast was near its most inferior position (Figure 4). Lower peak magnitudes of breast joint force were observed in the anterior/posterior (left breast: −7.9 N, right breast: −8.4 N) and medial/lateral directions (left breast: −4.0 N, right breast: 5.1 N) (Figure 4)

Net lumbar spine moment was similar between the integrated breast (Model 1) and fixed breast (Model 2) models with a difference of <0.02 Nm/kg (6%) within the running gait cycle (Table 1). However, the dynamic breast model (Model 3) exhibited greater differences in net lumbar spinal moments, ranging from increases of 0.05 Nm/kg (+21%) in flexion/extension and decreases of 0.10 Nm/kg (−500%) in the axial rotation, compared with Model 1 within the gait cycle. Peak lumbar spine moments also showed small differences between Models 1 and 2 (up to 0.05 Nm/kg) with increased differences associated with Model 3 (0.55 Nm/kg; −89%) during flexion. When examining the lumbar spine joint moment time histories, the magnitudes varied between models at each time point, and the greatest difference (0.86 Nm/kg; 68%) occurred at 90% of the running gait cycle during axial rotation (Figure 5(a)). Thoracic spine moments revealed a similar pattern to the lumbar spine with minimal differences

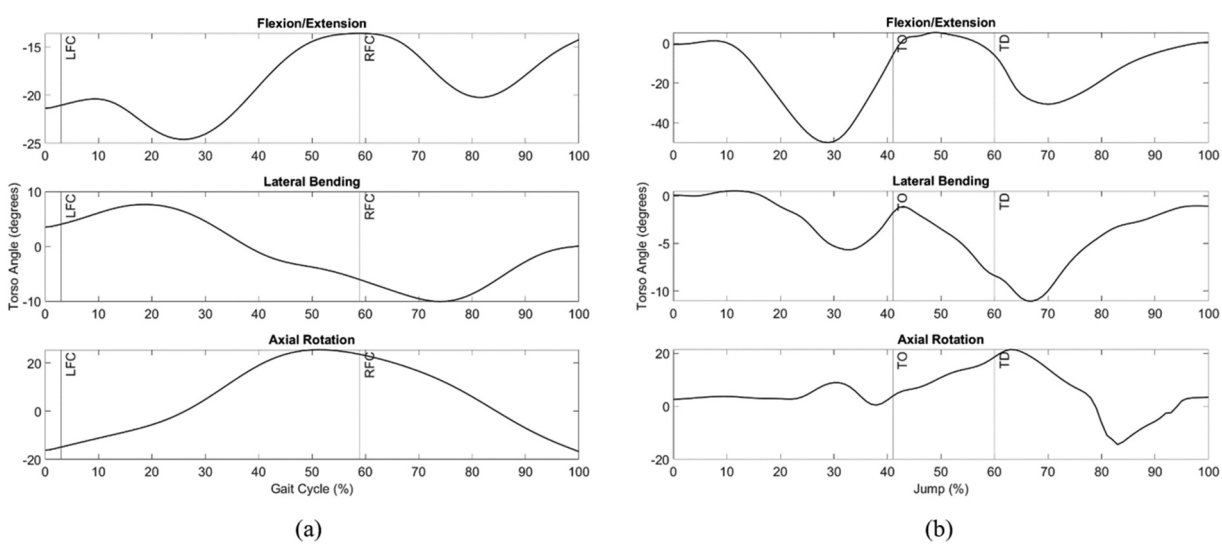

Figure 3. Torso angle during (a) one running gait cycle (LFC = left foot contact; RFC = right foot gait contact) and (b) a vertical jump (TO = take Off; TD = touch down) in the global coordinate system. (Flexion (-), extension (+); left bend (-), right bend (+); clockwise rotation (-), anticlockwise rotation (+)).

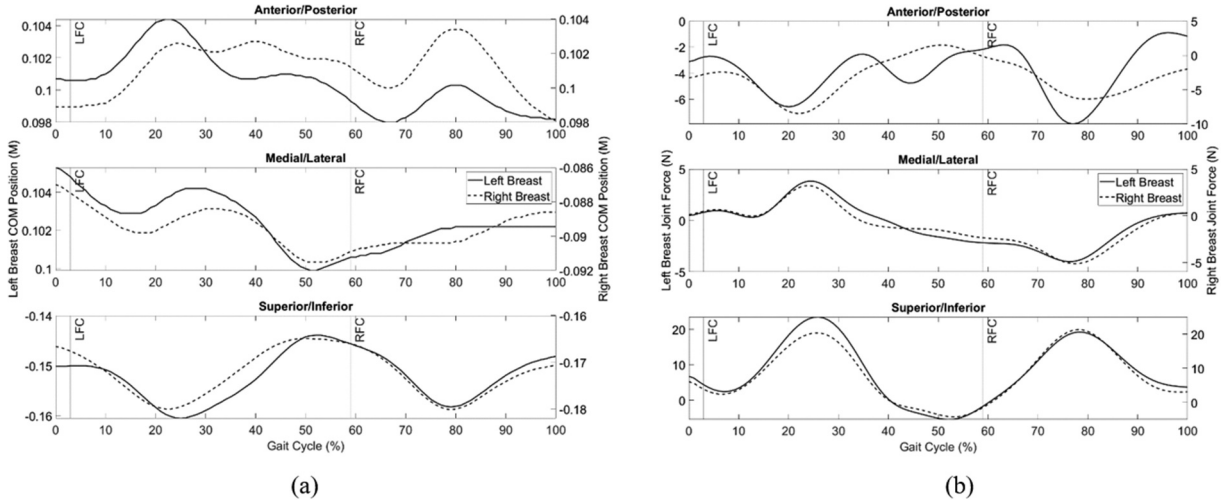

Figure 4. Left and right (a) breast centre of mass (COM) position and (b) breast joint force, in the local coordinate system of the torso during one running gait cycle (LFC = left foot contact; RFC = right foot gait contact).

Table 1. Peak and net lumbar and thoracic joint moments during one running gait cycle (Nm/kg), including a relative percentage difference to the integrated torso-breast model in brackets.

			Models		
Spine region	Peak or Net	Direction	1	2	3
Lumbar	Peak	Flexion	−0.63	−0.62 (−2%)	−0.07 (−89%)
		Extension	1.18	1.22 (3%)	1.32 (12%)
		Left Bend	−0.65	−0.66 (2%)	−0.59 (−9%)
		Right Bend	0.75	0.76 (1%)	0.60 (−20%)
		Clockwise	−1.28	−1.23 (−4%)	−0.71 (−45%)
		Anticlockwise	1.07	1.07 (0%)	0.71 (−34%)
	Net	Flexion/Extension	0.33	0.35 (6%)	0.40 (21%)
		Lateral Bend	0.04	0.04 (0%)	0.06 (50%)
		Axial Rotation	−0.02	−0.02 (0%)	0.08 (−500%)
Thoracic	Peak	Flexion	−0.53	−0.53 (0%)	−0.08 (−85%)
		Extension	0.45	0.50 (11%)	0.44 (−2%)
		Left Bend	−0.55	−0.55 (0%)	−0.41 (−25%)
		Right Bend	0.64	0.64 (0%)	0.42 (−34%)
		Clockwise	−0.69	−0.69 (0%)	−0.42 (−39%)
		Anticlockwise	0.59	0.59 (0%)	0.40 (−32%)
	Net	Flexion/Extension	0.10	0.11 (10%)	0.13 (30%)
		Lateral Bend	−0.02	−0.02 (0%)	0.01 (−150%)
		Axial Rotation	0.00	0.00 (NaN)	0.06 (NaN)

Flexion (-), extension (+); left bend (-), right bend (+); clockwise rotation (-), anticlockwise rotation (+); NaN = not a number.

(0.01 Nm/kg; 10%) in net joint moments between Models 1 and 2, yet up to 0.02 Nm/kg (30%) increases or 0.03 Nm/kg (−150%) decreases, when compared to Model 3. Peak thoracic spine moments also showed small differences between Models 1 and 2 (0.05 Nm/kg; 11%) with increased differences associated with Model 3 (0.45 Nm/kg; −85%)

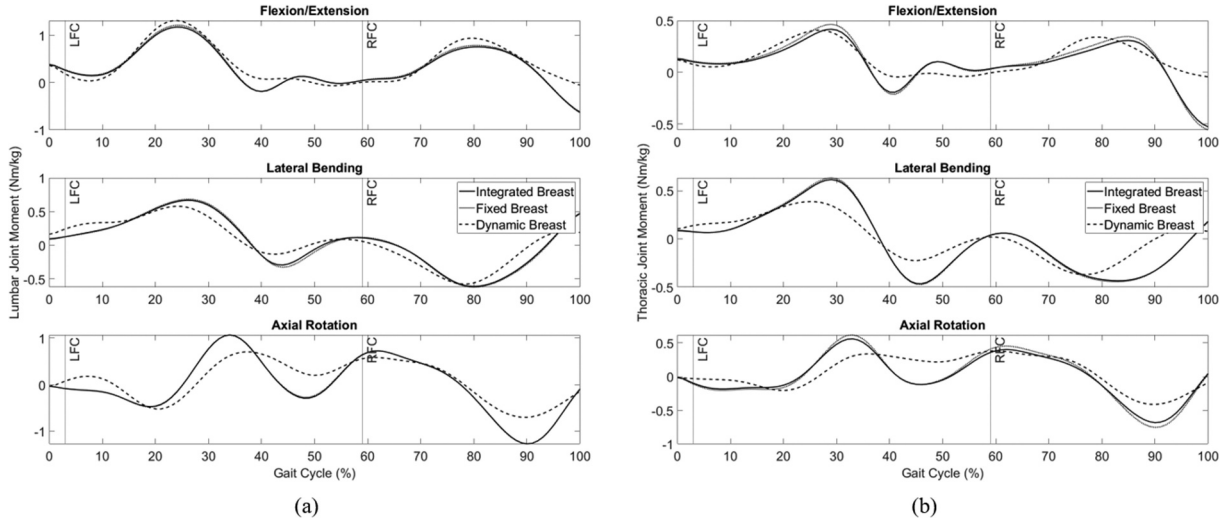

Figure 5. Average (a) lumbar and (b) thoracic joint moments during one running gait cycle (LFC = left foot contact; RFC = right foot gait contact). (Flexion (-), extension (+); left bend (-), right bend (+); clockwise rotation (-), anticlockwise rotation (+)).

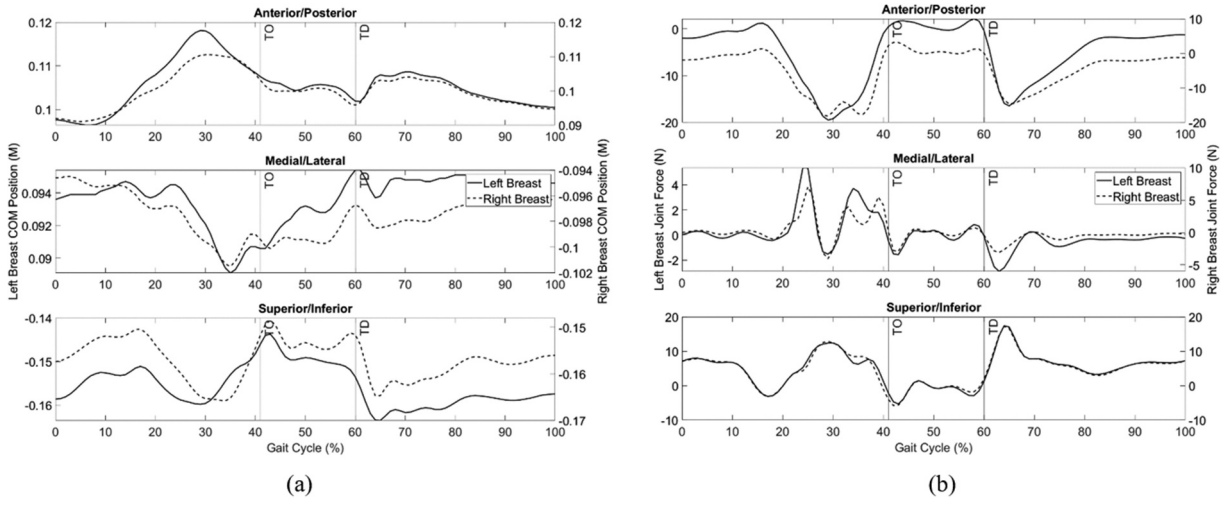

Figure 6. Left and right (a) breast centre of mass (COM) position and (b) breast joint force, in the local coordinate system of the torso during a vertical jump (TO = take off; TD = touch down).

during flexion. When examining the thoracic spine joint moment time histories, the magnitudes again varied between models at each time point, and the greatest difference was 0.48 Nm/kg (92%) (flexion/extension) at 100% of the gait cycle, when compared to Model 3 (Figure 5(b)).

3.2. Jumping

During the standing vertical jump, the torso angle ranged from −49.9° of flexion to 5.5° of extension; −0.5° left bend and 11.1° right bend; 21.4° anticlockwise to −14.5° clockwise rotation (Figure 3b). The largest joint breast force was observed in the anterior/posterior direction (left breast: −19.5 N, right breast: −18.2 N) (Figure 6), which was greatest when the torso was at the maximum flexion. Similar peak magnitudes of breast joint force were observed in the superior/inferior direction (left breast: 17.4 N, right breast: 17.2 N) and lower breast forces in the medial/lateral direction (left breast: 5.4 N, right breast: 7.1 N) (Figure 6).

During the vertical jump, net lumbar spine moments were similar between all female torso models, the integrated breast (Model 1) and fixed breast (Model 2) models exhibited a difference of 0.01 Nm/kg (2%), whilst the dynamic breast model (Model 3) reported a small increase 0.03 Nm/kg (10%), compared to Model 1 (Table 2). Flexor/extensor lumbar spine moment time histories were similar in magnitude and timings between the three models; however, the lumbar spine moments for lateral bending and axial rotation showed marked differences between both Models 1 and 2, when compared to Model 3. The dynamic breast model (Model 3) underestimated the lumbar spine axial rotation moment by 0.89 Nm/kg (82%), when compared to the integrated breast (Model 1) and fixed breast (Model 2) at 30–40% of the jump (just prior to take off) (Figure 7(a)). Similar differences in magnitude and timings occurred within the thoracic spine moments in both the lateral

Table 2. Peak and net lumbar and thoracic joint moments during a vertical jump (Nm/kg), including relative percentage to the integrated torso-breast model in brackets.

			Models		
Spine region	Peak or Net	Direction	1	2	3
Lumbar	Peak	Flexion	−0.72	−0.71 (−1%)	−0.64 (−11%)
		Extension	2.40	2.40 (0%)	2.34 (−3%)
		Left Bend	−0.46	−0.47 (2%)	−0.33 (−28%)
		Right Bend	0.50	0.50 (0%)	0.30 (−40%)
		Clockwise	−0.68	−0.68 (0%)	−0.44 (−35%)
		Anticlockwise	1.11	1.12 (1%)	0.49 (−56%)
	Net	Flexion/Extension	0.41	0.42 (2%)	0.45 (10%)
		Lateral Bend	0.00	0.00 (NaN)	−0.01 (NaN)
		Axial Rotation	−0.02	−0.02 (0%)	−0.01 (−50%)
Thoracic	Peak	Flexion	−0.71	−0.71 (0%)	−0.49 (−31%)
		Extension	0.96	0.96 (0%)	0.73 (−24%)
		Left Bend	−0.59	−0.59 (0%)	−0.28 (−53%)
		Right Bend	0.44	0.44 (0%)	0.31 (−30%)
		Clockwise	−0.45	−0.44 (−2%)	−0.35 (−22%)
		Anticlockwise	0.96	0.97 (1%)	0.33 (−66%)
	Net	Flexion/Extension	0.12	0.13 (8%)	0.11 (−8%)
		Lateral Bend	−0.01	−0.01 (0%)	−0.01 (0%)
		Axial Rotation	−0.01	−0.01 (0%)	−0.01 (0%)

Flexion (-), extension (+); left bend (-), right bend (+); clockwise rotation (-), anticlockwise rotation (+); NaN = not a number.

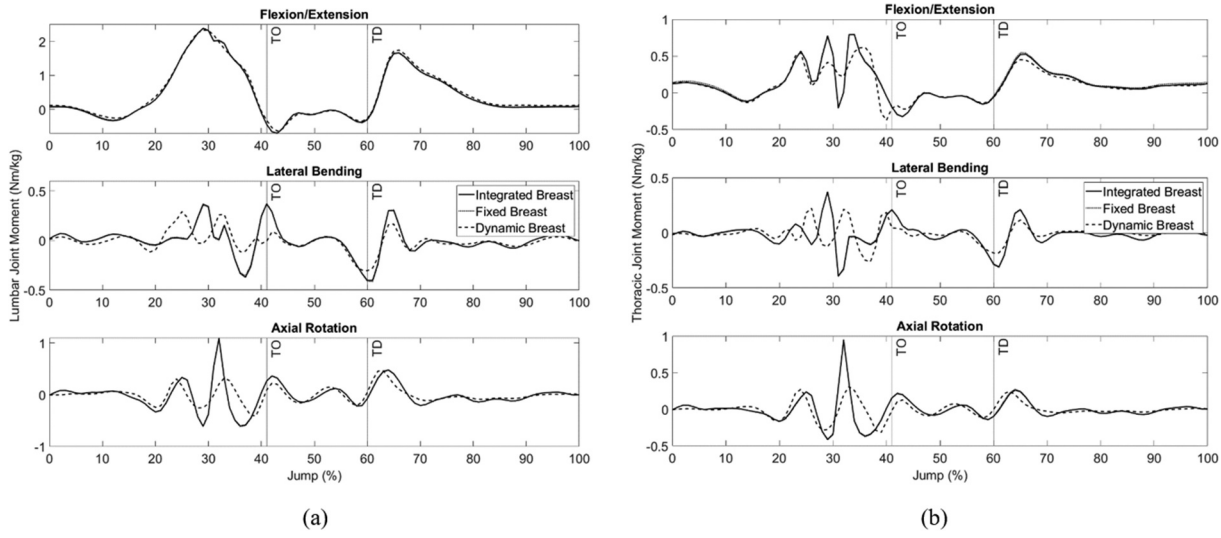

Figure 7. Average (a) lumbar and (b) thoracic joint moments during a vertical jump (TO = take Off; TD = touch down). (flexion (-), extension (+); left bend (-), right bend (+); clockwise rotation (-), anticlockwise rotation (+)).

bending and axial rotation moment time histories (Figure 7(b)). However, more notable differences also occurred during the flexor/extension thoracic spine moment time history, with an underestimation (0.39 Nm/kg; 49%) of extension moments in

Model 3, compared to Models 1 and 2, at 28–38% of the vertical jump (Figure 7(b)).

4. Discussion

The aim of this study was to investigate the effect of three female torso and breast modelling approaches upon lumbar and thoracic spinal loading during running and jumping. Key findings suggest that during physical activity, such as running, a fixed separation of the breast mass from the torso (Model 2) resulted in altered lumbar and thoracic joint moments (up to 0.05 Nm/kg; 4%); however, greater differences were reported when breast motion was incorporated within the torso model (Model 3), resulting in up to 0.89 Nm/kg (89%) changes in spinal joint moments. Similarly, during the vertical jump trial, when the motion of the breasts were included within the torso model, spinal moments were up to 0.74 Nm/kg (80%) lower than those predicted by either Model 1 or 2. Interestingly, spinal moment time histories displayed key differences (up to 0.39 Nm/kg; 49%) between models suggesting that breast motion had a direct impact upon both the magnitude and timing of spinal loading during physical activity. Finally, differences in the magnitude and timing of joint moments between the lumbar and thoracic spine were also evident, reinforcing the importance of examining both regions of the spine when incorporating breast mass and motion into female musculoskeletal models.

The torso range of motion was similar to previous research during a running gait cycle (Milligan et al., 2015) and a vertical jump (Khuu et al., 2015), the magnitude of peak breast forces (~22 N) were also similar to those previously reported for a similar breast mass (van Oeveren et al., 2021) during running. Minimal changes in spinal joint moments were reported between the integrated breast (Model 1) and fixed breast model (Model 2) during either running or jumping. The integrated breast model included the mass of the breasts within the torso, evenly distributing the breast mass within the geometry of the female torso. Although the fixed breast model (Model 2) included the breasts as separate segments attached via a weld joint to the anterior of the torso, this design approach only acts to move the centre of mass of the torso towards the anterior of the torso and altering the inertia tensor of the torso. The altered anterior position of the centre of mass without movement independent of the torso caused minimal changes (<0.05 Nm/kg) in spinal moments in order to maintain the segment kinematics during physical activity. This finding suggests that the differences in model complexity and design between Models 1 and 2 is not sufficient when considering clinical applications, since it has previously been reported that spinal moments between back pain and non-back pain sufferers vary by a similar magnitude to the difference between these two modelling approaches (Hasegawa et al., 2018). The inclusion of three degrees of freedom sliding joint between the breast and torso enabled the breast mass to move relative to the torso and induce a force on the torso resulting from the breast segment motion. This varying force, due to its weight and the acceleration of the breast segment over time, was applied to the torso causing a change in joint moments required to maintain whole body kinematics during running and jumping. This modelling approach used for the female torso and breast alters the calculated spinal moments by up to 0.89 Nm/kg during running and jumping; suggesting the inclusion of breast mass and motion (Model 3) is clinically significant for back pain and an important design feature for future musculoskeletal models using female participants.

The torso and breast modelling approaches resulted in changes in the magnitude of net and peak joint moments during physical activity. For example, the increase in net lumbar flexion/extension spinal moments during running and jumping associated with breast motion in the musculoskeletal model (Model 3) may require additional muscular demand to maintain torso kinematics and hence increased energy expenditure (Kyrolainen et al., 2001). Thus, when considering repeated running gait cycles or jumping within sports such as marathon racing or volleyball, the energy demand requirements may be mis-represented in female models that neglect breast motion. Furthermore, the results from this study have also highlighted up to 87% reductions in peak joint moments between a female torso model with (Model 2) and without (Model 3) breast mass motion. Previous work in sports injuries has reported a significant 52% reduction in flexion/extension or a significant 20% reduction in lateral flexion peak lumbar spine joint moments between injured and non-injured athletes in sports such as cricket (Bayne et al., 2016). The reported link between spinal moments and injuries in sports further highlights the need for accurate representation of spinal moments using female musculoskeletal models in future research.

The temporal changes in spinal moment time histories between the torso and breast modelling approaches illustrate how the inclusion of breast motion and hence force affected the magnitude of the spinal moments during the running gait cycle or vertical jump. During running, the greatest deviations in Model 3 and Model 1 (flexion/extension) occur at approximately 40% and 100% of the gait cycle, this corresponded to when the breast force was approximately zero. At these points in the gait cycle, the total torso mass (torso + breasts) was effectively lower, than Model 1 or 2, reducing the spinal moments required to maintain kinematics. A similar pattern also existed within lateral bending and axial rotation, as the greatest deviations within joint moments, between modelling approaches, occurred when the breast force was approximately zero. These temporal differences in joint moments are more noticeable within the thoracic region during jumping where Models 1 and 2 appear to over-estimate the spinal moments for the majority of the jump. At 20–40% of the jump trial (just prior to take off), there were notable differences in thoracic spinal moments between models. Examining a combination of joint moments, breast force and position and torso angle data within this time period helps to understand the possible underpinning mechanics of differences between models. The motion of the breasts and subsequent forces transferred to the torso reduce the thoracic spinal moments when breast forces are low (~22% of jump time) or when the breast motion is opposite to the torso (29–35% of jump time). The breast force can also contribute to the extension or flexion moment of the torso (for example, the breasts move superiorly whilst the torso extends during the take-off phase of a vertical jump, reducing the extension moment within the spine). The lack of

synchronisation (or lag) between breast and torso motion (Scurr et al., 2009) reduced spinal moments in this study. Additionally, this underpinning mechanism of relative soft tissue motion and timing has also been reported within other areas of the body, demonstrating that the inclusion of soft tissue motion within musculoskeletal models can reduce estimates of internal loading (Pain & Challis, 2006).

The spinal model used within this study incorporated intervertebral joint movement within the lumbar and thoracic regions. The results of this study confirm the importance of including both regions of the spine in future research as the torso and breast modelling approach used, influencing these regions differently. There was an increased difference in the temporal aspects between torso models within the thoracic region. Furthermore, whilst the percentage differences between models in peak moments were often greater in the thoracic region, particularly during jumping, the absolute magnitudes were greater within the lumbar spine. These differences between spine regions are likely due to the breast segments being located within the thoracic region of the spine. It is suggested that the breasts have a localised influence on thoracic spinal moments but, given the greater distance and longer moment arm to the lumbar spine, especially when jumping, the magnitudes of spinal moments were greater within this lumbar region.

The findings within this study offer a new insight into the effect of female torso segment design within musculoskeletal modelling. Whilst this study has implemented a new dynamic breast model (Model 3) within the context of physical activity, the findings of this study have wider reaching implications on other activities of daily living that involve impacts with the ground such as ascending and descending stairs or other devices. Future work associated with predicting surgical outcomes such as unilateral mastectomy surgery following breast cancer could provide greater insights into the subsequent effect of surgery on the musculoskeletal system, muscle activation and personalised rehabilitation advice. Sports apparel and bra designers could utilise this type of approach within forward dynamics solutions to understand how the viscoelastic properties of bra materials could influence breast motion and subsequent internal loading in females. Finally, the occupational and commercial sector could benefit from utilising the modelling approaches highlighted within this study to aid ergonomic design of equipment, such as chairs, or recommendations on posture to minimise spinal loading for females within the workplace.

The differences observed between the integrated breast, fixed breast and dynamic breast models indicate that whilst separating the breast mass from the torso improved the representation of the geometry of the female torso during physical activity, the estimation of the spinal loading is substantially different when incorporating breast motion. The inclusion of breast motion (Model 3) resulted in further changes in lumbar and thoracic spine moments, particularly during jumping, when compared to the commonly used integration of breast mass into the torso approach. During high impact physical activity (running and jumping), the motion of the breasts can counter the movement of the torso; hence, the integrated breast or fixed breast model over or underestimate the effect of the breast mass on spine moments. It is recommended that not only the breast mass but also breast motion are considered when developing female musculoskeletal models in the future.

Although the results of this study demonstrate differences in spinal loading between model designs, the limitations associated with this modelling approach are important to discuss. The actual data presented may vary given the single participant specific nature of the model; thus, the individual gait and jumping kinematics of the participant may influence the exact spinal moments reported. However, it is also important to acknowledge that the same input kinematics were used throughout and the influence of model design investigated. The inverse dynamics approach depends upon the quality of the kinematics data used to drive the model, and a future forward dynamics model would enable manipulation of viscoelastic soft tissue properties to determine how they affect soft tissue motion. Furthermore, these inverse dynamics solutions do not directly consider how muscular contributions influence joint moments and the possible effects of muscular co-contraction on joint loading. Therefore, whilst the results may infer possible changes in muscular demand due to breast motion, future modelling research could examine whether muscular effort is increased or decreased to maintain torso kinematics when breast motion occurs. The breast kinematics were captured using a marker placed on the bra over the breast; although careful bra fit criteria were followed, it may be possible for relative motion between the breast tissue and bra to occur, altering the magnitude of actual breast motion and hence possibly spinal loading.

In conclusion, whilst caution must be applied when interpreting the magnitude of differences in spinal loading between the models in this study, there is evidence to suggest that the modelling approach used to incorporate the breast mass and motion into female musculoskeletal models should be considered when attempting to estimate spinal loading in females during physical activity.

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Chris Mills and Melissa Jones significantly contributed to the study conception and design. Material preparation, data collection and analysis were performed by Chris Mills. The first draft of the manuscript was written by Chris Mills, and all authors commented on previous versions of the manuscript. All authors read and approved the final manuscript.

Consent for publication

The authors confirm that this manuscript is not submitted for publication elsewhere and approve the final version.

Consent to participate

The participant provided informed written consent to participate in the experiments within the present study.

Data availability statement

The data that support the findings of this study are available from the corresponding author, [CM], upon reasonable request.

Ethics approval

Experiments in the present study were approved by the University of Portsmouth Research Ethics Committee (SFEC 2018–043).

References

Akhavanfar, M., Uchida, T., Clouthier, A., & Graham, R. (2022). Sharing the load: Modeling loads in OpenSim to simulate two-handed lifting. *Multibody System Dynamics*, *54*(2), 213–234. https://doi.org/10.1007/s11044-021-09808-7

Alvim, F., Muniz, A., Lucareli, P., & Menegalo, L. (2019). Kinematics and muscle forces in women with patellofemoral pain during the propulsion phase of the single leg triple hop test. *Gait & Posture*, *73*, 108–115. https://doi.org/10.1016/j.gaitpost.2019.07.193

Banks, J., Alemi, M., Allaire, B., Lynch, A., Bouxsein, M., & Anderson, D. (2023). Using static postures to estimate spinal loading during dynamic lifts with participant-specific thoracolumbar musculoskeletal models. *Applied Ergonomics*, *106*, e103869. https://doi.org/10.1016/j.apergo.2022.103869

Bayne, H., Elloitt, B., Campbell, A., & Alderson, J. (2016). Lumbar load in adolescent fast bowlers: A prospective injury study. *Journal of Science & Medicine in Sport*, *19*(2), 117–122. https://doi.org/10.1016/j.jsams.2015.02.011

Bridgman, C., Scurr, J., White, J., Hedger, W., & Galbraith, H. (2010). Three-dimensional kinematics of the breast during a two-step star jump. *Journal of Applied Biomechanics*, *26*(4), 465–472. https://doi.org/10.1123/jab.26.4.465

Bruno, A., Bouxsein, M., & Anderson, D. (2015). Development and validation of a Musculoskeletal model of the fully articulated thoracolumbar spine and rib cage. *Journal of Biomechanical Engineering*, *137*(8), e081003. https://doi.org/10.1115/1.4030408

Bruno, A., Burkhart, K., Allaire, B., Anderson, D., & Bouxsein, M. (2017). Spinal loading patterns from biomechanical modeling explain the high incidence of vertebral fractures in the thoracolumbar region. *Journal of Bone and Mineral Research*, *32*(6), 1282–1290. https://doi.org/10.1002/jbmr.3113

Burkhart, K., Grindle, D., Bouxsein, M., & Anderson, D. (2020). Between-session reliability of subject-specific musculoskeletal models of the spine derived from optoelectronic motion capture data. *Journal of Biomechanics*, *112*, 110044. https://doi.org/10.1016/j.jbiomech.2020.110044

Cazzola, D., Holsgrove, T., Preatoni, E., Gill, H., Trewartha, G., & Hind, K. (2017). Cervical spine injuries: A whole-body musculoskeletal model for the analysis of spinal loading. *PLOS ONE*, *12*(1), e0169329. https://doi.org/10.1371/journal.pone.0169329

Coltman, C., Steele, J., & McGhee, D. (2018). Can breast characteristics predict upper torso musculoskeletal pain? *Clinical Biomechanics*, *53*, 46–53. https://doi.org/10.1016/j.clinbiomech.2018.02.002

Delp, S., Anderson, F., Arnold, A., Loan, P., Habib, A., John, C., Guendelman, E., & Thelen, D. (2007). OpenSim: Open-source software to create and analyze dynamic simulations of movement. *IEEE Transactions on Biomedical Engineering*, *54*(11), 1940–1950. https://doi.org/10.1109/TBME.2007.901024

Gittoes, M., Brewin, M., & Kerwin, D. (2006). Soft tissue contributions to impact forces simulated using a four-segment wobbling mass model of forefoot–heel landings. *Human Movement Science*, *25*(6), 775–787. https://doi.org/10.1016/j.humov.2006.04.003

Haddox, A., Hausselle, J., & Azoug, A. (2020). Changes in segmental mass and inertia during pregnancy: A musculoskeletal model of the pregnant woman. *Gait & Posture*, *76*, 389–395. https://doi.org/10.1016/j.gaitpost.2019.12.024

Hasegawa, T., Katsuhira, J., Oka, H., Fujii, T., Matsudaira, K., & de Lussanet, M. H. E. (2018). Association of low back load with low back pain during static standing. *PLOS ONE*, *13*(12), e0208877. https://doi.org/10.1371/journal.pone.0208877

Khuu, M., Musalem, L., & Beach, T. (2015). Verbal instructions acutely affect drop vertical jump biomechanics—Implications for athletic performance and injury risk assessments. *The Journal of Strength & Conditioning Research*, *29*(10), 2816–2826. https://doi.org/10.1519/JSC.0000000000000938

Kyrolainen, H., Belli, A., & Komi, P. (2001). Biomechanical factors affectingrunning economy. *Medicine & Science in Sports and Exercise*, *33*(8), 1330–1337. https://doi.org/10.1097/00005768-200108000-00014

Lee, H., Hong, K., & Kim, E. (2004). Measurement protocol of women's nude breasts using a 3D scanning technique. *Applied Ergonomics*, *35*(4), 353–359. https://doi.org/10.1016/j.apergo.2004.03.004

Masters, S. E., & Challis, J. H. (2022). The effects of wobbling mass components on joint dynamics during running. *Journal of Applied Biomechanics*, *38*(2), 69–75. https://doi.org/10.1123/jab.2021-0051

Milligan, A., Mills, C., Corbett, J., & Scurr, J. (2015). The influence of breast support on torso, pelvis and arm kinematics during a five kilometer treadmill run. *Human Movement Science*, *42*, 246–250. https://doi.org/10.1016/j.humov.2015.05.008

Mills, C., Loveridge, A., Milligan, A., & Scurr, J. (2016). Trunk marker sets and the subsequent calculation of trunk and breast kinematics during treadmill running. *Textile Research Journal*, *86*(11), 1128–1136. https://doi.org/10.1177/0040517515609257

Mills, C., Pain, M., & Yeadon, M. (2009). Reducing ground reaction forces in gymnastics' landings may increase internal loading. *Journal of Biomechanics*, *42*(6), 671–678. https://doi.org/10.1016/j.jbiomech.2009.01.019

Mills, C., Scurr, J., & Wood, L. (2011). A protocol for monitoring soft tissue motion under compression garments during drop landings. *Journal of Biomechanics*, *44*(9), 1821–1823. https://doi.org/10.1016/j.jbiomech.2011.04.019

Pain, M., & Challis, J. (2006). The influence of soft tissue movement on ground reaction forces, joint torques and joint reaction forces in drop landings. *Journal of Biomechanics*, *39*(1), 119–124. https://doi.org/10.1016/j.jbiomech.2004.10.036

Raabe, M., & Chaudhari, A. (2016). An investigation of jogging biomechanics using the full-body lumbar spine model: Model development and validation. *Journal of Biomechanics*, *49*(7), 1238–1243. https://doi.org/10.1016/j.jbiomech.2016.02.046

Racz, K., & Kiss, R. (2021). Marker displacement data filtering in gait analysis: A technical note. *Biomedical Signal Processing and Control*, *70*, e102974. https://doi.org/10.1016/j.bspc.2021.102974

Rose, J., Mendel, E., & Marras, W. (2013). Carrying and spine loading. *Ergonomics*, *56*(11), 1722–1732. https://doi.org/10.1080/00140139.2013.835870

Sanchez, A., Mills, C., & Scurr, J. (2017). Estimating breast mass-density: A retrospective analysis of radiological data. *The Breast Journal*, *23*(2), 237–239. https://doi.org/10.1111/tbj.12725

Scurr, J., White, H., & Hedger, W. (2009). Breast displacement in three dimensions during the walking and running gait cycles. *Journal of Applied Biomechanics*, *25*(4), 322–329. https://doi.org/10.1123/jab.25.4.322

Scurr, J., White, J., & Hedger, W. (2011). Supported and unsupported breast displacement in three dimensions across treadmill activity levels. *Journal of Sports Sciences*, *29*(1), 55–61. https://doi.org/10.1080/02640414.2010.521944

Trompeter, K., Fett, D., & Platen, P. (2017). Prevalence of back pain in sports: A systematic review of the literature. *Sports Medicine*, *47*(6), 1183–1207. https://doi.org/10.1007/s40279-016-0645-3

Turner, A., & Dujon, D. (2005). Predicting cup size after reduction mammaplasty. *British Journal of Plastic Surgery*, *58*(3), 290–298. https://doi.org/10.1016/j.bjps.2004.11.008

van Oeveren, B., de Ruiter, C., Beek, P., & van Dieën, J. (2021). The biomechanics of running and running styles: A synthesis. *Sports Biomechanics*, *23*(4), 516–554. https://doi.org/10.1080/14763141.2021.1873411

White, J., & Scurr, J. (2012). Evaluation of professional bra fitting criteria for bra selection and fitting in the UK. *Ergonomics*, *55*(6), 704–711. https://doi.org/10.1080/00140139.2011.647096

Wu, G., van der Helm, F., Veeger, H., Makhsous, M., Van Roy, P., Anglin, C., Nagels, J., Karduna, A., McQuade, K., Wang, X., Werner, F., & Buchholz, B. (2005). ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion—Part II: Shoulder, elbow, wrist and hand. *Journal of Biomechanics*, *38*(5), 981–992. https://doi.org/10.1016/j.jbiomech.2004.05.042