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The effects of squatting while pregnant on pelvic dimensions: A computational simulation to understand childbirth

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ABSTRACT

Biomechanical complications of childbirth, such as obstructed labor, are a major cause of maternal and newborn morbidity and mortality. The impact of birthing position and mobility on pelvic alignment during labor has not been adequately explored. Our objective was to use a previously developed computational model of the female pelvis to determine the effects of maternal positioning and pregnancy on pelvic alignment. We hypothesized that loading conditions during squatting and increased ligament laxity during pregnancy would expand the pelvis. We simulated dynamic joint moments experienced during a squat movement under pregnant and non-pregnant conditions while tracking relevant anatomical landmarks on the innominate bones, sacrum, and coccyx; anteroposterior and transverse diameters, pubic symphysis width and angle, pelvic areas at the inlet, mid-plane, and outlet, were calculated. Pregnant simulation conditions resulted in greater increases in most pelvic measurements – and predominantly at the outlet – than for the non-pregnant simulation. Pelvic outlet diameters in anterior-posterior and transverse directions in the final squat posture increased by 6.1 mm and 11.0 mm, respectively, for the pregnant simulation compared with only 4.1 mm and 2.6 mm for the non-pregnant; these differences were considered to be clinically meaningful. Peak increases in diameter were demonstrated during the dynamic portion of the movement, rather than the final resting position. Outcomes from our computational simulation suggest that maternal joint loading in an upright birthing position, such as squatting, could open the outlet of the birth canal and dynamic activities may generate greater pelvic mobility than the comparable static posture.

1. Introduction

Biomechanical complications of childbirth, such as obstructed labor, are a major cause of maternal and newborn morbidity and mortality (Neilson et al., 2003; Say et al., 2014; Tsu and Coffey, 2009). Obstructed labor accounts for 2.8% of maternal deaths worldwide with the majority of those occurring in developing countries (Say et al., 2014); it moreover can lead to devastating injuries such as obstetric fistula in the person giving birth as well as clavicle fracture and brachial plexus rupture in the newborn.

Larger maternal pelvic dimensions are believed to contribute to ease of delivery of the baby, though the exact mechanism by which pelvic alignment facilitates or hinders childbirth is still poorly understood (Reitter et al., 2014). Historically, pelvimetry was conducted to predict cephalopelvic disproportion (a geometric mismatch between the pelvis and presenting part of the fetus) (Ferguson and Sistrom, 2000). However, this assessment is considered ineffectual in that patient management and outcomes remain unchanged (Blackadar and Viera, 2004; WHO Reproductive Health Library, 2018). In developed countries such as Canada, where a laboring person has access to the resources required to manage complications that may arise when normal delivery is attempted, obstetric care providers no longer perform pelvimetry.

Evidence suggests that sometimes only a small expansion of the birth canal is needed to safely manage obstructed labor (Monjok et al., 2012; Verkuyl, 2007). Symphysiotomy, a surgical procedure in which the pubic symphysis is divided to enlarge the pelvis thereby facilitating vaginal delivery, is still supported in developing countries where options for caesarean section may not be available or appropriate (Monjok et al., 2012; Verkuyl, 2007). Similarly, upright maternal positioning has been shown to increase pelvic diameters (Reitter et al., 2014) and reduce biomechanical complications such as shoulder dystocia (Bruner et al., 1998).

Along with the potential to improve pelvic alignment, other – possibly related – benefits to upright birthing positions, such as

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squatting, include the additional force of gravity, shorter labor, more effective uterine contractions, reduced association with assisted delivery, and fewer abnormal heart rate patterns of the fetus (Desseauve et al., 2017; Gupta et al., 2012). However, a clear understanding of how birth positioning affects physiologic outcome is lacking.

Our objectives were, therefore, to use a computational model of the female pelvic region to better understand how pregnancy-related physical parameters – specifically, joint loading during upright squatting and ligament laxity – affect movement of pelvic bones and associated clinical measurements. A deeper understanding of the interrelationship between birthing position and pelvic mechanics has the potential to improve birth outcomes.

2. Methods

A multibody computational model of the female pelvis that was previously described (Hemmerich et al., 2018a) was used to simulate the movement of individual pelvic bones during squatting movements (Fig. 1). Three movement conditions were simulated: squatting by someone who is not pregnant, squatting by someone who is pregnant with "non-pregnant" ligament laxity, and squatting by a pregnant person with increased "pregnant" ligament laxity. A brief description of the model is included to explain how these squatting conditions were simulated.

2.1. Computational model

The three-dimensional pelvic model was developed using Mimics™ segmentation software (Materialise, Belgium) from magnetic resonance images (MRI) of a primiparous, non-pregnant subject lying in supine. Model components included pelvic anatomy (left and right innominate bones, pubic disc, sacrum, and coccyx), as well as the L5 vertebra and femurs. Parts were imported into a multibody dynamic simulation package (RecurDyn™, FunctionBay Inc., Seoul, Korea), where each component was modeled as a rigid segment and major ligaments were represented by groups of springs across the sacroiliac and pubic joints.

2.2. Material properties

Nonlinear stiffness equations were fit to each individual sacroiliac and pubic ligament spring complex based on data presented by Eichenseer et al. (2011) and Dakin et al. (2001), respectively. A 2% pre-strain was applied to each spring element since our previous validation data demonstrated slightly more consistent simulation outcomes with both cadaveric sacral rotation tests in the literature (Miller et al., 1987) and our own MRI pelvimetry results for most measurements (Hemmerich et al., 2018a).

A non-linear contact force was generated at the sacroiliac and pubic symphysis joint surfaces using appropriate "Geo Contact" properties in RecurDyn™ (Hemmerich et al., 2018a) with dynamic friction coefficients set to 0 to simulate hyaline cartilage at these joints (Alsanawi, 2016).

2.3. Boundary conditions

The model was actuated by rotational torques at the lumbosacral and hip joints with movement at the pelvic joints constrained only by contact at the joint surfaces and tension in the springs. The lumbosacral joint was restricted to sagittal plane rotation, while hip joints were permitted spherical articulation at the centre of each femoral head. The L5 vertebra was fixed, with linear mediolateral translation of the femurs enabling convergence and divergence of the hips and innominate bones as needed.

Actuating torques represented the muscle forces exerted during squatting and were derived from actual motion analysis data from pregnant participants in their third trimester and control subjects (height and pre-pregnancy weight-matched, non-pregnant participants) as part of a previous study (Hemmerich et al., 2018b). Hip joint moments from the previous investigation were calculated in the pelvic coordinate system defined according to the Visual3D

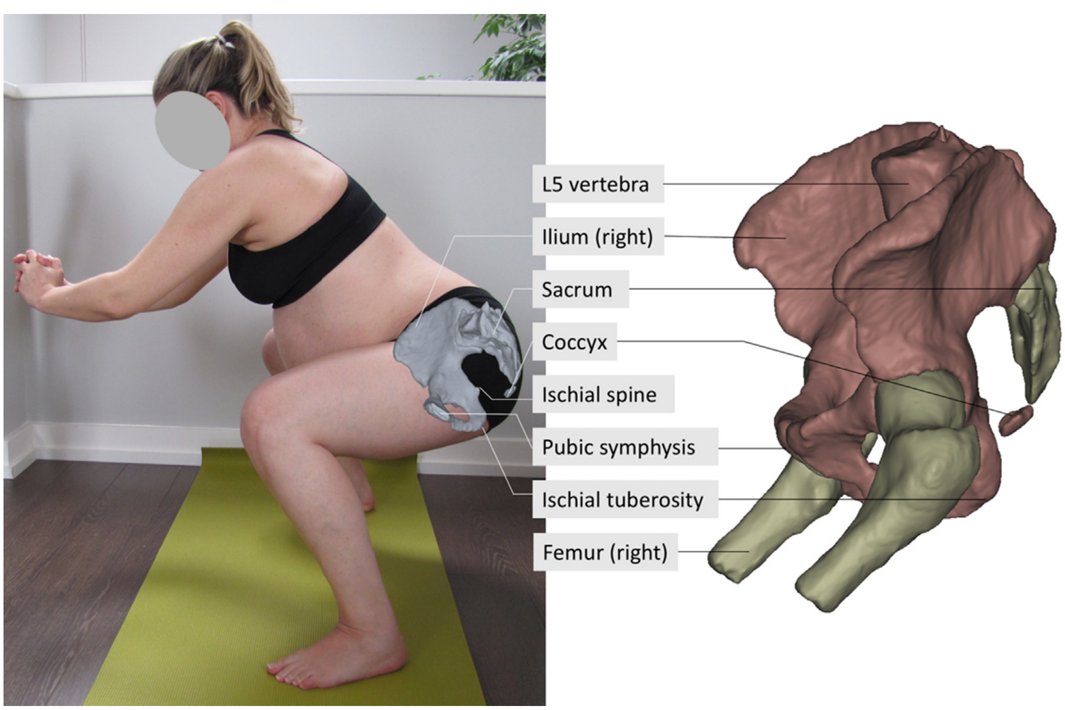


Fig. 1. Left - Position and orientation of pelvis are shown during squatting. Right – An enlarged oblique sagittal view of the model of the pelvic bones generated in Mimics™ software is reoriented to approximately match the position and orientation in squatting.

algorithm (Visual3D Wiki Documentation, 2019) in order to more accurately implement them in the computational model environment. Lumbosacral joint moment calculations were limited to the sagittal plane (i.e. flexion–extension).

2.4. Simulation conditions

Ligament laxity has been shown to increase during pregnancy (Charlton et al., 2001; Dumas and Reid, 1997; Schauberger et al., 1996); it has been reasoned that this phenomenon enables greater pelvic mobility to facilitate parturition (Vleeming et al., 2012; Wolf et al., 2013; Young, 1940). In order to account for this increased laxity, spring stiffness was reduced according to published data for simulations representing pregnant ligament laxity conditions. Based on average load-translation data at 20 and 40-lb force for the anterior cruciate ligament (ACL) in twenty subjects during their third trimester of pregnancy and six weeks postpartum (Charlton et al., 2001), we estimated a pregnancy ligament stiffness of 0.6 times the postpartum (i.e. non-pregnant) values. A detailed explanation of how this value was reached is included in Appendix A. Given the previous research showing that ligament laxity continues to increase until birth and does not fully recover to pre-pregnancy values until more than six weeks postpartum (Schauberger et al., 1996), we expect that our calculated value of ligament stiffness during childbirth is, in fact, a conservative estimate.

In addition to decreasing ligament stiffness for the pregnant condition, pregnant joint torques were used to actuate the pregnant simulation. Due to technical problems during data collection in our previous investigation (Hemmerich et al., 2018b), lumbosacral joint moments were estimated from a second pregnant subject by normalizing and scaling to the mass of the matched pregnant subject. We considered this a valid and accurate representation of lumbosacral moments in our primary pregnant subject since our previous investigation demonstrated closely matching angles and normalized moments with those of the other pregnant participant – and distinctly different from the non-pregnant participants – throughout this activity (Hemmerich et al., 2018b). Hip and lumbosacral joint moment curves generated for both non-pregnant and pregnant subjects are shown in Fig. 2.

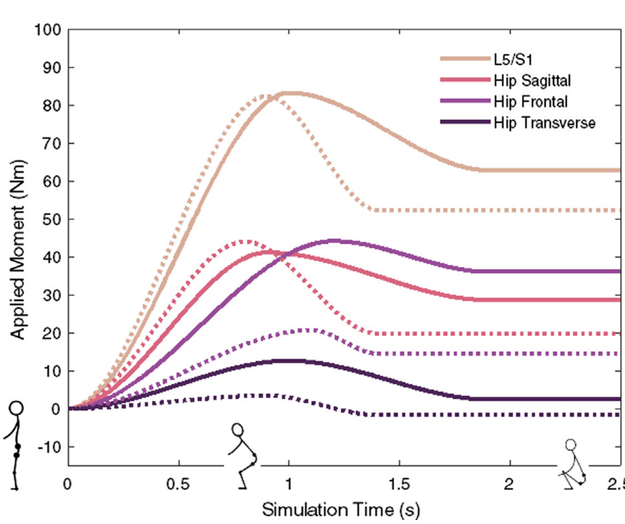


Fig. 2. Pregnant (solid lines) and non-pregnant (dotted lines) joint moments applied to computational model to simulate squatting. Pregnant and non-pregnant simulations were run separately. During each simulation, the following moments were applied simultaneously: L5/S1 joint sagittal plane, hip in sagittal, frontal, and transverse planes for left and right joints. Moments are positive for extension (sagittal plane), abduction (frontal plane), and external rotation (transverse plane).

2.5. Clinical measurements

The simulation software, RecurDyn™, tracked the positions of specified anatomical landmarks on the innominate and sacral bones in the model environment in order to measure clinically relevant dimensions throughout the simulated squatting motion. In addition to traditional anteroposterior (AP) and transverse plane pelvimetric diameters as described by Reitter et al. (2014), we calculated pelvic areas at the inlet, mid-plane, and outlet using anatomical landmarks on the innominate bones and sacrum (Fig. 3). Each triangular pelvic plane was defined using the following landmarks: right and left intersection of pubis and ilium on the pelvic brim together with the sacral promontory for the inlet; the right and left ischial spines as well as the third sacral bone for the midplane area; and the right and left ischial tuberosities together with the tip of coccyx for the outlet. As triangular representations, these reference planes do not portray the full anatomical pelvic area at that plane; results were thus analyzed as change in area percentage (i.e. proportional), rather than discrete measurements, throughout the dynamic squatting motion. By integrating both AP and transverse measurements, however, a more comprehensive overview of changes in pelvic area with maternal positioning was provided.

3. Results

Pregnant simulation data include the combined effect of pregnant squat moments and ligament laxity for comparison with non-pregnant results. Simulation results from individual effects (e.g. pregnant squat moments without increased ligament laxity) are included in Appendix B.

Fig. 4 compares the simulation final squatting position measurements with the results published by Reitter et al. (2014) for both pregnant and non-pregnant analyses; initial pelvimetry measurements are presented in Table 1 for reference. Diameters consistently increased in squatting for both simulation and published data except for the *in vivo* obstetric conjugate data presented by Reitter et al. (2014) which decreased in magnitude in the squat position (Fig. 4). However, the overall divergence in obstetric conjugate measurements between simulation and published data (less than 8 mm) fell well within the limits of variability presented by Reitter et al. (2014).

Measurement increases from standing to squatting were also consistently greater for the pregnant than non-pregnant simulations. Maximum differences between initial and final position measurements predicted by the simulations were similar to the literature for the majority of measurements with non-pregnant bispinous diameter showing the most pronounced discrepancy (19 mm versus 0.7 mm for literature and simulation data, respectively).

Most AP pelvic diameters demonstrated a vast enlargement as the model simulated a descent into squatting (Fig. 5); however, the non-pregnant obstetric conjugate diameter initially decreased before increasing and coming to rest in a position just 0.4 mm greater than the initial measurement. Maximum changes typically occurred during the movement (i.e. prior to reaching the final position) with maximum increases of up to 12 mm for the pregnant squat outlet diameter occurring 0.8 s into the 2.5 s simulation.

Final squatting position transverse diameters were larger for the pelvic outlet (bituberous diameter) and smaller for the inlet measurements when compared to initial dimensions (Fig. 6). That said, for the non-pregnant squat simulation, diameters initially changed in the opposite direction; for example, the bispinous (midplane) and bituberous diameters first decreased before increasing beyond the start position values. Consequently, the lar-

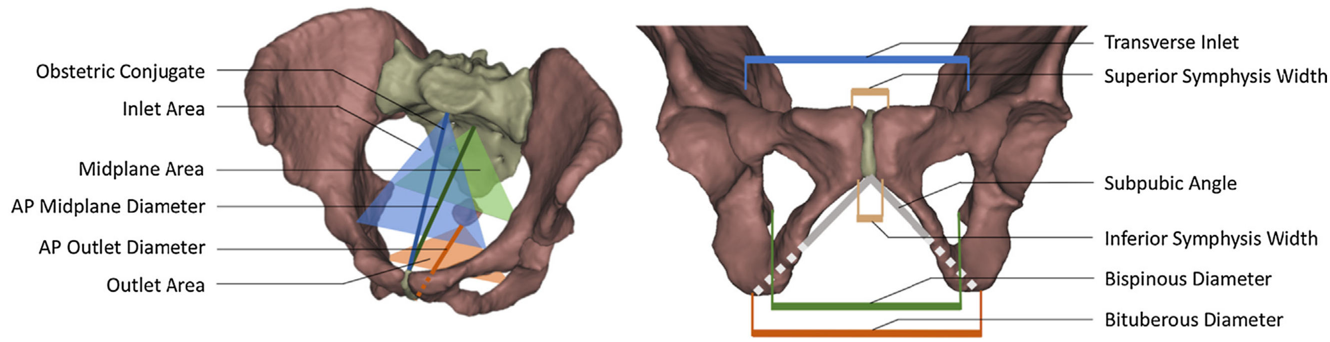


Fig. 3. Clinical dimensions that were measured during each simulation. Figure Left – Oblique view of full pelvis. Pelvic planes are defined using the following anatomical landmarks: inlet (blue) - right and left intersection of pubis and ilium on the pelvic brim, sacral promontory; Midplane (green) - right and left ischial spines, third sacral bone; Outlet (orange) - right and left ischial tuberosities, tip of coccyx. Figure Right – Close-up, frontal view of pubic area with sacrum removed for clarity. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

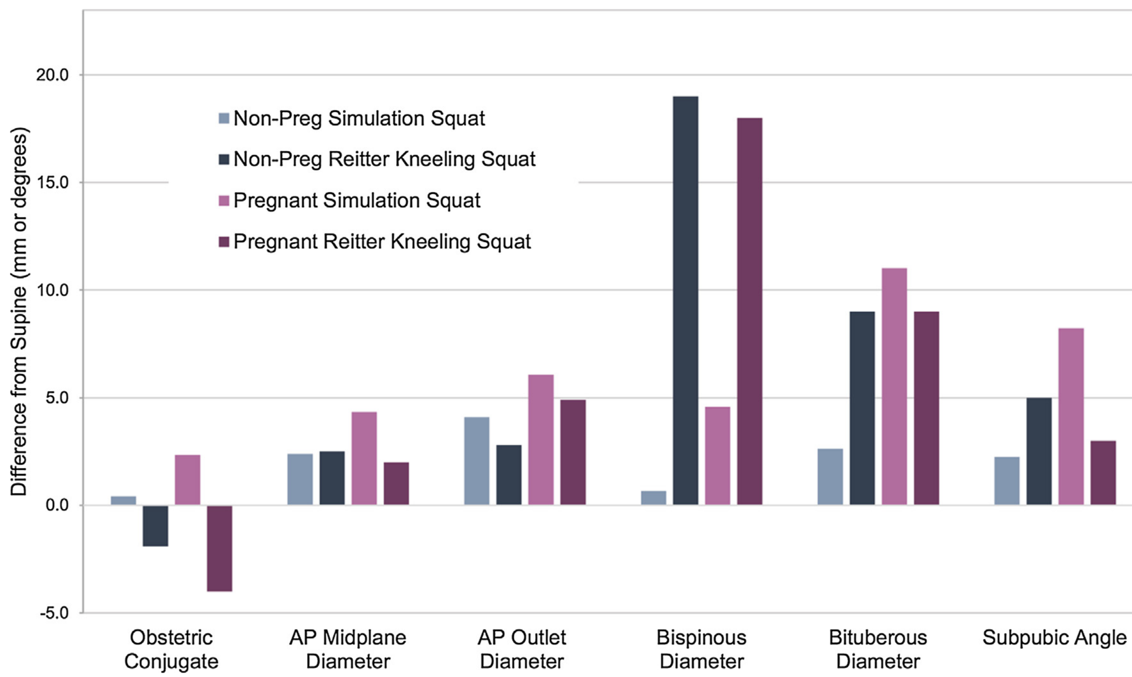


Fig. 4. Simulation results compared with data in the literature (Reitter et al., 2014), in which mean differences in measurements between similar supine and kneeling squat positions were presented for 28 to 50 subjects. All measurements are given in mm except angle in degrees.

Table 1
Pelvimetry measurements in supine for pelvic model based on *in vivo* MRI of single subject (Hemmerich et al., 2018a) and from *in vivo* data presented in the literature (Reitter et al., 2014). Literature data were presented as mean values in cm for 28 to 50 subjects and are converted to mm (with standard deviations in parentheses) for comparison here.

Supine Measurement		Pelvic Model	Literature Non-pregnant	Literature Pregnant
Anteroposterior				
	Obstetric Conjugate (Pelvic Inlet)	115.3	126.0 (11.3)	126.2 (8.0)
	AP Mid-Plane Diameter	134.5	131.7 (8.8)	134.5 (7.7)
	AP Outlet	90.7	85.9 (8.5)	86.1 (10.3)
Transverse				
	Bispinous Diameter	96.2	120 (7.6)	126 (6.5)
	Bituberous Diameter	103.7	126 (9.2)	136 (9.3)
	Anterior Angle (degrees)	71.2	70 (5)	74 (5)

gest change in transverse measurement from the minimum to maximum value for the non-pregnant simulation was 9 mm for the bituberous diameter, similar to that of the pregnant squat simulation (Fig. 6). Maximum bispinous and bituberous diameters for both pregnant and non-pregnant simulations were achieved just prior to reaching the final resting position and maintained in the simulated squat posture.

Similar to the transverse pelvic outlet measurements, both the subpubic angle and width of the inferior aspect of pubic symphysis had expanded by the end of the squatting simulation (Fig. 7). The

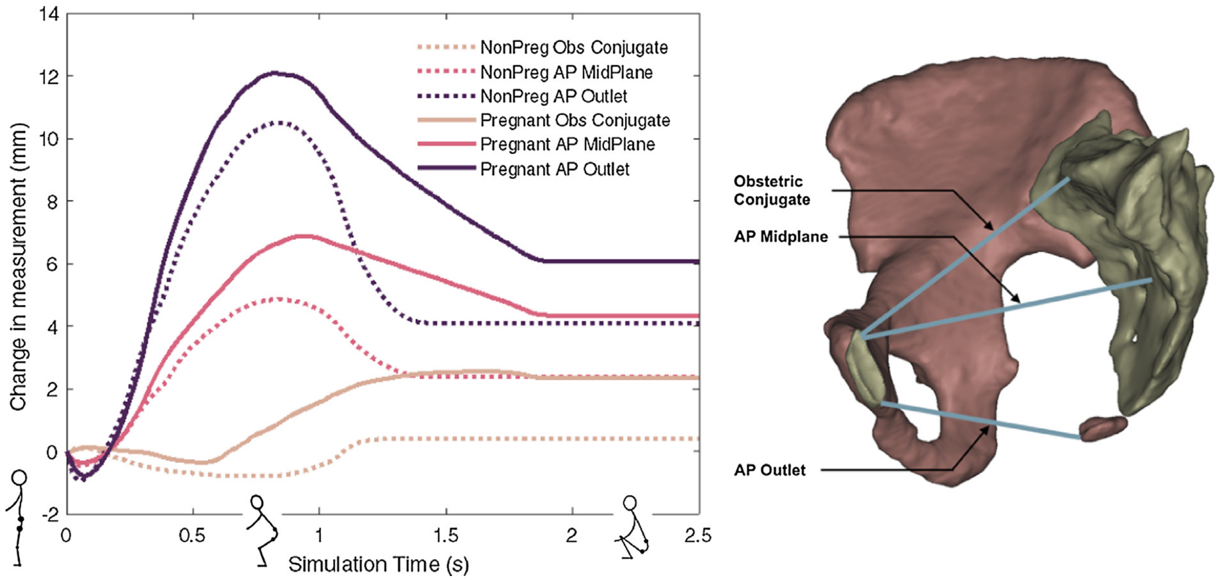


Fig. 5. Anteroposterior (AP) pelvic measurements for pregnant and non-pregnant squatting simulations. Pregnant conditions included increased ligament laxity and specific joint rotational moment inputs.

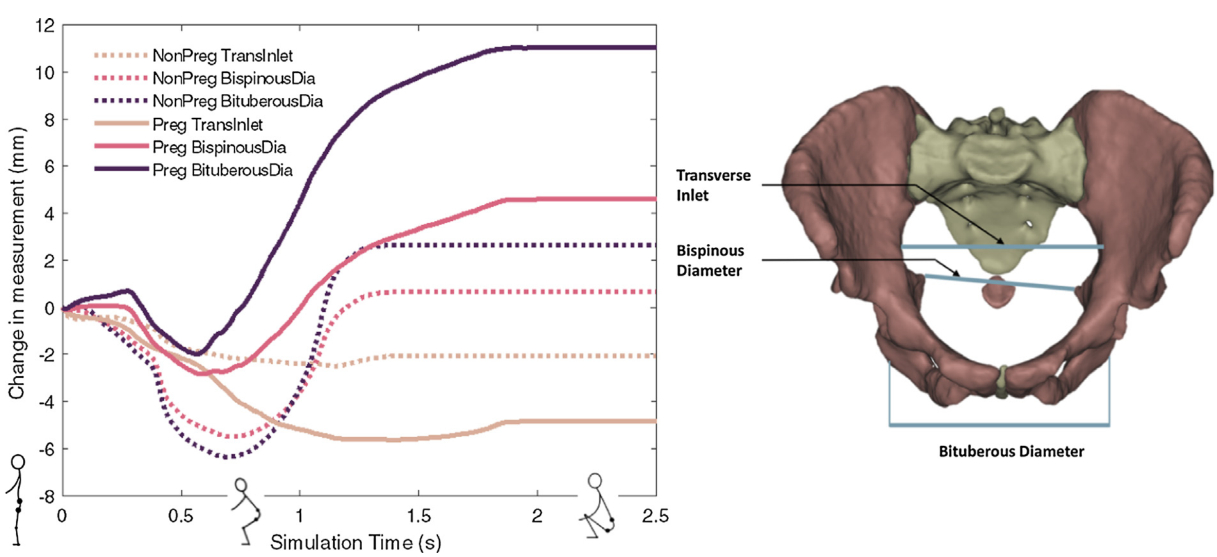


Fig. 6. Transverse pelvic measurements for pregnant and non-pregnant squatting simulations. Pregnant conditions included increased ligament laxity and specific joint rotational moment inputs.

final position increase in subpubic angle was substantially larger for the pregnant versus non-pregnant simulation: 8.2 versus 2.2 degrees, respectively (Figs. 4 and 7). However, the non-pregnant subpubic angle initially decreased to a greater extent during the simulation than the pregnant condition angle. By contrast, the final width of the superior aspect of the pubic symphysis decreased slightly in comparison to the initial position for both pregnant and non-pregnant simulations (Fig. 7).

Pelvic area measurements clearly increased in mid-plane and outlet areas for both non-pregnant and pregnant squat simulations with the pregnant squat increase being approximately twice that of the non-pregnant (Fig. 8). Conversely, pelvic inlet areas decreased slightly (less than 3%) for both non-pregnant and pregnant squat models over the course of the squatting movement.

By comparing the simulation results in which pregnant squat moments were applied without increased ligament laxity (Appendix B left side plots) with the simulation data in which only ligament laxity was increased (Appendix B right side plots), we see the individual effects of joint loading and ligament laxity on simulation outcome. The left side simulation plots, in which pregnant and non-pregnant joint moments were used to actuate the model, generally show a distinct change in shape between pregnant and non-pregnant curves. By contrast, plots on the right show a similar pattern between simulation conditions in which only the ligament stiffness was altered; however, a greater displacement occurred with the increased ligament laxity.

4. Discussion

We used a computational model to study the effects of an upright birthing position (squatting) on pelvic movement. Our model demonstrated how the loads experienced at the hip and lumbosacral joints during squatting move the pelvic bones. Although the joint loading calculated from our previous study (Hemmerich et al., 2018b) for our pregnant and matched non-

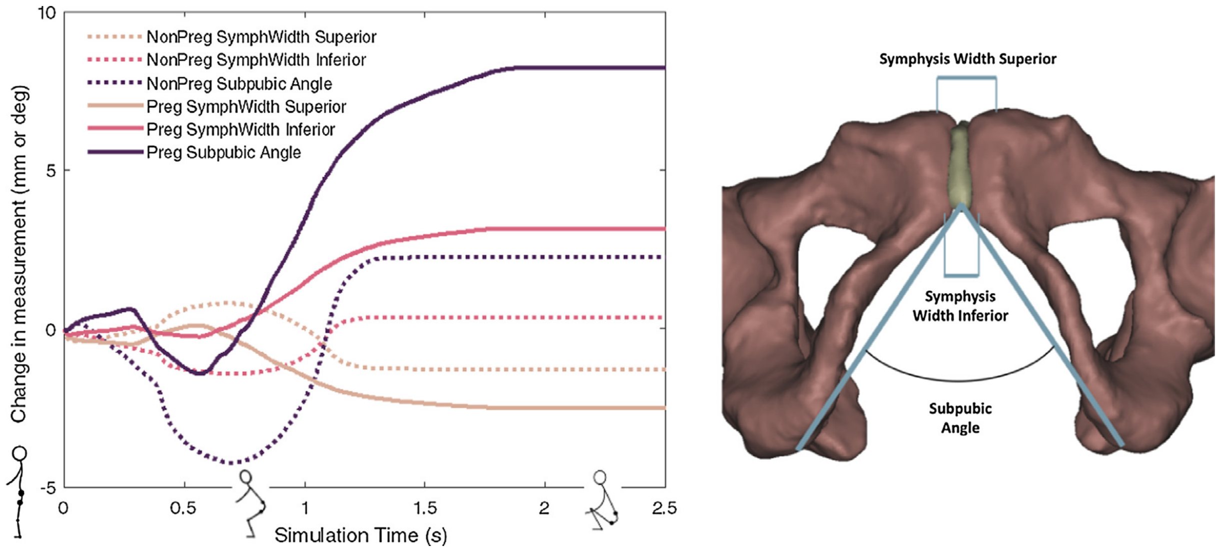
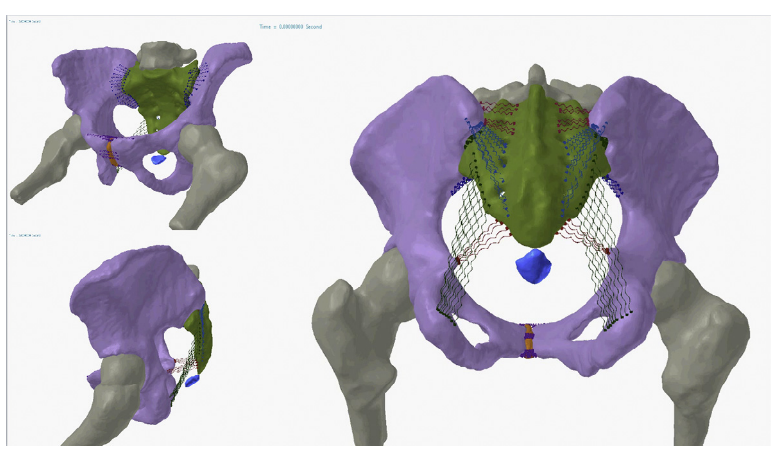


Fig. 7. Subpubic angle and pubic symphysis widths for pregnant and non-pregnant squatting simulations. Pregnant conditions included increased ligament laxity and specific joint rotational moment inputs.

pregnant participants were slightly different (Fig. 2), the final outcomes were similar: AP and transverse plane outlet measurements – and, to a lesser extent, mid-plane measurements – increased when compared with the initial pelvic position (measured in supine). The computational model allows us to see these pelvic movements in a way that is currently impossible with conventional imaging techniques; the simulation helps us to visualize how the squatting lumbosacral and hip moments (i.e. the torque generated by the muscles around the joints) cause the sacrum and innominate bones to rotate so as to open the outlet of the birth canal (Fig. 9 and Supplementary Video 1).

a substantial increase in pelvic outlet area, especially in the pregnant model, with only a slight decrease in overall pelvic inlet area (Fig. 8).

We can observe from the model that this is due to the joint positions in relation to the anatomical landmarks being measured. The lumbosacral joint, for example, is almost in line with the obstetric conjugate (Fig. 10); therefore, negligible change occurred in AP inlet diameter when the sacrum rotated (Fig. 5). By contrast, the perpendicular distance between the joint and AP outlet line is much larger, producing a greater effect on this measurement with the same degree of sacral rotation.



Supplementary Video 1. Computational model demonstrates opening of pelvic outlet during squatting movement for pregnant simulation condition.

By measuring the movement of specific anatomical landmarks on the model throughout the simulation, we were able to calculate clinical diameters in a three-dimensional environment. Rotation and translation of individual model components expanded not just the AP and transverse outlet diameters, but also the subpubic angle and the inferior aspect of the pubic symphysis width. Pelvic area, while not included in traditional pelvimetry due to the additional complexity required to calculate it, provided a better understanding of the integrated AP and transverse effects on pelvic alignment. Again, our simulation results showed

Similarly, we can consider the hip joints – about which the innominate bones rotate – which are positioned only slightly caudally relative to the pelvic inlet defined by the pelvic brim and a similar distance cranially to the bispinous diameter (Fig. 10). As a result, the decrease in transverse inlet diameter was similar to the increase in bispinous (mid-plane) diameter (Fig. 6) during both pregnant and non-pregnant simulations. The line joining the ischial tuberosities is substantially further from the hip joints, thereby increasing the bituberous diameter (outlet) to a greater extent for the same angle of rotation. The overall effect was greater

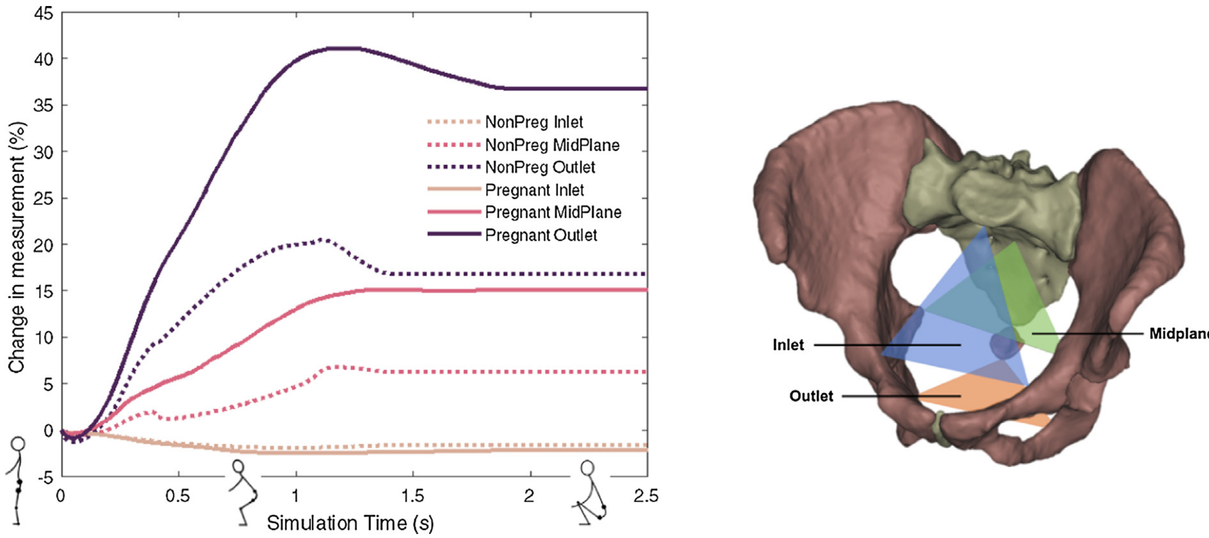


Fig. 8. Pelvic area measurements for pregnant and non-pregnant squatting simulations. Pregnant conditions included increased ligament laxity and specific joint rotational moment inputs.

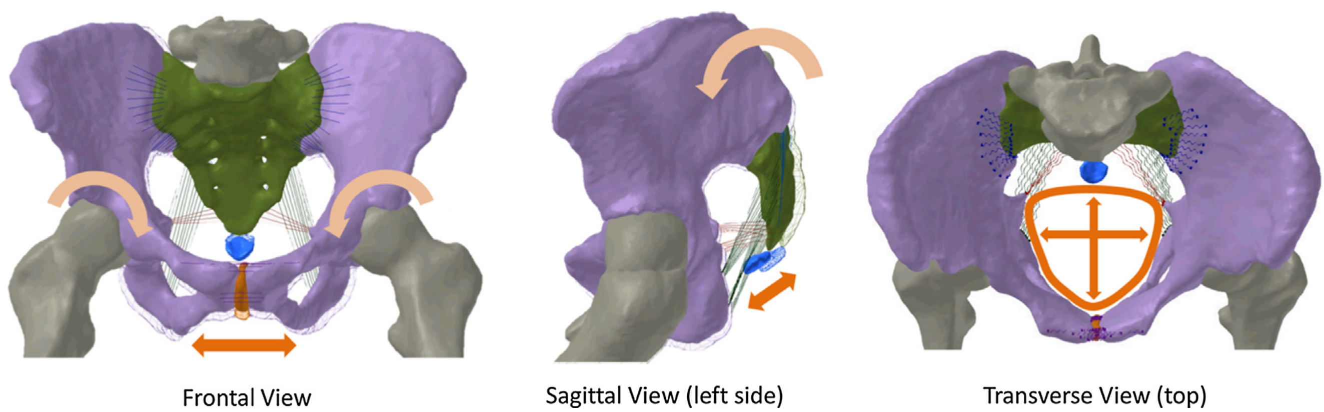


Fig. 9. Computational model is shown in the initial position (solid) and in squatting (wireframe) during the pregnant condition simulation. Curved and straight arrows indicate joint moments (L5/S1 and hips) and resulting bone movement, respectively. The wireframe model demonstrates an opening of pelvic outlet in anterior-posterior and transverse directions.

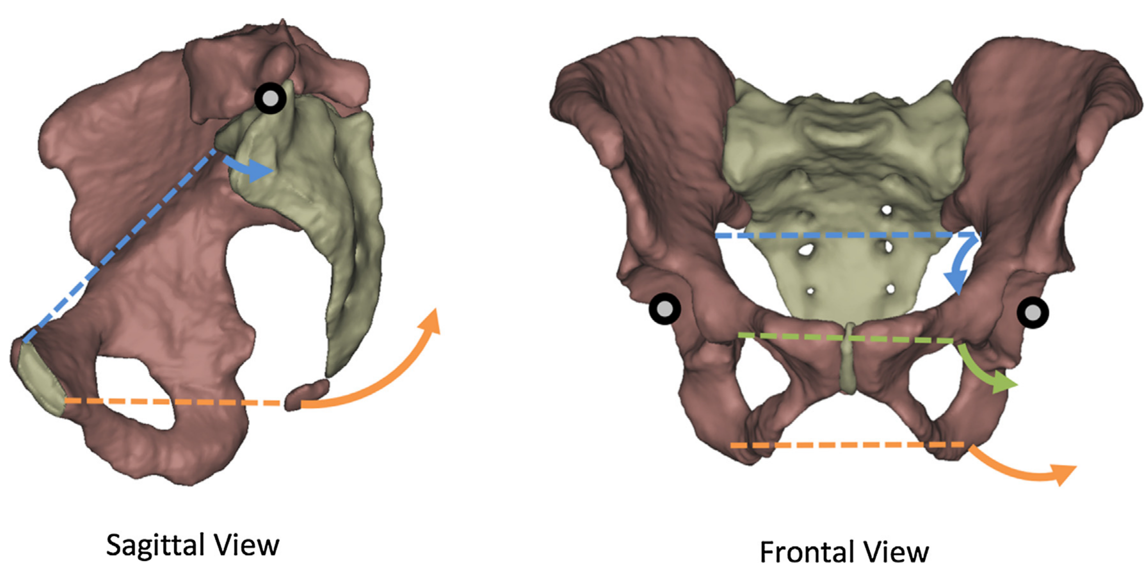


Fig. 10. Sagittal and frontal plane views of pelvis showing joint positions relative to inlet, midplane, and outlet dimensions. The pelvis model is used to indicate the perpendicular distance between the joint of interest and the line connecting anatomical landmarks used to measure clinical diameters. Left: The perpendicular distance from the L5/S1 joint (black-grey) to the AP outlet line (orange) is much greater than the obstetric conjugate (inlet shown in blue). Right: The perpendicular distance from the hip joint (black-grey) to the transverse inlet (blue) and bispinous diameter (green) is similar; the distance to the bituberous diameter (orange) is greater. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

expansion of the subpubic angle (Fig. 7) and pelvic outlet (Fig. 8) under these loading conditions than contraction of the pelvic inlet.

Another benefit to the computational model was that it permitted a dynamic analysis of pelvic motion *throughout* the simulated squatting activity and demonstrated substantial movement beyond what occurs in the final resting position. Most measurements, in particular in the AP direction, showed peak increases in diameter during the dynamic portion of the movement, rather than the final resting position. Furthermore, the peaks in pelvic area roughly correspond with the time of maximum joint loading during the movement. This peak in joint loading is, in fact, required to reduce the downward velocity of the body before coming to rest (Hemmerich et al., 2018b). In other words, the comparatively higher joint loading that is experienced during dynamic activities generates greater pelvic mobility than the comparable static posture. This observation supports clinical recommendations for the parturient person's freedom of movement during labor ("Joint Policy Statement on Normal Childbirth," 2008; Romano and Lothian, 2008) whereby mobility and changing positions may help the laboring person's pelvis move and change shape to facilitate the progression of labor.

As anticipated, the effects of squatting on pelvic movement are more pronounced in pregnancy than non-pregnancy according to the model results. This may be attributed to three biomechanical differences: the magnitude of the joint loads as a result of the increase in body mass during pregnancy, the difference in joint loading pattern as a result of changes in squatting style in our pregnant subject, and the increased ligament laxity during pregnancy. By using a computational model, we were able to simulate the effects of pregnancy related changes separately so that we could understand which differences in simulation outcomes were a result of the differences in squatting joint moments that were applied to the model and which were due to the increase in ligament laxity.

From the data in Appendix B (pregnant squat moments without increased ligament laxity simulation) we can discern that the difference in hip and lumbosacral joint *loading* experienced by pregnant and non-pregnant subjects (Fig. 2) primarily accounts for the change in movement pattern – i.e. curve *shape* – of the pelvic bones and associated clinical diameters. By contrast, the effect of increasing *ligament laxity* predominantly increased the *magnitude* of the bone movement without affecting the direction of movement (Appendix Fig. A.1 through Appendix Fig. A.4). It should be noted that the simulated increase in ligament laxity during pregnancy is a first approximation based on an extrapolation of published data (Charlton et al., 2001). Our simulation data are intended to show the trend in bone movement that would be expected if pelvic ligament laxity increases during pregnancy; however, sensitivity analyses are recommended for future studies.

This change in movement pattern as a result of variation in joint loading is evident when analysing the anteroposterior and transverse diameters (Appendix Fig. A.1 and Appendix Fig. A.2). The initial increase in AP outlet diameter for the non-pregnant simulation is due to the lumbosacral extension moment that pulls the sacrum outward; the opposing "reaction" force on the model then draws the innominate bones inward, decreasing bispinous and bituberous diameters, before the hip frontal plane moment is great enough to overcome this force and pull the pubic bones outward again.

The peak hip frontal plane moment for the pregnant simulation was not only twice as large as for the non-pregnant subject (44 versus 21 Nm), but also represented a greater share of the overall applied loads since the sagittal plane moments – both hip and lumbosacral – were similar in magnitude between pregnant and non-pregnant subjects (Fig. 2). Consequently, the pregnant hip frontal plane moments were able to draw the caudal aspect of the innominate bones out, thereby increasing the transverse outlet diameters almost immediately during the pregnant squat simulation (Fig. 6).

Why, then, did the pregnant subjects demonstrate greater hip frontal plane moments than the non-pregnant? Further biomechanical investigation is required to confirm our theory; however, the pregnant subjects adopted a wider stance in squatting, presumably to allow their knees to get around the belly while maintaining a stable posture. The larger the distance between the hip and the ground reaction force at the foot, the larger the hip joint moment required to offset this force (i.e. widening one's squatting stance results in increasing these joint moments) (Hemmerich et al., 2018b).

Clinicians and researchers alike have suggested that the effect of gravity in an upright birthing position, such as squatting, contributes to the progression of labor (Desseauve et al., 2017). The implication is that gravity acting on the baby is now providing a force in the direction of movement – out the birth canal – rather than, for example, towards the labouring person's spine as would be the case in supine. Our computational analyses reveal that it is gravity acting on the body of *the laboring person* (having a much greater mass than that of the baby and, consequently, more substantial force) that affects the alignment of the pelvic bones. Such considerable gravitational forces could potentially influence childbirth mechanics far more than those on the "descending passenger."

Another advantage that is often put forward regarding upright compared with recumbent birthing positions is that the sacrum and coccyx are unencumbered; in other words, the force of the bed under the sacrum in supine would close the pelvis, while in squatting the sacrum and coccyx would be free to move. Again, our analyses indicate that there is, in fact, an additional biomechanical element in which the squat posture actually generates joint loading at the lumbosacral joint that would pull the sacrum outwards, rather than simply removing an obstruction to free motion.

In addition to physiological vaginal delivery, clinical implications for upright positioning would include facilitating delivery in cases where additional maneuvers can be indicated, such as breech or shoulder dystocia. Many clinical maneuvers, including McRoberts', Woods' Screw, and Rubin, are performed with the laboring person lying on their back. However, it is possible that a recumbent position may diminish mechanical advantages of being upright and assisting a parturient person who is in an upright position may be more effective. This may be especially relevant in cultures where upright birth positioning is common or access to technological resources is limited.

While a computational analysis of the effects of squatting on pelvic dimensions provided numerous advantages, one must consider the simplifications and assumptions included in the model. We previously validated the computational simulation using MRI in non-pregnant subjects. However, since (in Canada) MRI during pregnancy is restricted to indications considered to be clinically necessary, our second author advised against a similar validation of the simulation of pregnant conditions.

That said, simulation results were reasonably similar to data presented by Reitter et al. (2014) for pregnant people in a kneeling squat position (Fig. 4). The only discrepancy was in the direction of change in measurement for the obstetric conjugate where literature data showed a mean decrease in diameter in the kneeling squat position. However, our slight increase in the measurement in the pregnant squat simulation corresponded with the MRI data for the non-pregnant subjects (Hemmerich et al., 2018b); since the model was generated from the MRI scans of one of those participants it is possible that this was simply a subject-specific difference rather than a flaw in the model.

This point then illustrates another limitation of this model: it is *not* representative of all pregnant and non-pregnant people. As a subject-specific model, both with respect to the bone geometry (discerned from MRI) and applied joint loads during squatting, the simulation provided an estimate of what is experienced by the person for whom the data was modeled. People having different pelvic shapes or whose squatting motion does not resemble what was simulated in this study may exhibit different pelvic bone motion. Recommended future work would be to investigate the effect of pelvic shape on bone kinematics using a computational model and compare this variation with *in vivo* data.

When applied to the computational model, hip and lumbosacral joint moments experienced during squatting increased most pelvic measurements and primarily opened the pelvic outlet in the AP and transverse directions. It is important to understand that the loading on the pelvis occurs in a three-dimensional configuration. Muscles around the hip and lumbar spine generate substantial joint moments in both the sagittal and frontal planes in order to cooperatively open the outlet of the pelvis in squatting. Our simulations, furthermore, demonstrated considerable increases to bone motion during the dynamic movement, supporting clinical observation with regards to the benefits of mobility during labor. Unsurprisingly, the extra ligament laxity that is experienced during pregnancy resulted in a further increase in bone motion within the simulation environment.

Outcomes from our computational simulation suggest that advantageous maternal joint loading in an upright birthing position such as squatting could open the outlet of the birth canal and potentially facilitate delivery.

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Conflict of interest

None declared.

Appendix A. Estimation of ligament stiffness during pregnancy based on data provided by (Charlton et al., 2001)

Table A1 lists the ligament stiffness data from which the ratio of pregnant to non-pregnant stiffness was calculated.

Based on the equation

$$F = k\Delta x$$

where F is force in lbs, k is stiffness, and Δx is translation in mm, we can use the data provided in the table above to solve for the unknown, k, for both third trimester and postpartum data.

In other words, using the third trimester data, since

$$20 = 4.7k$$

and

$$40 = 7.2k$$

then

$$k = 8\frac{lbs}{mm}$$

Similarly, for the postpartum data

$$k = 13.3\frac{lbs}{mm}$$

Therefore, the ratio of ligament stiffness during pregnancy to postpartum is

$$\frac{k_{pregnancy}}{k_{postpartum}} = 0.6$$

Table A1
Ligament stiffness data provided by (Charlton et al., 2001) showing the average values for translation (mm) at 20 lbs and 40 lbs displacement tests from the third trimester of pregnancy to the postpartum.

	20 lbs	40 lbs
Third trimester	4.7	7.2
Postpartum	3.3	4.8

Appendix B

(See Figs. A1–A4).

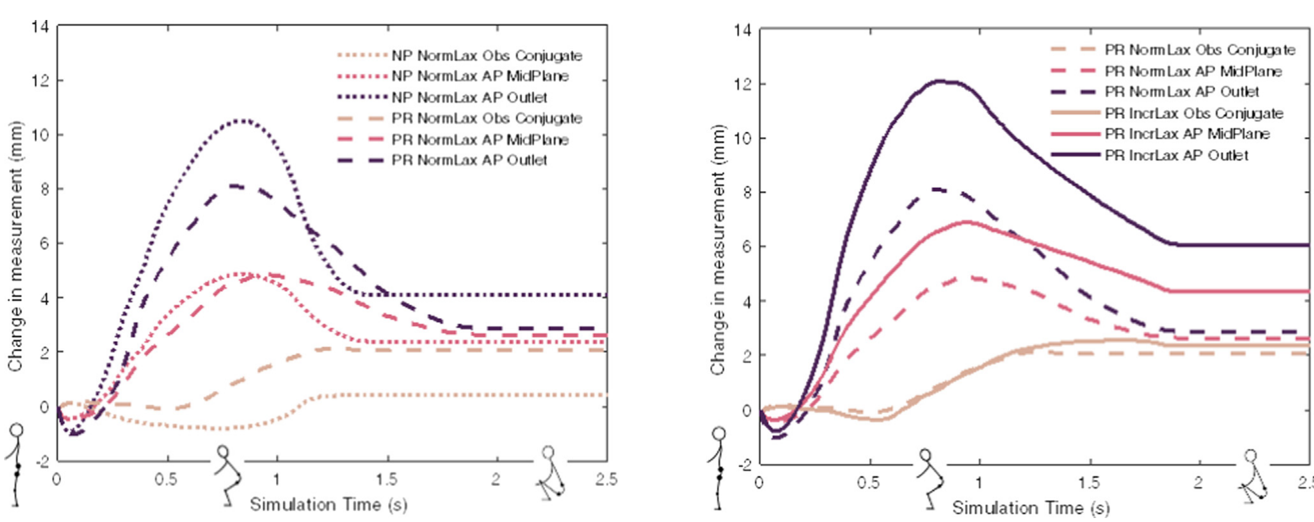

Fig. A1. Anteroposterior (AP) pelvic measurements for pregnant (PR) joint loading and normal ligament laxity condition compared with non-pregnant (NP) simulation [left] and pregnant simulation with increased ligament laxity [right].

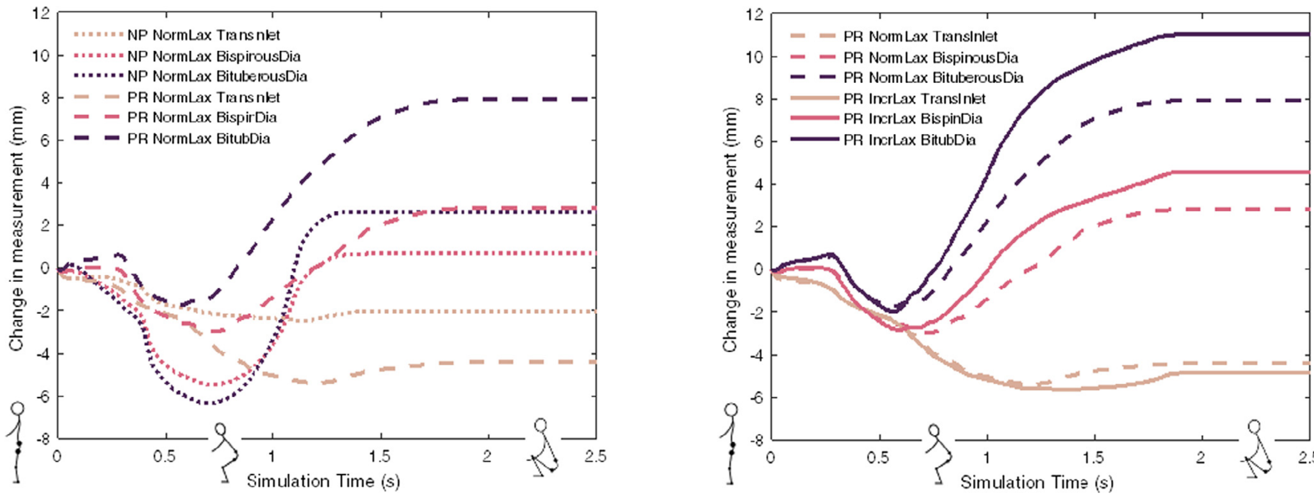

Fig. A2. Transverse pelvic measurements for pregnant (PR) joint loading and normal ligament laxity condition compared with non-pregnant (NP) simulation [left] and pregnant simulation with increased ligament laxity [right].

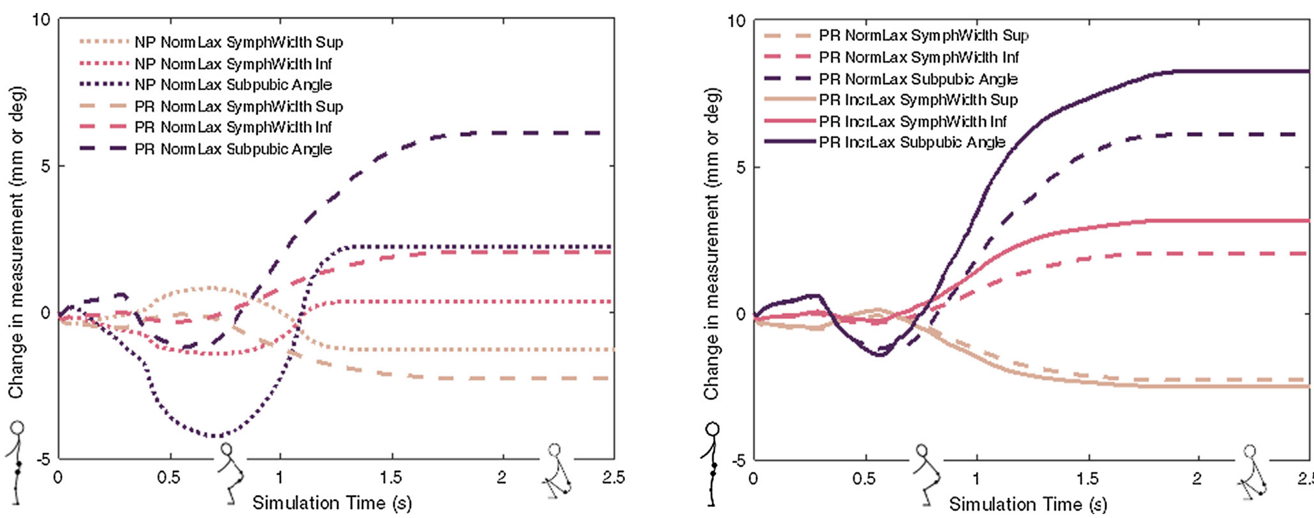

Fig. A3. Pubic symphysis width and subpubic angle for pregnant (PR) joint loading and normal ligament laxity condition compared with non-pregnant (NP) simulation [left] and pregnant simulation with increased ligament laxity [right].

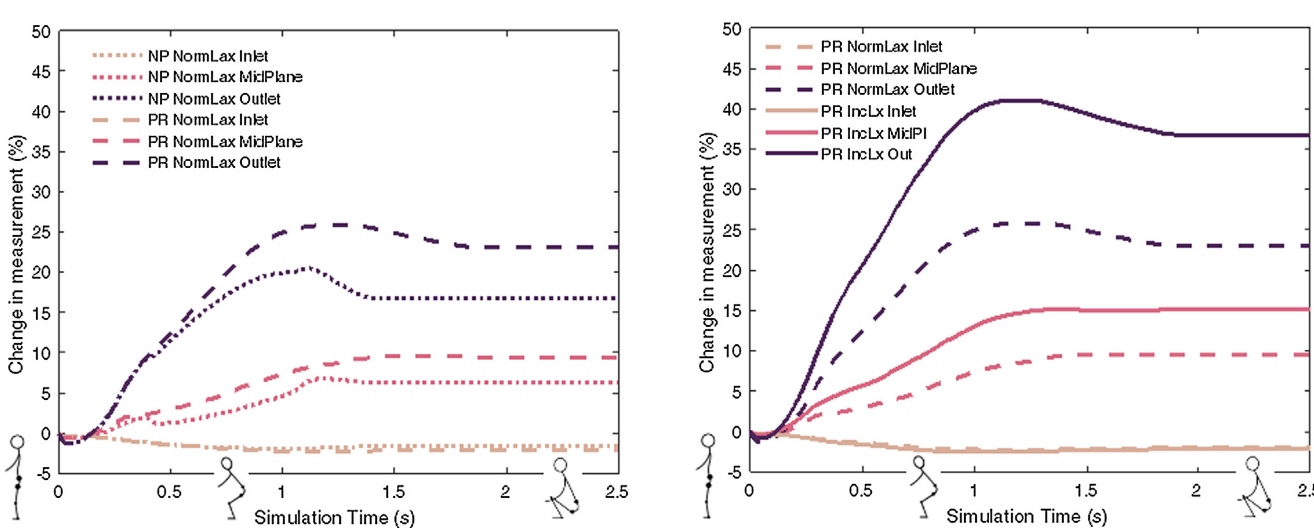

Fig. A4. Pelvic area for pregnant (PR) joint loading and normal ligament laxity condition compared with non-pregnant (NP) simulation [left] and pregnant simulation with increased ligament laxity [right].

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