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Original Article

Changes in segmental mass and inertia during pregnancy: A musculoskeletal model of the pregnant woman

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ABSTRACT

Background: One in four pregnant women falls at least once during her pregnancy. During pregnancy, the body undergoes tremendous vascular, hormonal, physiological, and psychological changes to accommodate the growing fetus. The pregnancy-induced mass gain of 10 to 25 kg is not evenly distributed and results in a large change in mass distribution and shift in segmental centers of mass. To accurately understand how the change in mass distribution leads to an increase in fall events, a musculoskeletal model of the pregnant body is necessary. Generic musculoskeletal models cannot accurately represent the morphology of pregnant women and the study of postural stability of pregnant women is limited by the lack of adapted musculoskeletal models.

Research question: Could a model reflecting the change in segmental inertia during pregnancy explain the pregnancy-related risk of falling?

Methods: We built a musculoskeletal model of the pregnant women, combining literature anthropomorphic measurements with generic models. We optimized the dimensions of the anthropomorphic model shapes to fit the average measurements of 25 pregnant women. The mass, center of mass, and inertia of each segment are then computed throughout pregnancy. Finally, the stance phase of a gait cycle was modeled using the pregnancy-specific and the generic models. The ankle, knee, hip and lumbar joint moments during gait were compared between the two models.

Results: The built musculoskeletal model of the pregnant woman includes changes in mass and geometry of the thorax, pelvis, thighs, and legs. The model reproduces the change in lumbar curvature during pregnancy. Gait simulation results show a limited impact of pregnancy on the ankle, knee, and hip moment, but a large impact on the lumbar moment.

Significance: Such a musculoskeletal model will help elucidate the mechanisms leading to falls or low back pain during pregnancy.

1. Introduction

Pregnancy induces tremendous changes in the body to accommodate a growing fetus. Pregnant women gain between 10 and 25 kg with a mean mass gain at 15 kg [1]. Pregnancy is also associated with vascular, hormonal, and physiological changes. Hormonal changes notably lead to joint laxity and physiological changes to fluid retention compressing soft tissues. As a result of these changes, musculoskeletal pain can arise [2] and daily activities or work-related tasks that were simple and inconsequential when not pregnant become problematic. For example, walking, standing for long periods, ascending and descending stairs, and bending lead to an increased risk of falling [3]. To accommodate anthropomorphic changes and maintain balance, pregnant women adapt their posture and gait biomechanics [4,5]. Despite these adjustments, more than one in four women (27%) falls at least once during her pregnancy [3], and on average, 4.5% of American women of reproductive age are pregnant at any point in time [6]. Falls are rarely as costly as when they happen during pregnancy, leading to substantial healthcare, emotional, and societal costs.

The loss of stability amongst pregnant women is mainly due to increases in overall mass and changes in mass distribution, as has been shown by studies simulating pregnancies [7,8]. However, it is not well understood how the mass changes (amount and distribution) specifically affect the postural stability (during standing) and the dynamic postural stability (during gait) of pregnant women. This is partly due to the lack of a dedicated musculoskeletal model for pregnant women,

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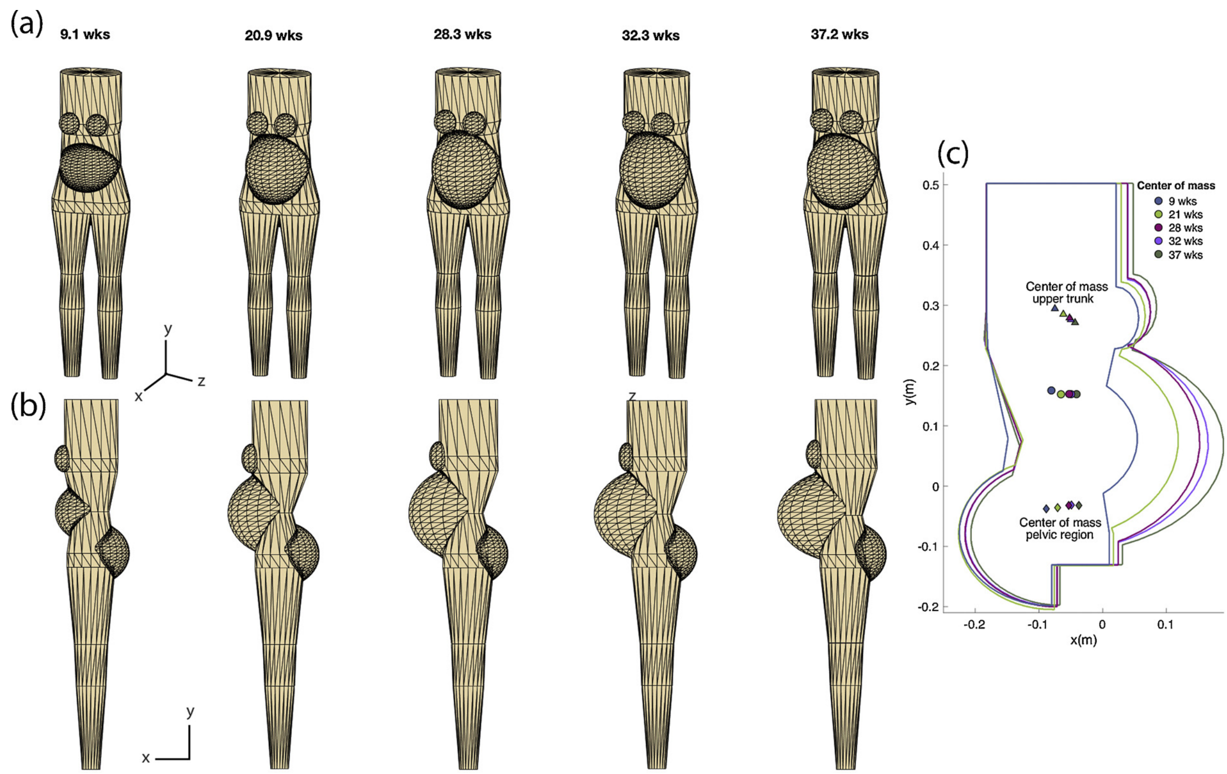

Fig. 1. Anthropomorphic model of the pregnant body (a) face (b) profile. The model shows that the dimensions measured on pregnant woman are reproduced with limited constraints on the shapes. (c) Representation of the contour of the trunk and of the positions of the upper trunk center of mass, the pelvic region center of mass, and the torso center of mass at 9, 21, 28, 32, and 37 weeks of pregnancy. As expected, the torso center of mass shifts forward and downward. In addition, the model reproduces the increase in lumbar curvature observed during pregnancy, as well as predicts a shift upward of the breast to accommodate the bump.

which allows the computation of joints' ranges of motion and moments.

Multiple studies have identified strategies developed by pregnant women to compensate for the loss of postural stability and the effort has recently intensified taking advantage of new technologies such as wireless pressure insoles measuring the center of pressure (COP) displacement [9,10]. Measurements of pregnant subjects show that the COP velocity decreases with pregnancy [11]. The stance width increases [5] while the stride length and frequency decrease [12]. The medio-lateral sway [5,9], the base of support [13], and the medio-lateral ground reaction forces [5] also increase during pregnancy. These adaptations have significant consequences for the joints moments and range of motion, leading to modifications of the COP and center of mass (COM) paths during quiet standing and gait.

A pregnancy-specific musculoskeletal model is needed because the pregnancy-induced anthropomorphic changes are significant and subject-specific, sometimes even pregnancy-specific for a single subject [14–17]. As pregnant women segmental masses and inertial parameters differ significantly from the generic human model largely used in biomechanics, a specific pregnancy model has to be developed [17,18]. Finally, using magnetic resonance imaging or X-ray imaging regularly is impractical and not recommended for pregnant women [17], and a non-invasive weighted sum of segmental centers of mass is generally chosen to build models.

Previous proposed models have not allowed the simulation of movements of the American pregnant woman body. Morino et al. [18] focused on the end of the pregnancy and on demographics with different anthropomorphic characteristics, namely Japanese women. Catena et al. [17] only provided the segmental masses and the torso center of mass. Finally, Jensen et al. [14] did not differentiate between the segments corresponding to the torso and the pelvis, thus prohibiting their relative movements in a corresponding musculoskeletal model even though this specific motion has proven relevant to understanding pregnant gait [19,20]. The goal of this study is to build a musculoskeletal model of the pregnant woman at all stages of the pregnancy that allows modeling of any movement. The usefulness of such a model to understand musculoskeletal disorders in pregnant women will be demonstrated by the simulation of the stance phase of a gait cycle.

2. Literature data

There are few sets of data providing a comprehensive view of the anthropomorphic evolution of women during pregnancy and the model has been developed based on a data set published by the USAFRL [21] after 3D scanning pregnant female subjects as part of the Accommodation and Occupational Safety for Pregnant Military Personnel project. A set of averaged anthropomorphic measurements were recorded and published. A total of 35 subjects were recruited after including civilians into the study, 25 completed the initial baseline session and at least one additional one. The data was recorded in six sessions (number of weeks ± 1 standard deviation): first trimester (9.1 ± 2.8 weeks on average), second trimester (20.9 ± 1.4 weeks), 3 sessions in the third trimester (28.3 ± 0.9, 32.3 ± 0.8, 37.2 ± 0.6 weeks), and post delivery (3.7 ± 1.7 weeks postpartum). To build the model, we used the measurements of the trunk sitting height, the cervical height and sitting height, the circumference, breadth, depth, and height of the chest, the circumference, breadth and height of the chest below the bust, the circumference, breadth, depth, and height of the waist, the breadth and circumference of the hips, the circumference of the thighs measured at the lowest point of the thigh buttock juncture, the height of the knee, the maximum circumference of the calf, and the circumference of the ankle during pregnancy.

3. Method

3.1. Musculoskeletal model development

3.1.1. Modeling the volume change

Each body segment was modeled as an elementary shape and added to the standing body model (Fig. 1a and b). The upper trunk, lower trunk, and pelvis were modeled as extruded obround, the breast and gluteal region as ellipsoids, the thighs and lower legs as double frustums. An obround is a stadium shape and a frustum is a truncated cone. The pregnant abdomen was modeled as an ellipsoid centered on the umbilicus. The head, neck, arms, hands, and feet were directly scaled from a generic musculoskeletal model as very little change in mass was measured during pregnancy [21].

To compensate for missing data, the thigh height was assumed to be the cervical height minus the cervical sitting height minus the knee height. The knee radius was estimated at 1.9 times the ankle radius. The height of the maximum calf circumference as measured in the study was assumed at 43.52% of the knee height following [22]. The height of the pelvis/crotch was estimated at 75%/25% of the known height between the umbilicus line and the top of the thighs.

3.1.2. Breast dimensions

The breast was modeled as two ellipsoids located on the chest coronal plane. Fig. 2a indicates the unknown (a, b, d, and r) and known (*breadth*, *depth*) quantities. The dimensions were optimized according to Eqs. (1)–(6), using a Levenberg-Marquardt algorithm implemented in MATLAB® (Mathworks, Natick, MA).

$$D_{\text{model}} = D_{\text{chest}} \tag{1}$$

$$C_{\text{model}} = C_{\text{chest}} \tag{2}$$

$$B_{\text{model}} = B_{\text{chest}} \tag{3}$$

$$B_{\text{breast}} = 0.8B_{\text{chest}} \tag{4}$$

$$2r = 1.05D_{\text{Chest belowbust}} \tag{5}$$

$$d = 0.20B_{\text{Chest}} \tag{6}$$

where B, C, and D stand for breadth, circumference, and depth, respectively. Eq. (4) indicates the breast should occupy about 80% of the chest breadth. Eq. (5) allows the depth of the obround of the trunk (equals to $2r$) to be 5% larger than the depth measured below the bust. Finally, the breast was assumed symmetrical and the distance between the center of each ellipsoid and the medial plane, d, was estimated at 20% of the chest breadth. The third axis of the ellipsoid was determined by the difference between the chest height and the chest below bust height.

3.1.3. Gravid abdomen and gluteus

The gluteal region was modeled as one ellipsoid in the pelvis/crotch coronal plane. The increased abdominal girth was modeled as an ellipsoid centered on the umbilicus coronal plane. Fig. 2b represents the geometry in the coronal plane, where r and l are the radius and length of the obround representing the trunk and b is the minor axis of the ellipsoid representing the buttocks. The ellipsoid was located on the trunk so that the major axis of the ellipsoid was $a = r + l/2$.

The dimensions of the gluteal region were determined by optimization (Eqs. (7)–(9)), using the Levenberg-Marquardt optimization algorithm.

$$B_{\text{hip}} = 2r + l \tag{7}$$

$$R_{\text{thigh}} = r \tag{8}$$

$$C_{\text{hip}} = C_{\text{model}} \tag{9}$$

where B, C, R stand for breadth, circumference, and radius, respectively. The third axis of the ellipsoid was estimated at $1.2b$. The circumference of the model (Eq. (9)) took into account the presence of the anterior abdomen as pregnancy progresses.

Similarly to the gluteus, the dimensions of the gravid abdomen were determined by optimization (Eqs. (10)–(12)).

$$B_{\text{waist}} = 2r + l \tag{10}$$

$$C_{\text{waist}} = C_{\text{model}} \tag{11}$$

$$D_{\text{waist}} = r + b \tag{12}$$

where B, C, D stand for breadth, circumference, and depth, respectively. The dimensions of the abdomen were limited in the mid-sagittal plane to reaching the breast and the lower base of the pelvis.

3.1.4. Segmental center of mass and inertia

The inertia of each segment was numerically computed. Each segment was defined as a triangular surface mesh. The inertia was determined in MATLAB [23] by transforming the triple integral over the volume into a double integral over the surface using the divergence theorem. As the surface was defined by a triangular mesh, the surface was piecewise linear and the integral over the surface was written as a sum of the moments over each triangle, which was solved in barycentric coordinates. This method allowed us to compute the inertia of any closed shape, regardless of its complexity.

Since the density in each segment is constant, the position of the center of mass was directly determined from the geometry (Fig. 1). The influence of the fetus, placenta, and amniotic fluid is neglected because they do not significantly modify the density of the trunk compared to a cadaver density [14]. The following densities have been used: thorax 0.92 g cm^{-3} [24], pelvis 1.01 g cm^{-3} [24], thigh 1.04 g cm^{-3} [25], knee 1.10 g cm^{-3} (from hand in [25]), calf 1.08 g cm^{-3} [25].

For ease of utilization, the coordinates of the center of mass are provided in Table 1 in each segment's coordinate system, as defined in OpenSim [26,27].

3.2. Evaluation of joint moments

The effects of pregnancy-specific inertial parameters on net joint

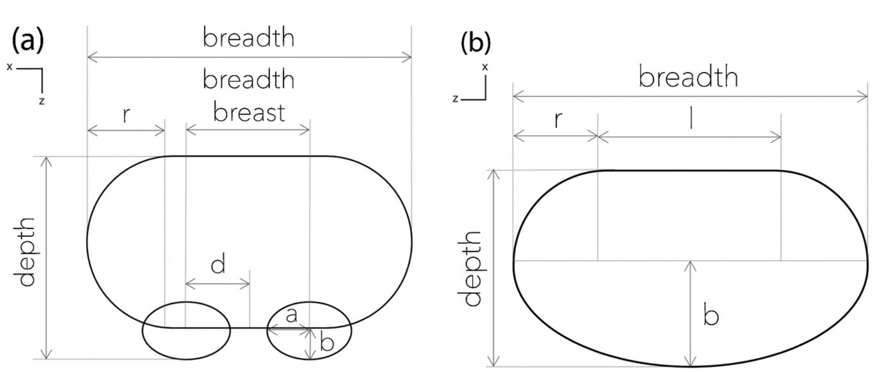

Fig. 2. (a) Chest and breast model in the transverse plane. The parameters defining the shape of the cross-section of the body at breast height are described. Some parameters, such as d and *breadth breast*, are assumed equal to 20% and 80% of the chest breadth, respectively. All other parameters are optimized to fit the known measurements. (b) Gluteal region and bump model in the transverse plane. The cross-section of the body at gluteal region height and umbilical height are modeled with identical shapes. Parameters are optimized on the measured breadth and depth.

Table 1
Mass, center of mass position, and inertia of the segments of the musculoskeletal model varying through pregnancy. X represents the antero-posterior, Y the vertical, and Z the medio-lateral direction. For each segment of the model that shows a pregnancy-induced change, the mass, center of mass position, and inertia are given, thus entirely describing a musculoskeletal model at each stage of the pregnancy. Segments constant through pregnancy are described in Table 2.

	Pregnancy (weeks)	9.1	20.9	28.3	32.3	37.2
Thorax	Mass (kg)	18.99	21.62	24.24	25.23	26.87
	Center of mass X (m)	0.0257	0.0156	0.0277	0.0293	0.0389
	Center of mass Y (m)	0.2127	0.2061	0.2023	0.2016	0.2019
	Center of mass Z (m)	0.0000	0.0000	0.0000	0.0000	0.0000
	Inertia XX (kg m^2)	0.3548	0.4243	0.4938	0.5177	0.5741
	Inertia YY (kg m^2)	0.1424	0.1883	0.2390	0.2557	0.2950
	Inertia ZZ (kg m^2)	0.3193	0.3982	0.4770	0.5047	0.5727
Pelvis	Mass (kg)	13.27	15.70	17.43	17.95	18.73
	Center of mass X (m)	−0.0885	−0.0941	−0.0742	−0.0705	−0.0558
	Center of mass Y (m)	−0.0377	−0.0331	−0.0271	−0.0242	−0.0196
	Center of mass Z (m)	0.0000	0.0000	0.0000	0.0000	0.0000
	Inertia XX (kg m^2)	0.1361	0.1598	0.17592	0.1758	0.1812
	Inertia YY (kg m^2)	0.1306	0.1778	0.2132	0.2233	0.2432
	Inertia ZZ (kg m^2)	0.0907	0.1313	0.1535	0.1617	0.1732
Thigh	Mass (kg)	5.06	5.23	5.44	5.45	5.57
	Center of mass X (m)	−0.0097	−0.0090	−0.0089	−0.0088	−0.0087
	Center of mass Y (m)	−0.1948	−0.1957	−0.1935	−0.1933	−0.1941
	Center of mass Z (m)	0.0072	0.0097	0.0125	0.0122	0.0137
	Inertia XX (kg m^2)	0.0492	0.0498	0.0528	0.0529	0.0550
	Inertia YY (kg m^2)	0.0135	0.0146	0.0161	0.0160	0.0170
	Inertia ZZ (kg m^2)	0.0492	0.0498	0.0528	0.0529	0.0550
Leg	Mass (kg)	3.87	3.93	4.10	4.09	4.30
	Center of mass X (m)	−0.0058	−0.0050	−0.0049	−0.0048	−0.0048
	Center of mass Y (m)	−0.1825	−0.1824	−0.1818	−0.1819	−0.1819
	Center of mass Z (m)	0.0072	0.0097	0.0125	0.0122	0.0137
	Inertia XX (kg m^2)	0.0452	0.0459	0.0475	0.0475	0.0500
	Inertia YY (kg m^2)	0.0039	0.0040	0.0044	0.0044	0.0047
	Inertia ZZ (kg m^2)	0.0452	0.0459	0.0475	0.0475	0.0500

moments during the stance phase of a gait cycle were estimated. We defined in OpenSim five generic pregnancy-specific models, corresponding to 9, 21, 28, 32, and 37 weeks of pregnancy and five generic model corresponding to non-pregnant woman of equal weight. A model is qualified of generic if it can be scaled to represent any element of the population fitting its basic characteristics. For example, a generic pregnancy-specific model is a model that can be scaled to fit any pregnant woman, while a generic model will fit any woman. The objective of this study is to show that the generic model cannot reproduce pregnant woman kinetics because of the changes of segmental inertia occurring during pregnancy.

We selected the kinematics corresponding to a stance phase of an OpenSim 4.0 data set (gait2354, [28]). Each model was then scaled using the scaling factors provided in the data set. Ground reaction forces corresponding to the stance phase were scaled based on each model's mass, but the kinematics were kept identical. Finally, we performed inverse dynamics to compute the net joint moments at the ankle, knee, hip, and lumbar joint for each model. The root mean square (RMS) errors for the lower-limb joint moments quantified the difference between pregnancy-specific and generic models for each pregnancy stage.

4. Results and discussion

4.1. Segmental mass validation

The segmental masses of the model through pregnancy (Table 1) were compared to previous measurements of segment mass and mass distribution [14] (Fig. 3a).

Compared to [14], the model slightly overestimated the mass for all segments except the thigh. The mean relative error between the model and [14] was 13.2 ± 2.7% for the lower trunk, 8.6 ± 2.6% for the upper trunk, 13.6 ± 2.2% for the thigh, and 28.9 ± 2.1% for the lower leg. The lower legs were especially overestimated in the model while the upper leg value was close. We attributed this to the fact that the dimensions of the legs in the model have been estimated and part of the upper leg may have been attributed to the lower leg.

The slope of each line is indicated in Fig. 3a to show the ability of the model to predict the segmental mass gain. As the order of magnitude of the slope is similar, the shape and density assumed for each segment were appropriate to simulate a pregnancy-induced mass gain. Notably, the shape of the abdomen as deduced from circumferential measurements at the umbilicus was close to what has been measured in the past [14]. However, [14] did not differentiate between the lower torso and the pelvis, rendering this data unsuitable to build a musculoskeletal model [17].

The total mass gain predicted by the model was also compared to the total mass gain independently measured in the study [21], and to the average change in BMI during pregnancy from a study specifically focused on mass gain and BMI increase during pregnancy [15] (Fig. 3b). The initial study [21] and the model are lying very close to the average change in BMI [15] (below 3% relative error). This result shows a very good agreement in the total mass change predicted by the model and measurements performed on pregnant women.

4.2. Musculoskeletal model

The musculoskeletal model of a pregnant woman has been built from a generic musculoskeletal model [29]. The generic model included 12 segments and 29 degrees of freedom (dofs). Each lower extremity had five degrees-of-freedom; the hip was modeled as a ball-and-socket joint (3 dofs), the knee was modeled as a custom joint (1 dof), and the ankle was modeled as a revolute joint (1 dof). Lumbar motion was modeled as a ball-and-socket joint (3 dofs) and the pelvis (6 dofs). Each

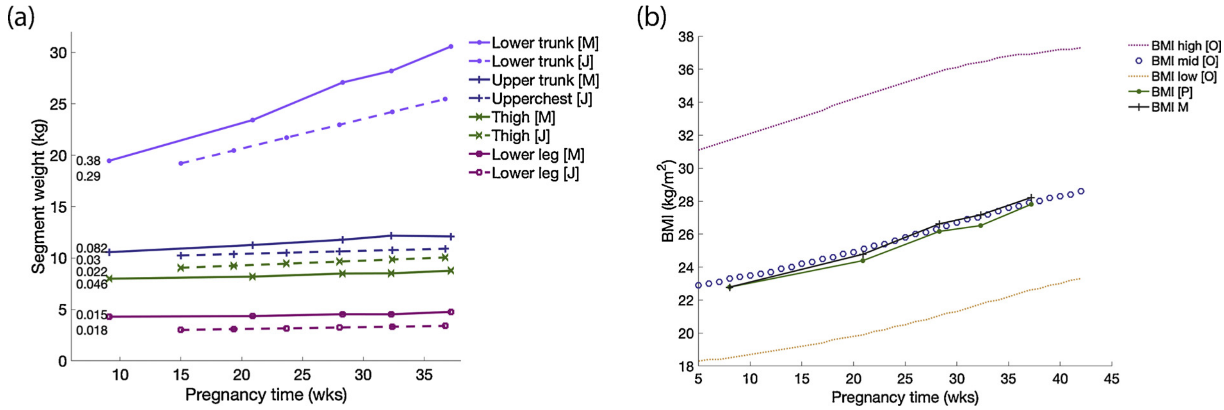

Fig. 3. (a) Comparison of segmental mass from the model [M] and from [14] [J]. Differences in absolute values may correspond to difference in tested populations. The model has been built on anthropomorphic measurements of 25 pregnant women with BMI below 26, [14] built their model on measurements of 15 pregnant women with no BMI criterion but a mean initial BMI of 24.7. The slope of the linear regression is indicated on the left of each curve. Orders of magnitude of the slopes are all similar between the two models. (b) Comparison of resulting total BMI from the model [M], [21] [P], and [15] [O]. The BMI and increase in BMI predicted by the model is close to the mean BMI observed by [15] and the BMI actually measured by the initial study [21]. The maximum relative error between the model and [15] is 2.2%.

Table 2
Mass, center of mass position, and inertia of the segments of the musculoskeletal model constant through pregnancy. X represents the antero-posterior, Y the vertical, and Z the medio-lateral direction. These values are identical to [29].

Segment	Mass	Center of mass			Inertia		
	(kg)	X (m)	Y (m)	Z (m)	I_{XX} (kg m^{-2})	I_{YY} (kg m^{-2})	I_{ZZ} (kg m^{-2})
Radius/ulna	0.61	0	−0.1205	0	0.0030	0.0006	0.0032
Hand	0.46	0	−0.0681	0	0.0009	0.0005	0.0013
Humerus	2.03	0	−0.1645	0	0.0119	0.0041	0.0134
Talus	0.10	0	0	0	0.0010	0.0010	0.0010
Calcaneus	1.25	0.10	0.03	0	0.0014	0.0039	0.0041
Toes	0.22	0.035	0.0060	−0.0175	0.0001	0.0002	0.001

arm consisted of 5 degrees-of-freedom; the shoulder was modeled as a ball-and-socket joint (3 dofs), and the elbow and forearm rotation were each modeled with revolute joints (1 dof). We grouped muscles based on their main function to define 28 muscle groups: 6 lower-back and 11 lower-limb muscles. The segmental masses and inertias have been modified to model the changes measured during pregnancy (Table 1). Models of the arms, hands, and feet were estimated [22,29] and are detailed in Table 2. These segments did not exhibit significant changes during pregnancy.

The complete musculoskeletal model allows us to perform inverse dynamics and estimate muscle forces for any movement of the pregnant body or any pregnancy-related syndrome, including a loss of balance. To study the risk of falling during pregnancy, we need to estimate as accurately as possible the location of the overall center of mass of the body, which depends on the location of the center of mass of each segment. Inaccuracies will result in under- or over-estimation of fall-related parameters such as COM-COP distance [30], and time-to-boundary [31].

4.3. Evolution of muscle activation during gait

We compared the joint moments between the generic and pregnancy-specific models through pregnancy (Fig. 4a). Although RMS errors are limited for the ankle, knee, and hip moments, they become much larger for the lumbar moment. These findings confirm that pregnancy-specific models including specific inertia parameters and COM locations are necessary to obtain reliable joint moment estimations.

Using pregnancy-specific models, we computed the evolution of joint moments through pregnancy (Fig. 4b–d). The most striking effects were an increase of the ankle moment during the push-off phase and an overall increase of the lumbar moment during pregnancy. Due to an overall increase in total body mass, the ankle joint needs to output more work to maintain the same kinematics and ensure forward propulsion. The peak lumbar moment increased about 33% during the breaking and push-off phases between 9 and 37 weeks of pregnancy. The increased lumbar moment is likely related to an increase in moments of inertia for the upper body and a forward displacement of the upper body COM location through pregnancy. A sensitivity analysis will be conducted to confirm this assumption.

4.4. Limitations

The model is limited to women with an initial BMI below 26 kg/m^2. The collected data set was aimed at military personnel and included 25 subjects who all met the height/mass entrance requirements of the U.S. Air Force and U.S. Navy before they became pregnant. Military personnel is required to keep a level of fitness and physical ability beyond the average level of the civilian population; the required age-dependent BMI lies between 18.9 and 26 kg/m^2 [32]. Since the average BMI of the American woman is 26.5 kg/m^2 [33], the measurements may not represent the average American woman. In addition, the amount of mass gain during a pregnancy is highly dependent on the initial BMI of the subject: low BMI subjects tend to gain more mass than subjects exhibiting a high pre-pregnancy BMI [16]. The model would largely overestimate the mass gain for any subject with an initial BMI higher than 26 kg/m^2.

We did not consider potential modifications of muscle paths. Since the geometry of the pelvic and lower back region is modified throughout pregnancy [34], there may be a need to derive pregnancy-

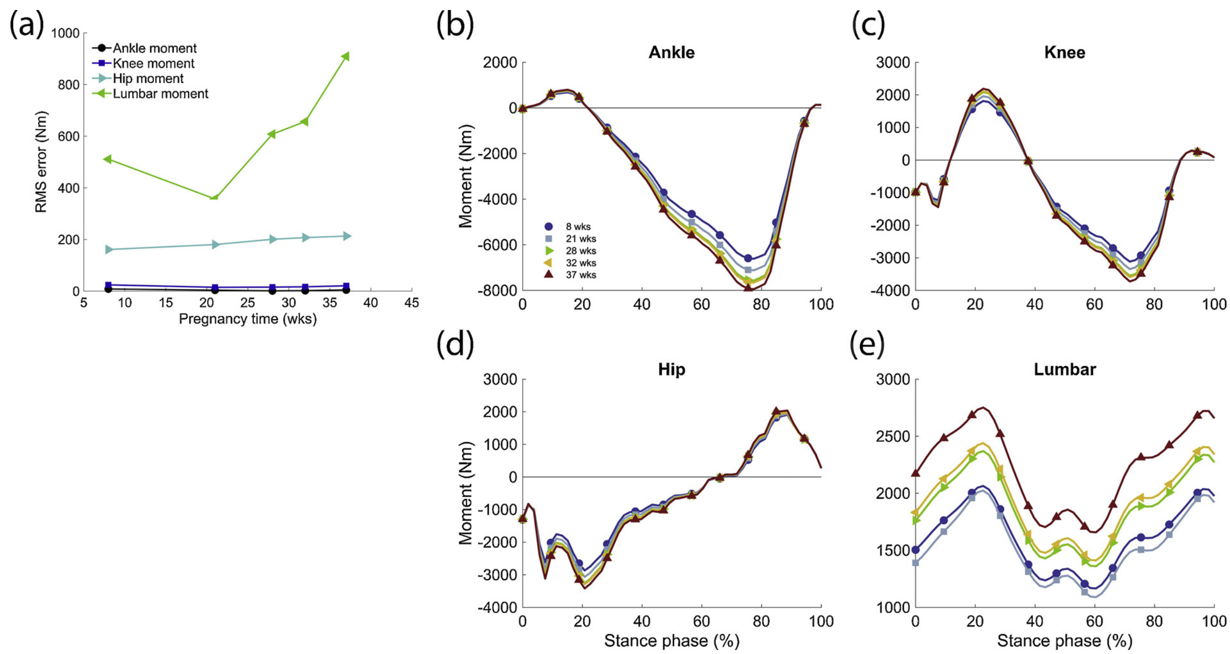

Fig. 4. (a)Root mean square of the error of the generic model relative to the pregnancy-specific model. Generic models scaled to the total amount of mass gain at each phase of the pregnancy are compared to the pregnancy-specific model where the mass distribution is modified. Results show that the change in mass cannot by itself explain the change in hip and lumbar moments during stance. As pregnancy progresses, the lumbar moment error doubles. (b–d) Net joint moments at the (b) ankle, (c) knee, (d) hip, and (d) lumbar joint during the stance phase of gait after 9, 21, 28, 32, and 37 weeks of pregnancy. Variations in ankle, knee, and hip net joint moments are very limited. The lumbar net joint moment shows an increase during the entire stance phase and the ankle net joint moment increases during the push-off phase.

specific muscle paths, based on Magnetic Resonance Images for example. However, effects of these changes on estimated muscle forces remain to be quantified.

5. Conclusion

We report here for the first time a musculoskeletal model for a Caucasian pregnant woman with initial BMI lower than 26 kg/m^2, including center of mass and inertia of each body segment throughout pregnancy. The model was developed from an extensive data set collected on pregnant subjects and validated by comparison with two independent data sets on the BMI increase and the mass and inertia increases during pregnancy, respectively. Results show that the lower trunk segment is, as expected, the most affected by pregnancy. Apart from the classical change in mass of the abdomen, the study also showed the change in mass distribution with part of the mass gain located in the gluteal region, upper trunk, thighs, and lower legs. This very specific distribution of the mass gain leads to imbalances and a potential increase in the risk of falling of pregnant women. It also leads to altered muscle forces, which in turn could lead to muscle fatigue and potential low back pain. Study of the modified biomechanics of the pregnant woman is now facilitated by the development of this musculoskeletal model. In the future, this model will be extended to pregnant women with initial BMI higher than 26 and to modified muscle paths. Future studies will focus on collecting kinematics and kinetics data on pregnant women and computing the corresponding muscle forces to estimate muscle fatigue.

Conflict of interest

The authors declare no conflict of interest regarding the content of this article.

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