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Section: Original Research

Article Title: Changes in Gait with Anteriorly Added Mass: A Pregnancy Simulation Study

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Journal: *Journal of Applied Biomechanics*

Acceptance Date: February 26, 2016

DOI: http://dx.doi.org/10.1123/jab.2015-0178

Changes in Gait with Anteriorly Added Mass: A Pregnancy Simulation Study

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Funding: Undergraduate Research Opportunities Program (UROP) at Boston University awarded to Maureen I. Ogamba, the National Institute of Arthritis and Musculoskeletal and Skin Diseases of the National Institutes of Health under Award Number K23 AR063235 awarded to Cara L. Lewis and the and the National Center for Medical Rehabilitation Research, National Institute of Child Health and Human Development/National Institute Neurological Disorders and Stroke of the National Institutes of Health under Award Number K12 HD055931 awarded to Simone V. Gill.

Conflict of Interest Disclosure: None

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“Changes in Gait with Anteriorly Added Mass: A Pregnancy Simulation Study” by Ogamba MI et al.
Journal of Applied Biomechanics

Abstract

During pregnancy, the female body experiences structural changes, such as weight gain. As pregnancy advances, most of the additional mass is concentrated anteriorly on the lower trunk. The purpose of this study is to analyze kinematic and kinetic changes when load is added anteriorly to the trunk, simulating a physical change experienced during pregnancy. Twenty healthy females walked on a treadmill while wearing a custom made pseudo-pregnancy sac (1 kg) under three load conditions: sac only, 10 pound condition (4.535 kg added anteriorly), and 20 pound condition (9.07 kg added anteriorly), used to simulate pregnancy, in the second trimester and at full term pregnancy, respectively. The increase in anterior mass resulted in kinematic changes at the knee, hip, pelvis, and trunk in the sagittal and frontal planes. Additionally, ankle, knee, and hip joint moments normalized to baseline mass increased with increased load; however, these moments decreased when normalized to total mass. These kinematic and kinetic changes may suggest that women modify gait biomechanics to reduce the effect of added load. Furthermore, the increase in joint moments increases stress on the musculoskeletal system and may contribute to musculoskeletal pain.

Keywords: gait biomechanics, pregnancy, anterior mass, kinematics, kinetics

Manuscript Word Count: 3994

Abstract Word Count: 188

Introduction

Women comprise at least half of the world's population and a large percentage of these women are, or will become, pregnant. The pregnant female body experiences hormonal, musculoskeletal, and structural changes. During the 36-40 week long gestation period of normal pregnancy, one of the most significant changes is weight gain. The Institute of Medicine recommends that women with a normal pre-pregnancy Body Mass Index (BMI) gain 11.34 to 15.88 kg (25 to 35 pounds)[1] during pregnancy. As the pregnancy progresses, the anterior mass on the trunk is increased with an average increase of 0.29 kg per week in the lower trunk segment.[2] This increase in anterior mass due to pregnancy often leads to changes in gait biomechanics.[3-8] These changes, which have been interpreted as attempts to increase stability,[4,6] may also increase the risk for musculoskeletal pain.[3,5,8]

Multiple studies have been conducted to analyze how the biomechanics of gait differs in pregnant women compared to nulliparous,[8,9] pre-pregnancy,[10,11] at varying gestational phases,[4,9-12] and post-partum women.[5-7,10-12] Recently, Branco and colleagues[13] performed a meta-analysis of the previous relevant literature regarding the biomechanics of the gait during pregnancy. The results of their meta-analysis, including nine previous studies,[3-6,9,11,12,14] illustrate a need for further investigation to clarify the effects pregnancy has on a woman's biomechanics, as findings have been varied. For instance, there are inconsistent results reported regarding stride length. Foti et al.[3] reported no change in stride length during pregnancy compared to post-partum while Branco et al.[4] reported a decrease in stride length during pregnancy comparing the 2nd and 3rd trimester. Additionally, Forczek & Staszkiewicz[11] reported an increase in base of support via increased stride width and decreased step length, which was thought to decrease single-limb support time. A common interpretation from all nine studies was that the spatiotemporal and kinematic changes observed were made in an attempt to ensure body stability during the later stages of pregnancy.[13]

One mechanism thought to drive changes in movement of pregnant women is the increased anterior mass created by the growing fetus, which changes the inertial properties of the lower trunk[2] and shifts their center of mass more anteriorly in the absence of postural changes.[15] Postural adaptations such as increased lumbar lordosis help to keep a consistent position of the woman's center of mass during pregnancy.[15] Posture and coordination of the pelvis and trunk have been investigated as other possible areas of adaptation during pregnancy, with evidence of increased anterior pelvic tilt,[3,4,8,10] decreased trunk[7] and pelvic rotation,[7,9] and decreased in-phase coordination of the trunk relative to the pelvis.[7] Additionally, previous studies interested in the effects of anterior mass on gait biomechanics in healthy (non-pregnant) adults have noted increased trunk extension,[16] decreased stride length,[17] and lower limb kinematic changes.[8,16,18] As a result, further understanding how anterior mass affects gait may reveal ways to accommodate for the pregnancy mass independent of the pregnancy-related hormonal changes.

In addition to the changes in mass distribution during pregnancy, pregnant women also experience musculoskeletal pain, such as lower back, pelvic, and limb pain. Fifty-six percent of postpartum women exhibit more symptoms of musculoskeletal pain than nulliparous women,[19] but the cause of these symptoms remains unclear. Branco and colleagues[13] urged that further research is needed focusing on lower limb joint kinetics to better understand the demands on muscles during gait and their implications on musculoskeletal pain during pregnancy. Unfortunately, few treatment recommendations exist for pregnancy-related musculoskeletal pain management, with most treatment strategies including simply waiting for resolution of the pain after childbirth.[3] Therefore, a better understanding of how the biomechanical changes exhibited with increased anterior mass may contribute to increased stress on musculoskeletal structures is warranted.

The purpose of this study was to examine kinematic and kinetic changes during gait with added anterior mass that simulated pregnancy at approximately 20 (2nd trimester) and 40 (full-term) weeks of gestation. We hypothesized that for every increase in mass added there would be significant changes to the kinematics and kinetics during gait. Understanding how women adapt to the increased anterior mass during pregnancy can aid in our understanding of modifications which may improve stability yet increase stress on musculoskeletal structures.

Methods

Twenty healthy nulliparous college-aged adult females (22 ± 3 years, 1.7 ± 0.1 m, 62 ± 9.4 kg, BMI 22.37 ± 2.52 kg/m^2) who reported no current musculoskeletal pain participated in this study. Additionally, participants with a history of hip or back surgery were excluded. All participants wore running shorts, a fitted top, and their own running shoes to allow for marker placement over the predetermined, palpated anatomical landmarks. All procedures were approved by Boston University Institutional Review Board and all subjects provided informed written consent.

Advancing pregnancy was simulated by inserting cuff weights (4.535 kg each) anteriorly into a custom made pseudo-pregnancy sac (1 kg) (Figure 1). For this study, we tested three conditions: a baseline condition with the empty sac where the only additional mass (1 kg) was distributed around the trunk (sac only condition), and two conditions simulating progressive stages in the fetal development where mass was added anteriorly within the pseudo-pregnancy sac. The sac only condition was tested first. Next, the second trimester was simulated by adding 4.535 kg anteriorly (10 pound condition) for a total of 5.535 kg of mass added. Lastly, full term

(20 pound condition) was simulated by adding an additional 4.535 kg, resulting in a total of 9.07 added anteriorly, and a total increase of 10.07 kg in overall mass.

Kinematic data were collected using a motion capture system (Vicon®, Oxford Metrics, Centennial, CO). The system included ten high resolution cameras (sampling rate of 100Hz). Kinetic data (collected at 1000Hz) during walking were measured using a split belt instrumented treadmill (Bertec ® Corporation, Columbus, OH). Twenty-five passive reflective markers were placed via double-sided tape over bony landmarks. Markers were placed bilaterally over the calcaneus, head of first and last metatarsal, medial and lateral malleoli, and medial and lateral femoral epicondyles. On the pelvis, markers were placed bilaterally on the iliac crest, greater trochanter, anterior superior iliac spine, and a single marker on the sacrum. To define the trunk segment, markers which were placed bilaterally on the acromion process, spinous process of 7th cervical vertebra, and xiphoid process were used in addition to the iliac crest markers. Non-collinear marker clusters with four reflective markers were positioned bilaterally on the distal thigh and shank. The clusters were attached to body segments with neoprene wraps (NuStimWrap, Fabrifoam® Exton, PA), that allowed the clusters to be attached via Velcro and further wrapped in pre-wrap to prevent movement.

Participants were instructed to walk five laps continuously around the lab to determine their preferred walking speed without the pregnancy sac. Participants were timed as they walked between two strips of tape on the floor, five meters apart. After calculating preferred walking speed, the pseudo-pregnancy sac and markers were placed on the participant. After marker placement, a static standing trial was taken, which was used to create a participant specific model.

Three conditions were selected for analysis: sac only condition (1 kg total mass), 10 pound condition (4.535 kg added to the sac, 5.535 kg total mass), and 20 pound condition (9.07 kg added to the sac, 10.07 kg total mass). The participant wore the pseudo-pregnancy sac in all three conditions. The participants were instructed to walk on the treadmill at their calculated preferred speed while looking straight ahead. Following a one-minute adaptation period, data were collected for each condition for two minutes, during which approximately 100 gait cycles were collected.

Three-dimensional marker trajectories and force plate analog outputs were processed using Visual3D (C-Motion, Rockville, MD). Marker trajectories were low-pass filtered using a 4^{th} order Butterworth filter with a cut-off of 6 Hz.[20] Force plate analog outputs were also low-pass filtered using a 4^{th} order Butterworth filter 10 Hz.[20] The static standing trial was used to create a participant specific model and three-dimensional joint and segment angles were determined using a Cardan X-Y-Z rotation sequence.[21] Ankle, knee and hip joint angles were defined by the angle between the distal and proximal segments. Segment angles for the pelvis and trunk were determined with respect to the global coordinate system. The pelvis was defined using the CODA model.[22] Visual3D was used to determine kinematic variables of the trunk, pelvis, and lower extremities and kinetic variables of the lower extremities; as well as a spatiotemporal measure. The kinematic variables of interest were peak ankle, knee, and hip joint angles and pelvic and trunk segment angles in the sagittal and frontal planes, as well as trunk and pelvic transverse plane excursions during the gait cycle (right heel strike to right heel strike). Kinetic variables of interest were peak ankle, knee, and hip moments in the sagittal and frontal planes. All moments were normalized in two ways: 1) using the participant's body mass and the empty sac (baseline normalized) and 2) using the participant's body mass, the sac, and the additional anterior mass (total mass normalized). This dual normalization allowed us to determine the change in the magnitude of the moments (baseline normalized) and the change in the moment relative to the new mass (total mass normalized). The spatiotemporal measure examined was stride length. Variables of interest were extracted using a custom Matlab (Mathworks® Natick, Ma) program for each stride and then averaged across the entire trial for analysis.

Using SPSS 20.0 (IBM Corp. Chicago, IL) statistical software, a repeated measure ANOVA was used to determine the main effect of load on each of the dependent variables. *Post hoc* Fisher's least significant difference (LSD) test was used to determine where any significant differences existed between the three conditions (sac only, 10 pound condition, and 20 pound condition). An alpha value of 0.05 was set for all statistical tests of significance.

Results

There were no significant kinematic differences in peak ankle angles due to added mass (Table 1). However, there was a significant effect of added mass on kinetics in the sagittal plane ($p < 0.001$ Figure 2, Table 2). Specifically, when normalized to baseline mass, the peak plantar flexion moment was greater in the 20 pound condition compared to the 10 pound condition ($p = 0.001$) and the sac only condition ($p < 0.001$). However, when normalized to total mass, the peak plantar flexion moment was less in the 20 pound condition compared to either the 10 pound ($p = 0.002$) or the sac only condition ($p < 0.001$). The plantar flexion moment was also less in the 10 pound condition than in the sac only condition ($p < 0.001$). Additionally, the peak ankle dorsiflexion moment was less in the 20 pound condition than in either other condition (10 pound condition $p = 0.032$, sac only condition $p < 0.001$), and was less in the 10 pound condition than the sac only condition ($p = 0.001$). While the baseline normalized ankle eversion moment was not different with load, the load normalized moment was less in the 20 pound condition than in the sac only condition ($p = 0.005$).

The additional mass had significant kinematic and kinetic effects at the knee in the sagittal (Figure 2, Table 1-2) and frontal planes (Figure 3, Table 1-2). In the sagittal plane, peak knee extension angle decreased with the addition of mass (10 pound condition $p = 0.028$, and 20 pound condition $p = 0.014$) compared to the sac only condition. Additionally, the baseline normalized peak knee flexion moment was greater in the 20 pound condition compared to the sac only condition ($p = 0.003$). Conversely, both the total mass normalized peak knee flexion and extension moments were less in the 20 pound condition than either the 10 pound condition (flexion $p = 0.007$, extension $p = 0.018$) or sac only condition (flexion $p < 0.001$, extension $p = 0.002$) and the 10 pound condition was less than the sac only condition (flexion $p = 0.012$, extension $p = 0.015$). In the frontal plane, the baseline normalized peak knee adduction moment was greater in the 20 pound condition than in either the 10 pound condition ($p = 0.014$) or the sac only condition ($p = 0.003$). However, there was no difference in the peak knee adduction moment when normalized by the total mass. Conversely, the baseline normalized peak knee abduction moment was not different across conditions while the total mass normalized knee abduction moment decreased with added mass ($p < 0.001$). The total mass normalized moment was less in the 20 pound condition than in either other condition (10 pound condition $p = 0.007$, sac only $p < 0.001$), and the 10 pound condition was less than the sac only condition ($p < 0.001$).

At the hip, the addition of the anterior mass had significant kinematic and kinetic effects in the sagittal (Figure 2, Table 1-2) and frontal planes (Figure 3, Table 1-2). In the sagittal plane, hip extension angle decreased with the additional mass ($p < 0.001$ and $p < 0.001$ for the 10 pound and 20 pound conditions, respectively) compared to the sac only condition. Furthermore, in the 20 pound condition peak hip extension angle decreased compared to the 10 pound condition ($p = 0.007$). The peak hip flexion angle increased with addition of mass compared to the sac only condition ($p = 0.009$ for 10 pound, $p < 0.001$ for 20 pound conditions). Additionally, the baseline normalized peak hip extension moment increased with mass ($p < 0.001$) while the total mass normalized moment did not change. The baseline normalized hip extension moment increased in the 10 pound condition ($p = 0.003$) and 20 pound condition ($p < 0.001$) compared the sac only condition. Also, the baseline normalized peak hip extension moment was greater in the 20 pound condition compared to the 10 pound condition ($p = 0.006$). The total mass normalized hip flexion moment decreased with added mass. The 20 pound condition was less than the 10 pound condition ($p < 0.001$) and sac only condition ($p < 0.001$), and the 10 pound condition was less than the sac only condition ($p < 0.001$) when normalized to total mass. In the frontal plane, the baseline normalized peak hip abduction moment increased in the 20 pound condition compared to the sac only condition ($p = 0.001$), while the total mass normalized moment decreased with added mass. The hip abduction moment in the 20 pound condition was less than in the 10 pound condition ($p = 0.004$) and the sac only condition ($p < 0.001$), and the moment in 10 pound condition was less than the sac only condition when normalized to total mass.

The addition of the anterior mass had significant effects on the peak pelvic angles in the sagittal (Figure 2, Table 1) and frontal planes (Figure 4, Table 1). Specifically, in the sagittal plane, anterior pelvic tilt increased with the addition of mass ($p = 0.003$ for the 10 pound, and $p < 0.001$ for the 20 pound conditions) compared to the sac only condition. The 20 pound condition also had greater anterior pelvic tilt than the 10 pound condition ($p = 0.025$). Similarly, peak posterior pelvic tilt decreased with the addition of mass compared to the sac only condition

($p = 0.024$ for 10 pound, and $p = 0.001$ for 20 pound conditions). Additionally, pelvic rotation in the transverse plane decreased with the addition of mass compared to the sac only condition ($p < 0.001$ for 10 pound, and $p = 0.001$ for 20 pound conditions).

The addition of anterior mass had significant effects on the trunk. There was a decrease in peak ipsi-lateral trunk flexion in the 10 pound ($p = 0.013$) and 20 pound conditions ($p = 0.005$) compared to the sac only condition (Figure 4, Table 1). Finally, the addition of load significantly decreased trunk excursion in the transverse plane compared to the sac only condition ($p = 0.001$ for the 10 pound condition and $p < 0.001$ for the 20 pound condition).

Added anterior mass had a significant effect on stride length (Table 1). The addition of mass decreased stride length compared to the sac only condition ($p = 0.009$ for the 10 pound condition and $p = 0.001$ for the 20 pound condition). Additionally, stride length was decreased in the 20 pound condition compared to the 10 pound condition ($p < 0.001$).

Discussion

The purpose of this study was to examine the kinematic and kinetic adaptations women experience when exposed to increasing anterior mass, as would occur during pregnancy. The addition of anterior mass resulted in biomechanical adjustments which are thought to increase the safety of the mother and fetus.[13] Despite adjustments in kinematics and stride length, the addition of anterior mass resulted in increased baseline normalized moments at the ankle, knee, and hip, which may contribute to the musculoskeletal pain experienced by the majority of pregnant women.[19]

Similar to Branco and colleagues'[4] third trimester condition result, we found a decrease in the peak hip extension angle in the 20 pound condition. This decrease in hip extension is often attributed to the increase in anterior pelvic tilt evident late in pregnancy[3,4,8,10] and is consistent with our results from the added anterior mass conditions. The decrease in peak hip extension angle was accompanied by a decrease in trunk and pelvic rotation excursions with the addition of anterior mass. The reduction of trunk[7] and pelvic[7,9] rotation has similarly been reported in third trimester pregnant women during both over-ground and treadmill walking tasks. The combination of kinematic changes in the hip, trunk, and pelvis has been reported to contribute to a decrease in stride length with the progression of pregnancy.[4,9] The results of this study suggest that with increased anterior mass, women decrease their stride length which also may decrease the time in single limb support.[11] This decrease in stride length has also been reported in healthy adult populations with the addition of anterior and asymmetrical mass[17] compared to the no added mass conditions.

Interestingly, our study did not detect a change in sagittal plane trunk kinematics, which has been previously reported in added anterior mass research[16] and is thought to be an important proactive strategy to increase stability.[9,15,23] However, the kinematic changes noted at the knee, hip, and pelvis may suggest an attempt to increase stability when walking with anterior mass. The results of this study propose that increased stability may come from the decreased extension angles in the knee and hip, as well as a decreased stride length exhibited in the two conditions with additional mass compared to the sac only condition. These factors produce a posture that may result in lowering the center of mass closer to the ground aiding in weight acceptance.[24] Additionally, previous research has shown that increased anterior mass significantly affects end-point (foot clearance) control in healthy adults.[16,17] In this study, an increased knee abduction angle was seen around toe off in the additional mass conditions, suggesting a limb shortening strategy to increase foot clearance during swing.

As pregnancy progresses, the female body undergoes anatomical changes to adjust to the increasing anterior mass. With bipedal posture and gait, the increased mass gained during pregnancy imposes increased risk of musculoskeletal pain and injury for the mother. Two commonly reported areas of pain are the lower back[25-28] and the pelvic girdle.[13,25,29] Approximately 50% of pregnant women experience lower back pain,[28] ~30% of which are so severe that women cease to engage in at least one regular activity.[27] Additionally, pelvic girdle pain affects approximately 20% of pregnant women.[29] While relaxin hormones have been attributed as a possible cause of both lower back pain[25,29,30] and pelvic girdle pain[25,30], increases in stress on the muscles and supporting structures of the lower back, pelvis and hip have also been implicated.[2,3,5,8,25,29,30] In this study, we eliminated the pregnancy hormonal factors by selecting non-pregnant participants that were currently not experiencing pain. The resulting data suggest that elevated baseline normalized joint moments may play a role in the development of pregnancy-related musculoskeletal pain by increasing strain on the muscles and surrounding joint structures.

The results of our current study are consistent with results of previous studies where pregnancy related pains were attributed to increased lower limb moments.[3,5,8,25] Specifically, when normalized to the baseline condition, we found increased moments at the ankle, knee, and hip with the addition of the anterior mass. At the ankle, the increased peak plantar flexion moment, which occurred during mid-stance in the heaviest condition, may be a result of increased muscular torque needed to propel the center of mass forward with the added mass. Foti and colleagues,[3] found an increased non-normalized plantar flexion moment during pregnancy compared to 1-year post-partum. At the knee, the increased peak flexion moment, which occurred at heel strike, is an additional compensation to pull the center of mass forward in the heaviest condition. Furthermore, at the hip, increased extension and abduction baseline normalized moments occurring at heel strike suggest increased muscular torque is required to continue the forward progression of the center of mass. Huang and colleagues[5] also found increasing hip extension moments during pregnancy and related this increase to complaints of sacroiliac pain in their pregnant participants.

Interestingly, when evaluating the moments normalized to total mass, the majority of the moments decreased with the addition of mass indicating that there were modifications which reduced the effect of the added mass. The ankle plantar flexion and dorsiflexion moments, knee flexion, extension and abduction moments, and the hip flexion and abduction moments decreased with each mass increase. These reductions in moment could be interpreted as a decrease in the force required of the muscles as well as the surrounding joint structures. However, the moments normalized to the participant’s baseline mass clearly indicate that the forces required of these structures will increase with the addition of anterior mass compared to the baseline condition, and may contribute to the potential for increased pain or risk of injury to the musculoskeletal system during pregnancy.

One limitation of the study was the relatively brief walking trials. Gait trials were collected for two minutes for each condition (sac only, 10 pound, and 20 pound conditions), following a one-minute acclimation period. This allowed for detection of short term changes, but not for evaluation of long term adaptations to added mass. We also did not test pregnant women and used 10 and 20 pounds to approximate the second and third trimester weight gain. However, the use of nulliparous women eliminates hormonal factors such as elasticity of the ligaments during pregnancy, which allowed us to focus on acute biomechanical changes occurring with increased anterior mass placement. While the 20 pound condition we used was less than the recommended total mass gain for women with normal BMI[1], this added trunk mass (10.07 kg) was close to the average mass gain of the trunk at 40 weeks (full-term) as previously reported by Jensen and colleagues,[2] with an average of 0.29 kg of mass gain per week. We also used the same additional mass for each participant which resulted in different masses relative to the individual's body mass. However, we attempted account for this relative change by normalizing both to baseline mass and to total mass.

In conclusion, the addition of anterior mass on the trunk changes the body's center of mass if there is not a concomitant change in posture.[15] Consequently, kinematic adjustments of the trunk, pelvis, hip and knee, as well as stride length are made. These adjustments are thought to improve stability, allowing safer gait for the pregnant woman, but may be only mechanical in nature. The adjustments do reduce the kinetic effect of the increased mass as total mass normalized moments decrease; however, the increase in baseline normalized moments at the ankle, knee and hip indicated that the adjustments do not fully offset the effect of the added mass. The results of this study suggest that the musculoskeletal pain experienced by some pregnant women may be related to the increased joint moments associated with the increased mass. However, future studies should examine if similarities exist between kinematics of pregnancy and people with non-pregnancy related pain.

Acknowledgement

Research reported in this publication was supported by the Undergraduate Research Opportunities Program (UROP) at Boston University, the National Institute of Arthritis and Musculoskeletal and Skin Diseases of the National Institutes of Health under Award Number K23 AR063235 awarded to Cara L. Lewis and the National Center for Medical Rehabilitation Research, National Institute of Child Health and Human Development/National Institute Neurological Disorders and Stroke of the National Institutes of Health under Award Number K12 HD055931awarded to Simone V. Gill. The content is solely the responsibility of the authors and does not necessarily represent the official views of the National Institutes of Health.

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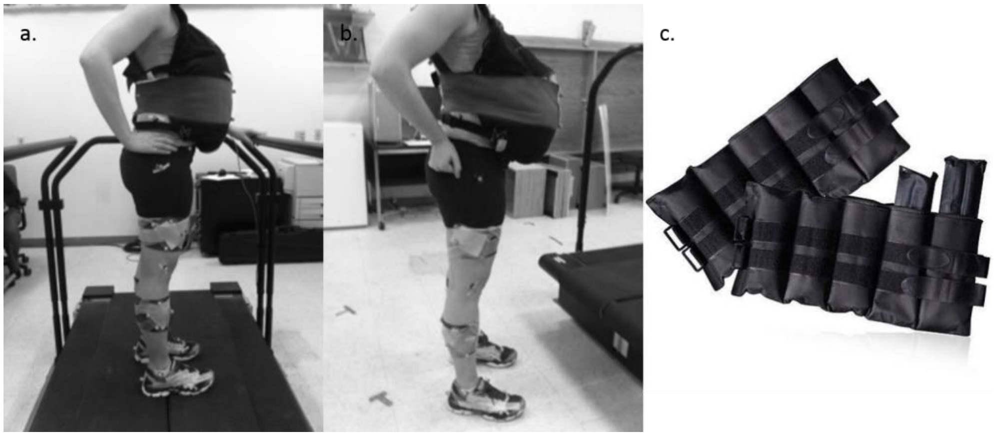

Figure 1: Participant setup with pseudo-pregnancy sac and reflective markers: a) sac only condition with empty pseudo-pregnancy sac (1 kg), b) 20 pound condition (9.07 kg), c) ankle cuff weights (4.535 kg each) used to add mass to the sac.

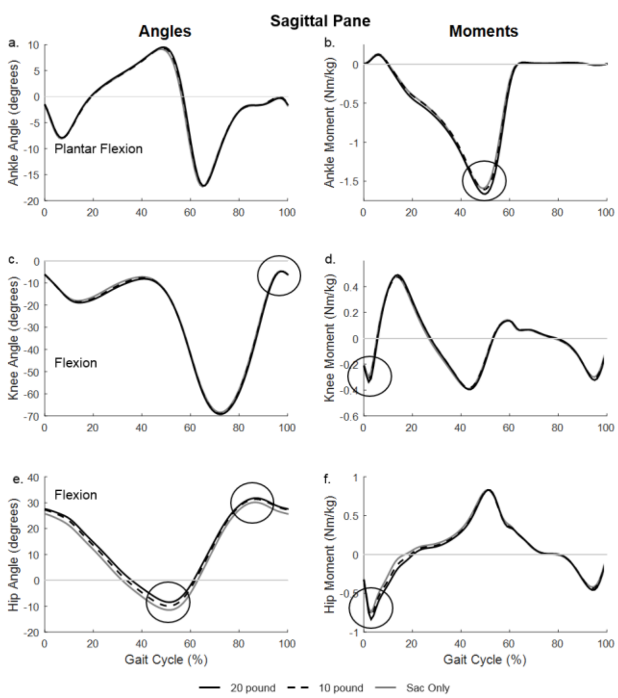

Figure 2: Sagittal plane kinematic and kinetic gait cycle plots for the ankle, knee, and hip. Circles indicate peaks where significant differences were found among conditions. Moments are baseline normalized.

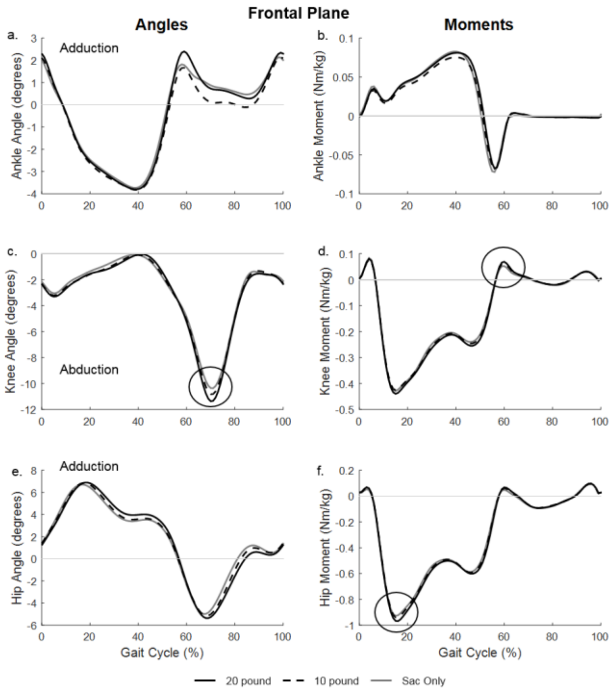

Figure 3: Frontal plane kinematic and kinetic gait cycle plots for the ankle, knee, and hip. Circles indicate peaks where significant differences were found among conditions. Moments are baseline normalized.

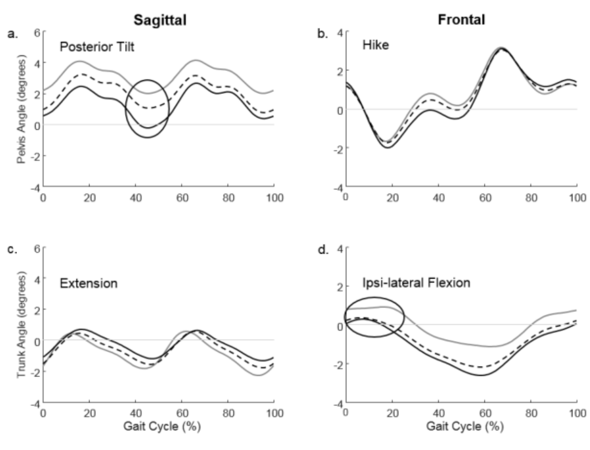

Figure 4: Sagittal and frontal plane kinematic gait cycle plots for the pelvis and trunk. Circles indicate peaks where significant differences were found among conditions.

Table 1: Mean and standard deviation (SD) for each kinematic and spatiotemporal variable during gait for the three mass conditions: sac only, 10 lbs, and 20 lbs conditions.

	Sac Only		**10 lbs**		**20 lbs**		**RM ANOVA**
Peak Values	Mean	(SD)	Mean	(SD)	Mean	(SD)	*p*-Value
Ankle							
Plantar Flexion Angle(°)	-18.2	(4.9)	-18.0	(5.2)	-18.1	(4.7)	0.904
Inversion Angle(°)	4.2	(4.1)	4.5	(4.7)	4.7	(4.3)	0.610
Knee							
Extension Angle(°)	-3.3	(5.2)	-3.7*	(5.5)	-3.8*	(5.4)	*0.014*
Abduction Angle(°)	-12.1	(7.0)	-12.3	(6.9)	-12.6*#	(7.1)	*0.042*
Hip							
Extension Angle(°)	-11.7	(3.4)	-10.1*	(3.6)	-8.7*#	(3.1)	*<0.001*
Flexion Angle(°)	30.7	(3.0)	31.8*	(3.8)	32.2*	(3.4)	*<0.001*
Pelvis							
Anterior Tilt(°)	0.7	(2.5)	-0.3*	(2.7)	-1.3*#	(2.2)	*<0.001*
Hike(°)	3.4	(1.1)	3.2	(1.3)	3.3	(1.8)	0.538
Trunk							
Extension(°)	1.5	(4.0)	1.4	(4.0)	1.5	(3.4)	0.939
Ipsi-Lateral Flexion(°)	1.3	(1.8)	0.8	(1.6)	0.7*	(1.6)	*0.002*
Excursion							
Pelvic Rotation(°)	9.3	(2.3)	8.2*	(1.7)	8.0*	(1.6)	*<0.001*
Trunk Rotation(°)	11.4	(3.0)	10.0*	(2.7)	9.4*	(2.6)	*0.002*
Stride Length(m)	1.30	(0.13)	1.29*	(0.13)	1.28*#	(0.14)	*<0.001*

*Note: *significantly different from sac only condition ($p < 0.05$), #significantly different from the 10 pound condition ($p < 0.05$)*

"Changes in Gait with Anteriorly Added Mass: A Pregnancy Simulation Study" by Ogamba MI et al.
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Table 2: Mean and standard deviation (SD) for each kinetic variable normalized to baseline mass and total mass during gait for the three mass conditions: sac only, 10 pound, and 20 pound conditions.

	Sac-Only		Baseline Normalized Nm/kg (body mass + sac) 10 pound		20 pound		RM ANOVA	Total Mass Normalized Nm/kg (body mass + sac + anterior mass) 10 pound		20 pound		RM ANOVA
	Mean	(SD)	Mean	(SD)	Mean	(SD)	*p*-Value	Mean	(SD)	Mean	(SD)	*p*-Value
Ankle												
Plantar Flexion	-1.59	(0.28)	-1.61	(0.29)	-1.67$^{*\#}$	(0.29)	*<0.001*	-1.50*	(0.26)	-1.45$^{*\#}$	(0.24)	*<0.001*
Dorsiflexion	0.14	(0.07)	0.13	(0.07)	0.13	(0.07)	0.191	0.13*	(0.06)	0.11$^{*\#}$	(0.06)	*<0.001*
Eversion	-0.10	(0.08)	-0.10	(0.02)	-0.10	(0.02)	0.889	-0.09	(0.07)	-0.09*	(0.08)	*0.013*
Inversion	0.12	(0.09)	0.12	(0.09)	0.12	(0.10)	0.381	0.11	(0.08)	0.11	(0.08)	0.052
Knee												
Flexion	-0.44	(0.11)	-0.45	(0.11)	-0.45*	(0.11)	*0.025*	-0.42*	(0.11)	-0.40$^{*\#}$	(0.09)	*<0.001*
Extension	0.45	(0.22)	0.51	(0.24)	0.52	(0.25)	0.254	0.41*	(0.22)	0.45$^{*\#}$	(0.21)	*0.002*
Abduction	-0.46	(0.20)	-0.46	(0.21)	-0.47	(0.21)	0.06	-0.43*	(0.19)	-0.41$^{*\#}$	(0.18)	*<0.001*
Adduction	0.12	(0.05)	0.12	(0.05)	0.13$^{*\#}$	(0.05)	*0.011*	0.11	(0.05)	0.11	(0.05)	0.256
Hip												
Extension	-0.75	(0.21)	-0.80*	(0.22)	-0.84$^{*\#}$	(0.22)	*<0.001*	-0.74	(0.20)	-0.73	(0.18)	0.185
Flexion	0.85	(0.19)	0.84	(0.21)	0.85	(0.21)	0.804	0.78*	(0.19)	0.74$^{*\#}$	(0.18)	*<0.001*
Abduction	-0.95	(0.34)	-0.97	(0.36)	-0.99*	(0.37)	*0.007*	-0.90	(0.33)	-0.86$^{*\#}$	(0.32)	*<0.001*
Adduction	0.17	(0.05)	0.17	(0.06)	0.18	(0.07)	0.225	0.15	(0.05)	0.16	(0.06)	0.14

*Note: *significantly different from sac only condition ($p < 0.05$), $^{\#}$significantly different from the 10 pound condition ($p < 0.05$)*