2024-05-01

Mechanical Adaptations To Anterior Load And Fall Risks In Nulliparous Women With Simulated Gestational Weight Gain

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MECHANICAL ADAPTATIONS TO ANTERIOR LOAD AND FALL RISKS IN NULLIPAROUS WOMEN WITH SIMULATED GESTATIONAL WEIGHT GAIN

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MECHANICAL ADAPTATIONS TO ANTERIOR LOAD AND FALL RISKS IN NULLIPAROUS WOMEN WITH SIMULATED GESTATIONAL WEIGHT GAIN

by

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DISSERTATION

Presented to the Faculty of the Graduate School of

The University of Texas at El Paso

in Partial Fulfillment

of the Requirements

for the Degree of

DOCTOR OF PHILOSOPHY

Interdisciplinary Health Sciences Doctoral Program

THE UNIVERSITY OF TEXAS AT EL PASO

May 2024

ACKNOWLEDGEMENTS

First and foremost, I would like to express my gratitude to Dr. Jeff Eggleston for his unwavering support throughout my tumultuous journey as a graduate student. His guidance, mentorship, and critical feedback have not only been instrumental in completing this dissertation, but also in my growth as an academic and professional.

I am also deeply thankful to my dissertation committee members, Dr. Vannesa Mueller and Dr. Jeannie Concha for their constructive criticism, valuable suggestions, and for supporting me and this body of work, despite the topic being outside of their scope of practice. I would also like to thank Dr. Robert Catena for being an inspiration for pregnancy research and providing his expertise in the field.

I appreciate my colleagues and friends for their encouragement. Special thanks to Brittaney Huskey for her invaluable support, both inside and outside of the laboratory; and to Alyssa Monero for her substantial assistance with data collection. Additional thanks to Samantha Gaytan and Micheal Sanchez for our late-night writing sessions.

I extend my heartfelt appreciation to my family for their love and encouragement. Special appreciation to my late father and late partner, both of whom played pivotal roles in the completion of this chapter in my life.

I would like to acknowledge the financial support provided by the Dodson Research Grant, which funded this research.

Finally, and most importantly, I am indebted to the extraordinary women who participated in the varying stages of this research project. Without their contributions of their time, physical presence, and experiences, this research would not have been possible. This dissertation is my humble contribution to improving the lives of women.

ABSTRACT

Approximately twenty-five percent of pregnant women experience falls during pregnancy, posing significant risks to maternal and fetal health outcomes. However, research on falls in pregnant women remains limited. **PURPOSE:** This study aimed to address this gap by comparing static postural stability between pregnant and nulliparous women with simulated gestational weight gain (GWG), thus validating the use of simulated pregnancy; and to observe the mechanical influences of simulated GWG on postural stability in static and dynamic conditions. It was hypothesized that increasing simulated GWG would decrease postural stability in single-limb stance, bilateral standing, margin of stability during gait initiation, and dynamic stability during a slip perturbation. **METHODS:** Eleven nulliparous women completed four separate data collections while adorning an anteriorly loaded weight-vest with 0, 2.26, 9.07, and 15.88 kg. Participants performed repeated balance, gait initiation, and treadmill slip perturbations under each weighted condition. **RESULTS:** Pregnant women and simulated pregnant women exhibited similar postural stability characteristics. Static balance was found to only be affected in anteroposterior sway magnitude between baseline and simulated second trimester in bilateral standing. Margin of stability was not significantly different across simulated trimesters during gait initiation. Dynamic postural stability displayed significant decreases at each heel strike between baseline and second trimester, and additional significance at heel strike one and heel strike three between baseline and third trimester. **CONCLUSION:** Overall, these findings suggest that the mechanical influences of simulated GWG influence the compensatory performance of postural stability during dynamic movements. This study contributes to the understanding of the mechanical aspects of GWG and may translate to the pregnancy-related postural stability changes and fall-risks.

LIST OF ACRONYMS

1T	First trimester
2T	Second trimester
3T	Third trimester
ANOVA	Analysis of variance
AP	Anteroposterior
APA	Anticipatory Postural Adjustments
B	Baseline
BAP	Bilateral anteroposterior sway magnitude
BCOPEa	Bilateral center of pressure excursion area
bh	Body height
BMI	Body mass index
BML	Bilateral mediolateral sway magnitude
BOS	Base of support
CDC	Centers for Disease Control and Prevention
CNS	Central nervous system
COM	Center of mass
COP	Center of pressure
COPEa	Center of pressure excursion area
EC	Eye closed
EO	Eyes open
g	Gravitational acceleration
GWG	Gestational weight gain

HS1	Heel strike before slip
HS2	First recovery slip
HS3	Second recovery slip
IOM	Institute of Medicine
LIFT	Foot lift; between leading limb heel-off and leading limb foot off
ML	Mediolateral
MOS	Margin of stability
P_{COM}	Center of mass position
ROM	Range of motion
SL	Step length
SLS	Single-limb stance
SLS-AP	Single-limb stance anteroposterior sway magnitude
SLS-ML	Single-limb stance mediolateral sway
SLS-COPEa	Single-limb center of pressure excursion area
STEP	Step execution; between leading limb foot off and leading limb heel contact
SW	Step width
vGRF	Vertical ground reaction force
V_{COM}	Center of mass velocity
xCOM	Extrapolated center of mass

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CHAPTER 1

INTRODUCTION

Pregnant women fall at rates that are similar to that of the elderly population, with one in four women falling while performing daily tasks (Dunning, LeMasters, & Bhattacharya, 2010). Following a severe fall, pregnant women are at a 2.3 times higher risk than nongravid women to require hospitalization (Weiss, 1999). Furthermore, pregnant women who fall are at an increased risk of developing poor maternal health outcomes, such as pre-term labor or delivery, placental abruption, cesarean delivery, and/or developing poor fetal health outcomes, such as fetal distress, fetal hypoxia, respiratory distress syndrome, and fetal death (Schiff, 2008). Fall prevention in pregnant women is imperative, leading to the current research in balance control (Bagwell et al., 2022; Butler, Colón, Druzen, & Rose, 2006; Catena, Campbell, Werner, & Iverson, 2019; Danna-Dos-Santos et al., 2018; Jang, Hsiao, & Hsiao-Wecksler, 2008a), sit-to-stand tasks (Catena, Bailey, Campbell, & Music, 2019; Gilleard, Crosbie, & Smith, 2002; Gilleard, Crosbie, & Smith, 2008), gait characteristics (Branco, Santos-Rocha, Aguiar, Vieira, & Veloso, 2013; Branco, M., Santos-Rocha, Vieira, Aguiar, & Veloso, 2015; Forczek & Staszkiewicz, 2012; Foti, Davids, & Bagley, 2000; Gilleard, Crosbie, & Smith, 2002; Gottschall, Sheehan, & Downs, 2013; Lymbery & Gilleard, 2005; McCrory, J., Chambers, Daftary, & Redfern, 2014b), and stair kinetics (McCrory, J., Chambers, Daftary, & Redfern, 2013; McCrory, J., Chambers, Daftary, & Redfern, 2014a).

Fall risks are multifaceted, involving a combination of the environment, the task, and the individual (Hsiao & Simeonov, 2001).The environment in which most pregnant women report a fall include walking on slippery surfaces, in poorly lit rooms, in loud environments, and while using stairs without a handrail (Dunning et al., 2003). In performing daily tasks, pregnant women reported that falls occurred more frequently while hurrying, changing direction quickly, carrying an object (such as another child), or while wearing inappropriate shoes (backless, slick, loose, high heels; Dunning, LeMasters, & Bhattacharya, 2010). On an individual level, many factors related to pregnancy can affect stability or balance control in women differently. On average, pregnant women gain upwards of twenty-five percent of their pre-pregnancy mass (Hagan & Wong, 2010), which could decrease stability by shifting the body's center of mass outside of the base of support made by the feet, due to the unevenly distributed gestational weight gain (GWG) in the torso (Whitcome, Shapiro, & Lieberman, 2007). Consequently, instability related to GWG disproportionately stresses the lumbar spine, which may lead to counteractive responses such as lumbar lordosis, weakened abdominal muscles, and an anterior pelvic tilt commonly associated with low back and pelvic pain (Norén, Östgaard, Johansson, & Östgaard, 2002). Additionally, hormonal fluctuations elicit physiological changes, affecting the normal function of major systems and extremities (Talbot & Maclennan, 2016). The combination of increased hormone secretion with increasing GWG, ligament laxity, altered posture, increasing load on the femoral arteries, and increasing venous pressure can contribute to edema, or swelling of the lower limbs (Soma-Pillay, Nelson-Piercy, Tolppanen, & Mebazaa, 2016) and morphological changes to the feet (Rao, Baumhauer, Tome, & Nawoczenski, 2009; Segal et al., 2013). Therefore, while individual factors of balance control may differ, pregnant women are at a greater disadvantage for fall risks than non-pregnant women.

The ability to maintain balance, often referred to as postural control, requires sensory information from the visual, audio-vestibular, and somatosensory systems (Peterka, 2018), which can be affected by the physiological adaptations occurring throughout pregnancy, thus creating a higher risk of falls. Ocular changes are common during pregnancy, wherein some women can experience increased corneal sensitivity and thickness, increased discomfort when wearing contact lenses, and decreased intraocular pressure (Samra, 2013); all of which can affect vision. Pregnancy may also exacerbate pre-existing ocular pathologies or incite ocular complications as a result of pregnancy-related diseases (i.e., pre-eclampsia, eclampsia, idiopathic intracranial hypertension, Grave's disease, etc.) (Samra, 2013). Similarly, the audio-vestibular system can be affected by the onset of hormones from pregnancy alone, or heighten pre-existing conditions such as hearing loss, tinnitus, autophony, and vertigo (Serna-Hoyos et al., 2022), which can negatively affect balance. Also, hormonal vertigo and the affected vestibular system may correlate to nausea, vomiting, and dizziness frequently reported during pregnancy (Black, 2002). Lastly, the somatosensory system is affected during pregnancy. Pain can limit neuromuscular control (Hlaing, Puntumetakul, Wanpen, & Boucaut, 2020) and reduce sense of joint position (Ruhe, Fejer, & Walker, 2011), which can impair postural stability. Notably, pain is often reported during pregnancy in the lower back and pelvic region (Mogren & Pohjanen, 2005), joints of the lower extremities (Vullo, Richardson, & Hurvitz, 1996), and feet (Karadag-Saygi, Unlu-Ozkan, & Basgul, 2010). Additionally, with increased ligament laxity, joint weakness can increase, and decreased proprioception has been reported in the knees and ankles (Bányai et al., 2009; Preetha & Solomon, 2011). Any of these sensory systems being affected during pregnancy may play a pivotal role in fall risks; especially when considering the conditions in which women reported falling, including in poorly lit, loud, and slippery environments (Dunning et al., 2003).

In addition to hormonal and physiological changes during pregnancy, physical adaptations may also increase the likelihood of instability, or sense of instability, during dynamic movements, such as gait (Branco, Santos-Rocha, Aguiar, Vieira, & Veloso, 2013; Carpes, Griebeler, Kleinpaul, Mann, & Mota, 2008; Foti, Davids, & Bagley, 2000; Krkeljas, 2018; Sawa et al., 2015). Pregnant women are often cited to have a cautious “waddle” to their gait, characterized by decreased gait velocity, increased step width, and increased double-limb support time (Gottschall, Sheehan, & Downs, 2013). Some studies have reported altered Joint kinematics, with reports of limited hip range of motion (ROM) (Foti, Davids, & Bagley, 2000), increased knee ROM (Branco, Santos-Rocha, Aguiar, Vieira, & Veloso, 2013), and decreased ankle ROM (Branco et al., 2016). Additionally, pelvic and thoracic torso rotations are limited, creating stiffened rotational movement in the torso, which creates movement as a single unit rather than separate pelvis/thoracic movement patterns in non-pregnant women (Wu et al., 2004).

As a caveat to the beforementioned findings, it is important to note that the majority of biomechanical pregnancy research does not represent women with an average pre-pregnancy body mass index (BMI) that is above normal; nor do they include women who gain excessive gestational mass (Foti, Davids, & Bagley, 2000; Marco Branco, Rita Santos-Rocha, & Filomena Vieira, 2014; McCrory, J. L., Chambers, Daftary, & Redfern, 2010). Whether these exclusions are intentional or systematically coincidental is unclear, as there is no mention of stratification of women by BMI category (for example, if regression of anthropometric data occurred). However, between the years of 1999 – 2008, the prevalence of reproductive-aged women with a BMI classified as overweight to obese was roughly 30 percent (Flegal, Carroll, Ogden, & Johnson, 2002); and as of 2018, the rate has since increased to roughly 40 percent (Ogden et al., 2020). Moreover, according to the Centers for Disease Control and Prevention (Centers for Disease Control and Prevention, 2022), only 32 percent of pregnant women gain within the Institute of Medicine (IOM) recommended guidelines (Institute of Medicine, 2009), with 21 percent gaining below, and 48 percent gaining above (CDC, 2022). Therefore, the literature on pregnancy biomechanics may only represent a small percentage of the women who are pregnant, and an even smaller representation of women at risk. This potential lack of representation is concerning considering that obesity has also been associated with increased fall risks (Fjeldstad, Fjeldstad, Acree, Nickel, & Gardner, 2008), and may attribute to an unexplored part of the problem.

While research in pregnancy biomechanics is expanding, the mechanical influences of pregnancy on falls remain debated. Longitudinal research studies in pregnancy biomechanics are limited, which could be due to a number of reasons: the ethical and logistical difficulty of recruiting women prior to pregnancy, lack of attrition, possible on-set health complications excluding women with high-risk pregnancies from research, and/or constrained time to collect at multiple time periods (i.e. pregnant women do not remain pregnant indefinitely). Additionally, there are few studies related to clinical assessments of falls that could be utilized to preemptively predict fall risks in pregnant women. The overall purpose of this dissertation is to quantify static and dynamic stability and assess fall risks related to the mechanical influences of pregnancy in nulliparous women with simulated gestational mass added at each "trimester," without putting pregnant women at risk. Within the overall purpose, there are three specific aims related to the simulated pregnancy conditions, including: 1) to examine potential static balance adaptations, 2) to examine dynamic balance adaptations and to determine if there is a correlation between static balance performance and dynamic stability, and 3) to examine dynamic stability during slip recovery and to determine if static and dynamic stability measurements correlate to slip recovery outcomes. Ideally, the potential connections made in this study between static and dynamic stability could translate to creating predictive measurements for fall risk assessments in pregnant women.

SPECIFIC AIMS

The overall aim of this project is to observe static and dynamic stability and to assess fall risks related to the mechanical influences of pregnancy, without putting pregnant women at risk.

Aim 1 – Static Balance

Static postural stability will be measured to compare strategies of maintaining an upright posture across increasing simulated gestational weight gain following the recommended guidelines for women with a normal pre-pregnancy body mass index. The aim of this study is to examine potential static postural stability mechanics influenced by gestational weight gain.

Aim 2 – Dynamic Balance

Dynamic stability will be measured while performing gait initiation with increasing simulated gestational weight gain following the recommended guidelines for women with a normal pre-pregnancy body mass index. The aim of this study is to examine dynamic postural stability mechanics influenced by gestational weight gain.

Aim 3 – Dynamic Stability in Slip Recovery

Slip recovery strategies will be measured following a slip perturbation with increasing simulated gestational weight gain following the recommended guidelines for women with a normal pre-pregnancy body mass index. The aim of this study is to examine dynamic postural stability recovery mechanics influenced by gestational weight gain.

Preliminary Study: Comparing Single-Limb Static Balance Between Pregnant Women Nulliparous Women with Simulated Gestational Weight Gain

The use of nulliparous women for pregnancy research remains highly controversial due to the number of hormonal, physiological, and biomechanical changes that make nulliparous women relatively incomparable to pregnant women (Talbot & Maclennan, 2016; Tan & Tan, 2013). However, utilizing nulliparous women, especially in fall research, eliminates risks to pregnant women while still allowing for the observation of the anthropometric characteristics and mechanical influences of pregnancy, exclusively. The inclusion of this preliminary study is to validate the use of nulliparous women with simulated gestational weight gain (GWG) in fall research by comparing static balance performance between pregnant and nulliparous women with similar anthropometrics created by adding simulated gestational mass.

Few studies have used nulliparous women with simulated pregnancy conditions for research, with very few similarities in methodology for determining the added anterior load. Two related studies from Gill and colleagues (2016) and Ogamba and colleagues (2016) compared simulated pregnancy conditions using a fabricated pregnancy sac, revealing that the increased load altered gait velocity, and joint kinematics changed in the frontal and sagittal planes for the knee, hip, pelvis, and trunk, respectively. However, the anterior loads used in both studies were the same for every participant, despite participants varying between healthy and overweight body mass index (BMI), and the added mass was less than the recommended range for GWG by the Institute of Medicine (Institute of Medicine, 2009), which could limit the translation of their results to pregnant women. Whereas Aguiar and colleagues (2015) were empirically the only group to compare pregnant women to simulated pregnant women in the same study; however, the anthropometrics of the non-pregnant participants and the pregnant participants were not matched, which could have affected their comparisons because the pregnant group had a higher average body mass and BMI compared to the non-pregnant group, even with the added simulated pregnancy mass. More recently, Danna-dos-Santos and colleagues (2022) observed the center of pressure (COP) posturography of women wearing seven percent of their individual body mass anteriorly for 24 hours, finding no significant alterations to postural sway throughout the prolonged period of an entire day; yet, seven percent of an individual's body mass may not elicit change to static stability, nor compare to pregnancy which can account for an additional twenty-five percent of pre-pregnancy body mass (Hagan & Wong, 2010). Therefore, comparing the mechanical movement patterns of pregnant women to nulliparous women, anthropometrically matched with added simulated gestational mass, have not been measured. The propose of this pilot study was to compare static balance mechanics between pregnant women and nulliparous women with anteriorly loaded mass to validate using simulated pregnancy conditions for research. It was hypothesized that static stability measurements would be similar between pregnant women and nulliparous women with matched simulated gestational masses. Additionally, differences due to acute accommodations to the load in the simulated pregnancy groups compared to the chronic accommodations to the load in the pregnancy groups were expected. If this hypothesis was supported, there could be some justification in using nulliparous women with simulated gestational masses in research, especially during high-risk tasks.

Methods

Participants

A convenience sample of four previously collected second trimester (2T) women (24.0 ± 3.6 years, 1.6 ± 0.1 m, 74.6 ± 13.1 kg, 28.0 ± 5.3 kg/m^2) and four previously collected third trimester (3T) women (27.5 ± 5.5 years, 1.7 ± 0.1 m, 89.6 ± 17.5 kg, 31.8 ± 3.4 kg/m^2) were individually matched in height, assumed pre-pregnancy mass, and BMI to eight nulliparous women (23.6 ± 3.1 years, 1.6 ± 0.1 m, 64.3 ± 11.1 kg, 23.5 ± 2.4 kg/m^2). All participants were required to be between the ages of 18 and 34 and free of lower limb injuries. Pregnant participants were excluded if deemed "high risk" by their own physician. Nulliparous participants were excluded if they had ever been pregnant. Pregnant women were collected a single time during their pregnancy and were classified into groups based on their self-reported trimester into either 2T (second trimester; between 13 and 27 weeks), or 3T (third trimester; 28 to 39 weeks). Prior to completing any study-related tasks, written informed consent was obtained on institutional approved documentation (Protocol No: 1414288-8; Protocol No: 1865800-3) and in accordance with 1964 Declaration of Helsinki.

Procedure

Previously collected pregnant women and newly collected nulliparous women performed the same balance tasks. Nulliparous women were fitted with a weight vest, loaded anteriorly, to match the mass and BMI of pregnant women at the time of their pregnancy collection; added mass could not exceed 22.7 kg to avoid injury to participants. To adjust to the anterior load, nulliparous women walked at a self-selected pace on a treadmill for 10 minutes prior to data collection.

During all static balance trials, participants were instructed to perform a single-limb balance for twenty seconds on the left and right limbs separately, at twenty second intervals. Participants performed balance tasks on the same mark on a single force platform, with the arms crossed over the stomach and eyes looking straight ahead. Laboratory personnel spotted participants while performing the tasks to reduce the risk of falling. Center of pressure (COP) data were obtained with the in-ground force platform (1,000 Hz, Advanced Mechanical Technology Inc., MA, USA).

Data Reduction

Raw COP data were exported from Vicon Nexus and imported into MATLAB (The Mathworks, Inc., Natick, MA, USA) to be filtered with a low-pass Butterworth digital filter with a 12.5 Hz cut-off (Callahan, 2017). Posturograms were created by the center of pressure excursion area (COPEa). COPEa was defined by the absolute maximum and minimum medial-lateral (X) and anterior/posterior (Y) coordinate data from the equation: $COPE_a = (X_{max} - X_{min}) \times (Y_{max} - Y_{min})$ in mm^2 (Callahan, 2017) to determine the overall area. Additionally, sway magnitude in the anteroposterior (AP Sway) and mediolateral (ML Sway) directions were analyzed separately.

Statistical Analysis

Statistical analyses were conducted using SPSS 29 (IBM Corp, Armonk, NY, USA), with all mean and standard deviations being determined for each variable. Separate independent t-tests were conducted to compare anthropometrics (age, height, mass, and BMI) between 2T pregnant women and nulliparous women with 2T simulated mass and between 3T pregnant and nulliparous women with 3T simulated mass.

Paired t-tests (α=0.05) were used to compare left and right limb COP values. Left and right limb COP values were collapsed into single-limb condition values (COPEa, AP Sway, ML Sway), as there were no significant differences between limbs (p > 0.05).

Separate independent t-tests were conducted to compare balance values (COPEa, AP Sway, ML Sway) between 2T pregnant women and nulliparous women with 2T simulated mass, and balance values (COPEa, AP Sway, ML Sway) between 3T pregnant and nulliparous women with 3T simulated mass.

Results

Independent t-test results comparing anthropometrics between 2T women and matched nulliparous women with 2T simulated gestational mass are presented in **Table 1.**

Table 1. Anthropometrics between second trimester women and matched nulliparous women with simulated gestational mass (mean (standard deviation))

	Pregnant	Simulated	t-value	df	p-value
Age (years)	24.00 (3.55)	24.00 (4.08)	5.89	6	1.00
Height (m)	1.64 (0.07)	1.63 (0.05)	0.41	6	0.83
Mass (kg)	74.63 (13.08)	81.00 (13.98)	0.67	6	0.53
BMI (kg/m^2)	28.00 (5.25)	30.46 (3.80)	0.76	6	0.48

Results revealed no statistical significant difference in COPEa (*M* = 327.64, 95% CI [-284.39, 939.67], t(6) = 1.31, *p* = 0.24), AP Sway (*M* = 5.37, 95% CI [-9.02, 19.75], t(6) = 0.97, *p* = 0.38), nor ML Sway (*M* = 4.32, 95% CI [-2.71, 11.35], t(6) = 1.50, *p* = 0.18) between 2T women and matched nulliparous women. **Figure 1, Figure 2, & Figure 3** display means and standard deviations for COPEa, AP Sway, and ML Sway, respectively, between 2T women and matched nulliparous women.

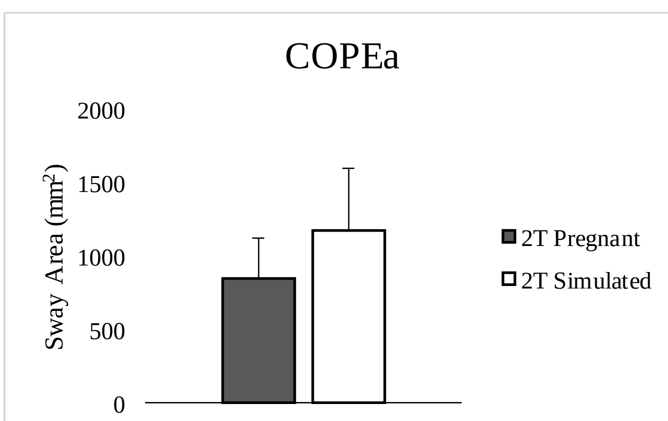

Figure 1. Center of pressure excursion area means and standard deviations between second trimester pregnant women and nulliparous women with simulated second trimester mass.

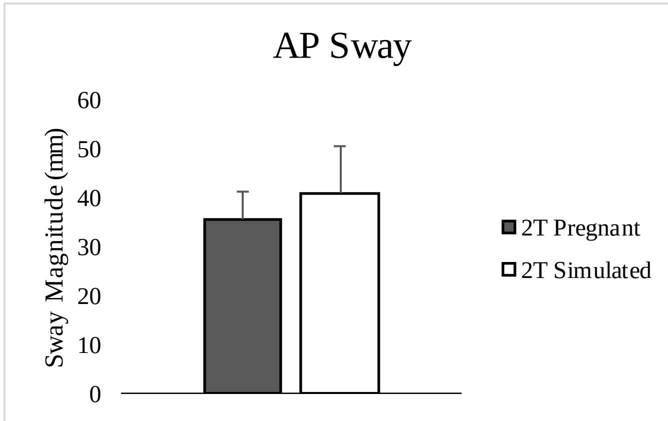

Figure 2. Anteroposterior sway magnitude means and standard deviations between second trimester pregnant women and nulliparous women with simulated second trimester mass.

Independent t-test anthropometric results comparing 3T women and matched nulliparous women with 3T simulated gestational mass are displayed in **Table 2.**

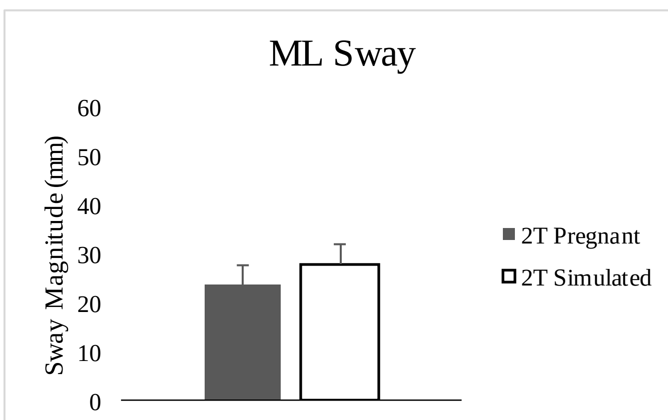

Figure 3. Mediolateral sway magnitude means and standard deviations between second trimester pregnant women and nulliparous women with simulated second trimester mass.

Table 2. Anthropometrics between third trimester women and matched nulliparous women with simulated gestational mass (mean (standard deviation))

	Pregnant	Simulated	t-value	df	p-value
Age (years)	27.50 (5.51)	24.75 (2.36)	-0.92	6	0.39
Height (m)	1.67 (0.08)	1.69 (0.05)	0.26	6	0.80
Mass (kg)	89.58 (17.50)	86.70 (22.06)	-0.21	6	0.85
BMI (kg/m^2)	31.76 (3.39)	31.63 (4.18)	-0.05	6	0.96

Results revealed no statistical significant difference in COPEa (M = -93.62, 95% CI [-1308.35, 1121.11], t(6) = -0.19, p = 0.86), AP Sway (M = -2.67, 95% CI [-22.85, 17.50], t(6) = -0.32, p = 0.76), nor ML Sway (M = 1.10, 95% CI [-10.46, 12.65], t(6) = 0.23, p = 0.82) between 3T women and matched nulliparous women. **Figure 4, Figure 5, & Figure 6** display means and standard deviations for COPEa, AP Sway, and ML Sway, respectively, between 3T and matched nulliparous women.

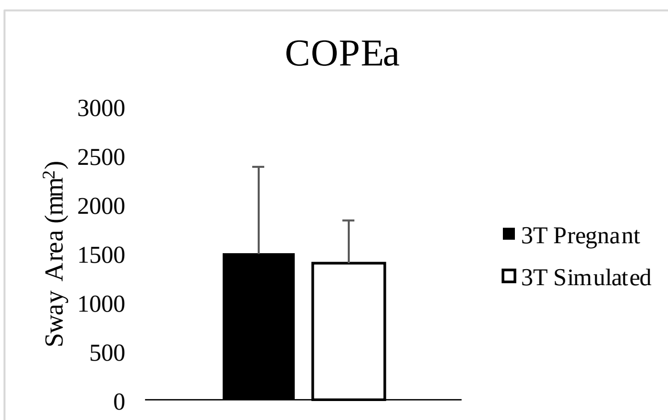

Figure 4. Center of pressure excursion means and standard deviations between third trimester pregnant women and nulliparous women with simulated third trimester mass.

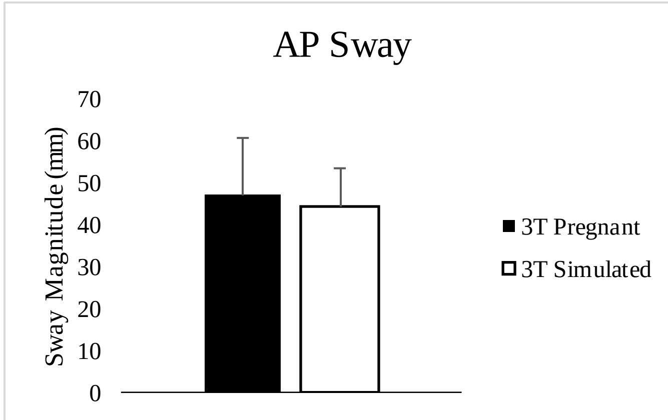

Figure 5. Anteroposterior sway magnitude means and standard deviations between third trimester pregnant women and nulliparous women with simulated third trimester mass.

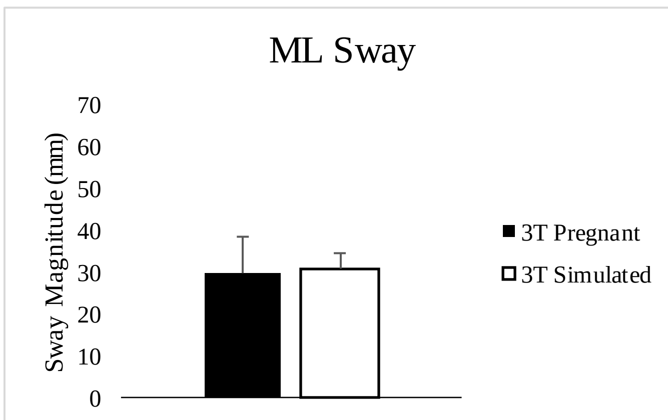

Figure 6. Mediolateral sway magnitude means and standard deviations between third trimester pregnant women and nulliparous women with simulated third trimester mass.

Discussion

The purpose of this preliminary study was to validate the use of simulated pregnancy conditions in research by comparing single-limb balance mechanics between pregnant women and nulliparous women with simulated gestational mass. The hypothesis that static stability measurements would be similar between pregnant women and nulliparous women when matched with simulated gestational masses was supported by the results of this study.

It can be assumed that participant anthropometrics were accurately matched between the pregnant and simulated gestational mass groups due to the lack of statistical differences between the groups. To the authors' knowledge, this is a novel approach of matching anthropometrics of pregnant and simulated pregnant groups on an individual and group level. Matching each pregnant woman individually may explain the similarities in single-limb balance performance observed between the groups.

Single-limb balance performance between pregnant and simulated gestational mass groups were statistically similar in the current study, which may suggest that the amount of mass elicits similar balance mechanics. While the distribution of mass could only be localized to the torso for the simulated gestational mass group in current study, it seemingly had no significant impact on sway magnitudes between the groups. These results contradict the findings from Aguiar and colleagues (2015), wherein they determined some similarities between gait characteristics (step time, stance time, double-limb support time, maximum hip extension, maximum pelvic obliquity, maximum pelvic obliquity range of motion, and peak hip flexion moments of force) between pregnant women and women with an external load, but that the distribution of the added external mass was a limiting factor. However, it is also possible that pregnant women and women with simulated gestational mass may have similar balance mechanics while in a static position, as opposed to during a task that requires dynamic balance, such as gait.

While the goal of this study was to validate the use of nulliparous women with gestational mass, single-limb balance was assessed specifically in the current study due to the task's clinical applications in balance performance and training (Bagwell et al., 2022). Previous findings from Bagwell and colleagues (2022) determined that single-limb balance sway magnitude and velocity decreased, but postural responses required to remain balanced increased into the second and third trimesters. Therefore, if women with simulated gestational mass perform similarly to pregnant women in the single-limb task, it may potentially allow for further investigation of other tasks relevant to fall risks—especially if the amount of simulated gestational mass is grounded in the IOM recommendations for gestational weight gain.

In conclusion, while simulated gestational mass cannot replace the complexities of pregnancy biomechanics, there may be beneficial and translational applications to testing the mechanics of increasing gestational mass in non-pregnant women, especially during tasks that may put pregnant women at risk of falling. Some limitations of this study include the very small sample size and single balance task that was measured; future studies should expand on the sample size and incorporate other balance and gait tasks that would be worth exploring. Additionally, pre-pregnancy mass was assumed rather than collected as self-reported data, which may have been more accurate in determining simulated gestational mass magnitude.

CHAPTER 2

Static Stability in Nulliparous Women with Simulated Gestational Weight Gain

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Static Stability in Nulliparous Women with Simulated Gestational Weight Gain

Significance of Chapter

This study compared the mechanical adaptations made during static balance tasks related to simulated gestational weight gain across trimesters. Posturography is a common diagnostic measurement of hindered static balance performance (Chaudhry, Bukiet, Ji, & Findley, 2011); therefore, these data may contribute to the prediction of stability in dynamic movements as pregnancy advances.

Introduction

Behind vehicular accidents, falls are the second most common form of accidental traumatic injury in pregnant women, as approximately one in four women experience a fall while pregnant (Dunning, LeMasters, & Bhattacharya, 2010). Severe falls can negatively impact maternal and/or fetal health outcomes, such as pre-term labor, cesarean delivery, placental abruption, fetal distress, fetal hypoxia, or fetal death (Schiff, 2008). Yet, a fall risk assessment is not typically involved in obstetric care (Heafner et al., 2013), leaving the risk of falls in individual pregnant women overlooked. Incorporating measures of postural stability throughout pregnancy could add to the comprehension of the risks and modalities of falls in pregnant women. Therefore, it is imperative to assess balance mechanics related to pregnancy, and standardize evidence-based fall risk assessment tools, to mitigate these risks.

During pregnancy, the body undergoes numerous adaptations that could contribute to postural instability or poor balance control (Cakmak, Ribeiro, & Inanir, 2016; Inanir, Cakmak, Hisim, & Demirturk, 2014). Morphologically, gestational weight gain (GWG) primarily accumulates anteriorly about the torso, resulting in anterior shift of the center of mass (COM) and an increased upper body torque about the hip (Whitcome et al., 2007). Hormonally, fluctuations of progesterone, relaxin, estrogen, and cortisol are believed to correlate to increased ligament laxity (Cherni et al., 2019; Talbot & Maclennan, 2016), contributing to swelling in the extremities (Talbot & Maclennan, 2016), muscle weakness and pain in the lower back and pelvic region (Mogren & Pohjanen, 2005), the joints of the lower extremities (Vullo, Richardson, & Hurvitz, 1996), and the feet (Karadag-Saygi, Unlu-Ozkan, & Basgul, 2010). Pain throughout pregnancy can hinder neuromuscular control (Hlaing, Puntumetakul, Wanpen, & Boucaut, 2020) and proprioception of joint position (Ruhe, Fejer, & Walker, 2011). Additionally, alterations to other sensory systems responsible for maintaining balance, such as the visual and audio-vestibular systems, can be affected by hormonal changes (Black, 2002; Fast et al., 1987; Preetha & Solomon, 2011; Samra, 2013; Serna-Hoyos et al., 2022; Vullo, Richardson, & Hurvitz, 1996), resulting in further instability. Consequently, diminished postural stability throughout pregnancy could result in fall; thus, making measurements of postural stability common indirect measures of fall risks.

Center of pressure (COP) measurements has been frequently used to quantify postural stability during pregnancy in previous literature (Butler, Colón, Druzen, & Rose, 2006; Cakmak, Ribeiro, & Inanir, 2016; Opala-Berdzik et al., 2015). Compared to nulligravida women, pregnant women displayed increased sway in the anteroposterior and mediolateral directions, suggesting that stability significantly decreases into the third trimester (Butler, Colón, Druzen, & Rose, 2006; Oliveira, Simpson, & Nadal, 1996). However, to combat decreases in stability, pregnant women have been observed to employ a wider stance width to mitigate stability issues in the mediolateral direction (Jang et al., 2008). In addition, pregnant women displayed greater sway magnitudes while standing with eyes closed, suggesting that they relied heavily on visual cues to maintain balance (Butler, 2006). Interestingly, most studies focused on bilateral standing measurements (Butler, Colón, Druzen, & Rose, 2006; Cakmak, Ribeiro, & Inanir, 2016; Inanir, Cakmak, Hisim, & Demirturk, 2014; Jang, Hsiao, & Hsiao-Wecksler, 2008b; Nagai et al., 2009; Oliveira, Simpson, & Nadal, 1996; Opala-Berdzik et al., 2015; Sunaga, Kanemura, Anan, Takahashi, & Shinkoda, 2016), despite the single-limb stance test being an acceptable balance assessment tool in clinical settings for other high fall risk groups (i.e. the elderly, patients with chronic stroke, Parkinson's disease) (Chomiak, Pereira, & Hu, 2015). Until recently, balance assessments measuring single-limb stance had not been investigated in a laboratory setting. Bagwell and colleagues (2022) were the first to compare postural stability differences between nulligravida and pregnant women at three stages (second trimester, third trimester, and four to six months postpartum) during single-limb stance tasks, with and without visual input. They revealed that women in the third trimester exhibited decreased sway magnitude and sway velocity, but increased median frequency and repetitive postural responses, which they deduced to represent a protective strategy or inability to flexibly adapt to a perturbation (Bagwell et al., 2022). Although the results from Bagwell are insightful, a longitudinal study comparing balance measurements in both bilateral and single-limb stance has yet to be conducted.

A longitudinal study design utilizing pregnant women posits unique challenges compared to other populations due to the ethical and logistical difficulty of recruiting women prior to pregnancy, in combination with the confounding hormonal influences on balance, the use of nulliparous women with simulated GWG may offer a solution. Admittedly, performance measurements taken from nulliparous women under simulated pregnancy conditions may only determine the mechanical adaptations that are related to GWG (Ogamba, Loverro, Laudicina, Gill, & Lewis, 2016). However, ruling out the intrinsic influences of pregnancy allows for a distinguished examination of a single contributor that is interrelated with other fall risks factors in pregnant women remains beneficial. Therefore, the purpose of this study is to examine potential postural stability mechanics associated with the mechanical influences of simulated GWG in bilateral and single-limb stance. It was hypothesized that in bilateral standing, as anterior load is increased, the overall sway magnitude and sway in the anteroposterior direction would increase (Butler, Colón, Druzen, & Rose, 2006; Oliveira, Simpson, & Nadal, 1996); and, in single-limb stance, overall sway magnitude and sway direction would decrease (Bagwell et al., 2022), thus suggesting reduced postural stability.

Methods

Participants

An *a priori* power analysis was conducted on G*Power (version 3.1, Universität Kiel, Germany), with data from Bagwell and colleagues (2022). Based on a proposed effects size of 1.08, power of 0.95, and alpha (α) of 0.05, it was determined that 11 participants were required to achieve adequate statistical power. A total of seventeen nulliparous women were recruited for this study in anticipation of potential participant drop-out; six participants were removed from analyses due to drop-out or insufficient data. The eleven nulliparous women (24.36 ± 4.20 years; 1.63 ± 0.06 m; 59.11 ± 7.46 kg; 22.06 ± 1.58 kg/m^2) analyzed in this study were required to be between the ages of 18 and 34, within “normal” body mass index (BMI; 18.5 – 24.9), and free of lower limb injuries. Prior to completing any study-related tasks, written informed consent was obtained on institutional approved documentation (Protocol No: 1727598-9) and in accordance with 1964 Declaration of Helsinki.

Procedures

Data collections were conducted on four separate days, with a minimum of two days between visits to avoid fatigue as a confounding factor (Cheung, Hume, & Maxwell, 2003) and a maximum of four weeks between visits, to avoid potential fluctuation in body mass and loss of retention in participants. Baseline (B) measurements were obtained during the initial data collection, wherein age, height, mass, and BMI of participants were measured and recorded. At each subsequent collection, participants were fitted with an adjustable Velcro-secured, plate-loaded, weight-vest with an additional mass secured anteriorly to be equivalent to the mass of three months pregnant at the first trimester (1T); added mass equivalent to six months pregnant at the second trimester (2T); and added mass equivalent to nine months pregnant at the third trimester (3T). It is worth noting that baseline tasks were performed while wearing the weight-vest without additional mass, which added an additional three kilograms to participants' body mass, but did not interfere with the BMI classification of the participants. Additional mass magnitude for simulated GWG was determined by participants' baseline BMI in order to simulate body mass at the third, sixth, and ninth months of pregnancy based on the mass gain recommendations by the Institute of Medicine (IOM) (National Research Council, 2010) (for example, a woman with a height of 1.62m, body mass of 59kg, and BMI of 22.5 kg/m^2 would have the following mass amounts added upon separate visits: 2.0, 9.2, and 15.9 kg to simulate months 3, 6, and 9 of pregnancy, respectively). Participants warmed up for 10 minutes at a self-selected, comfortable walking speed on a treadmill (Tracmaster TMX425, Newton, KS, USA) while wearing the weight-vest to acutely adapt to the additional anterior load.

Participants then performed balance tasks in two separate tasks, 1) bilateral standing (BL); and 2) single-limb stance (SLS) on both the left and right limbs, separately. Three trials were collected for each condition in 20 second intervals, for a total of nine trials. During all balance tasks, participants were instructed to stand quietly on a single force platform, with the hands placed over the stomach and eyes open and looking straight ahead, but not focused on a particular target. Participant stance width was controlled in the BL tasks by instructing the participants to place their feet on the same marked placement for each trial. Center of pressure (COP) data were obtained with a single force platform (1,000 Hz, Advanced Mechanical Technology Inc., MA, USA), embedded in the floor.

Data Reduction

COP coordinate data were exported from Vicon Nexus v2.15.1 and imported into MATLAB (The Mathworks, Inc., Natick, MA, USA) to be filtered with a low-pass Butterworth digital filter (12.5 Hz; (Callahan, 2017). Posturograms were created in both BL and SLS conditions, created by the center of pressure excursion area (COPEa). COPEa is defined by the absolute sway magnitude in the maximum and minimum medial-lateral (X) and anterior-posterior (Y) coordinate data from the equation: $COPE_a = (X_{max} - X_{min}) \times (Y_{max} - Y_{min})$ in mm^2 (Callahan, 2017). Additionally, the absolute sway magnitude in the anteroposterior (AP) and mediolateral (ML) directions were analyzed for both tasks. All postural sway variables were analyzed separately between the bilateral and single-limb tasks across trimester conditions, consisting of: bilateral center of pressure excursion area (BCOPEa), bilateral anteroposterior sway magnitude (BAP), bilateral mediolateral sway magnitude (BML), single-limb stance center of pressure excursion area (SLS-COPEa), single-limb anteroposterior sway magnitude (SLS-AP) and single-limb mediolateral sway magnitude (SLS-ML).

Statistical Analysis

Mean (SD) values were calculated for all variables. An independent one-way analysis of variance (ANOVA; α=0.05) was conducted to compare subject mass with the additional mass (simulated pregnancy body mass) at each visit (B, 1T, 2T, 3T). If a significant difference was detected in the omnibus ANOVA test, pairwise comparisons were interpreted after applying the Sidak adjustment.

Paired t-tests (α=0.05) were used to compare left and right limb COP values. Left and right limb variables were collapsed into SLS condition values, as there were no significant differences between limbs ($p > 0.05$).

Separate repeated measures tests were utilized to compare individual bilateral and single-limb postural sway variables. Data normality was evaluated for each variable using the Shapiro-Wilk test. Normally distributed data were assessed with individual one-way repeated-measures ANOVA tests. If Mauchly's test of sphericity was greater than 0.05, then sphericity was assumed. If data are not normally distributed, individual Friedman's tests were used for comparison. If statistical significance was detected, pairwise comparisons were performed for multiple comparisons between visits with a Holm-Bonferroni method adjustment to the *p* values. The Holm-Bonferroni method reduces the possibility of type I error and controls for familywise error rates among multiple test comparisons by ranking the *p* values and adjusting the significance level based on the number of comparisons (Holm, 1979).

Results

One-way ANOVA results revealed significant differences in simulated pregnancy body mass among visits, $F_{(3, 40)} = 12.01$, $p < 0.001, \eta^2 = 0.47$. **Table 3** displays the means and standard deviations of simulated pregnancy body mass among visits.

Table 3. Means and standard deviation values for simulated pregnancy body mass

	Baseline	First Trimester	Second Trimester	Third Trimester
Mass (kg)	62.59 ± 7.49 [a, b]	65.17 ± 7.51 [c]	73.64 ± 7.62 [a]	79.75 ± 7.53 [b, c]

Statistical significance is denoted by **a** $p < 0.05$, **b** $p < 0.001$, and **c** $p < 0.001$

Results from the Friedman test revealed statistical differences in BML measures across visits ($\chi^2(3) = 8.13, p = 0.04$). Initially, post hoc pairwise comparisons using the Wilcoxon test revealed that B was significantly different from 1T ($p = 0.03$), however, after applying the Holm-Bonferroni adjustment (adjusted $p = 0.017$), the null hypothesis was not rejected. **Figure 7** displays the median value and range of the BML sway magnitude across visits.

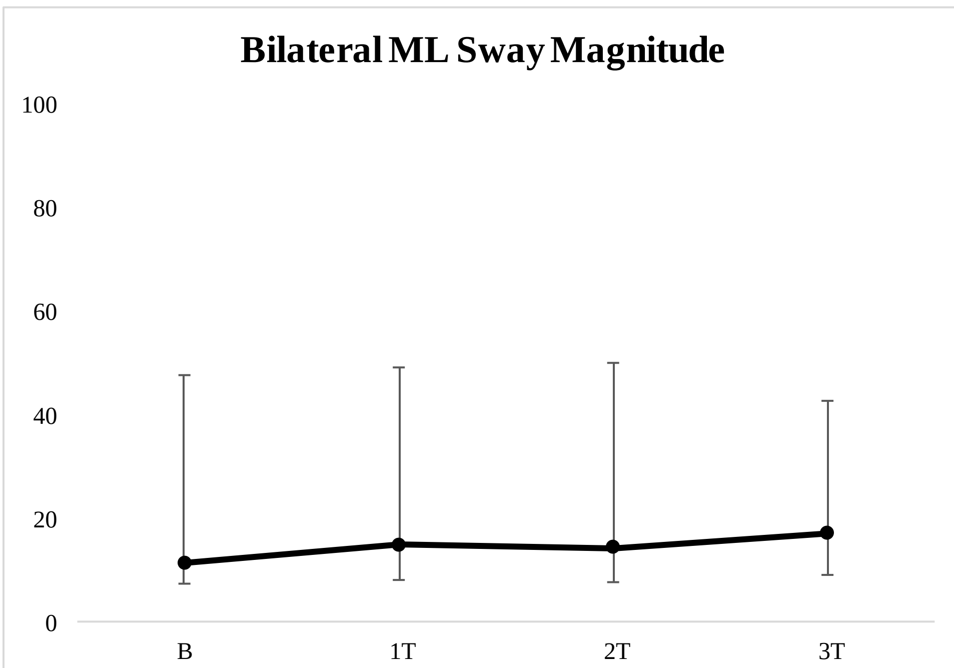

Figure 7. Bilateral mediolateral sway magnitude medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

The Friedman test revealed statistical differences in BAP measures across visits ($\chi^2(3) = 8.35, p = 0.04$). Initially, post hoc pairwise comparisons using the Wilcoxon test revealed that B was significantly different from 1T (p = 0.02), B was significantly different from 2T (p = 0.008), and B was significantly different from 3T (p = 0.02). However, following the Holm-Bonferroni adjustment (adjusted p = 0.0083), significance was only detected between B and 2T; all others were not significant. **Figure 8** displays the median value and range of the BAP sway magnitude across visits. The Friedman test revealed that BCOPEa was not statistically different across visits ($\chi^2(3) = 7.26, p = 0.06$). **Figure 9** displays the median value and range of the BCOPEa across visits.

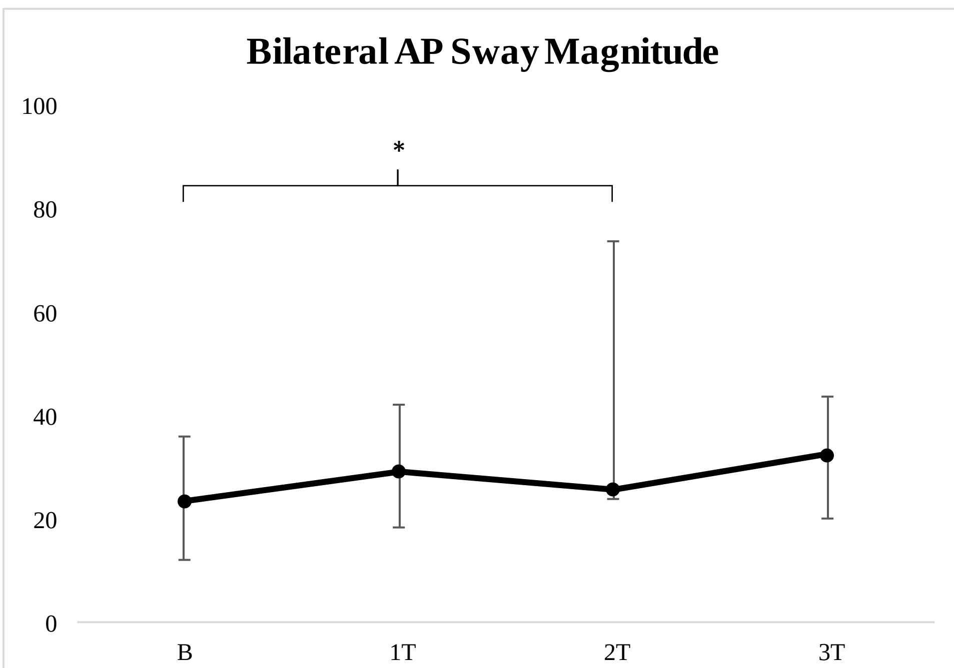

Figure 8. Bilateral anteroposterior sway magnitude medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.
* indicates p < 0.01

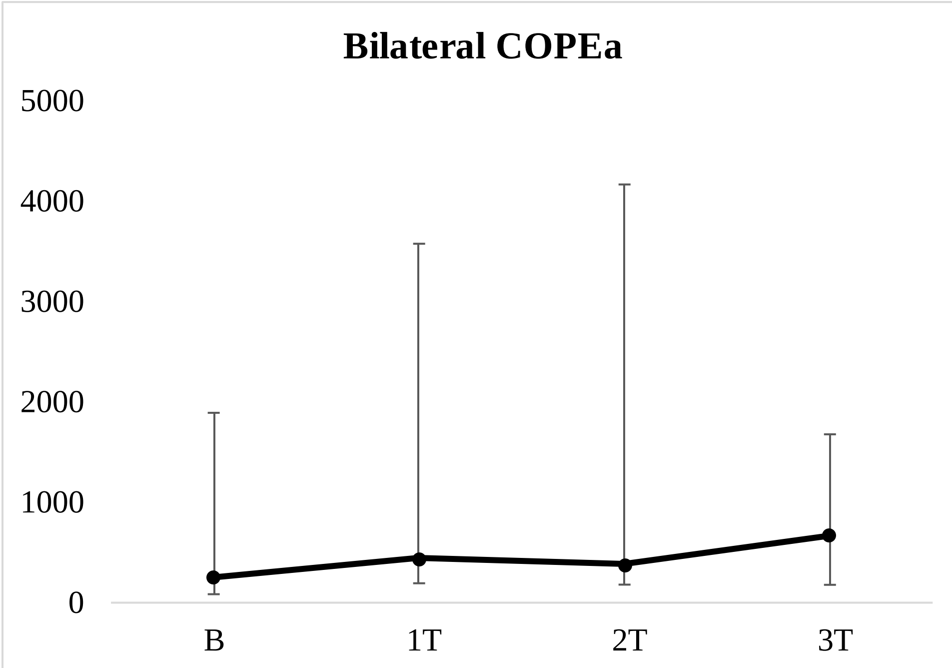

Figure 9. Bilateral center of pressure excursion area medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and 3rd trimester (3T) visits.

The repeated measures ANOVA test revealed no significant differences to SLS-ML among visits, $F_{(1.68, 16.77)} = 0.58$, $p = 0.92$, $\eta^2 = 0.01$. **Figure 10** displays the mean and standard deviation of the SLS-ML sway magnitude across visits. Results revealed no significant differences to SLS-AP ($\chi^2(3) = 1.47, p = 0.69$), nor SLS-COPEa ($\chi^2(3) = 0.05, p = 0.99$) among visits. **Figure 11 & Figure 12** display the median value and range of the SLS-AP sway magnitude and SLS-COPEa across visits, respectively.

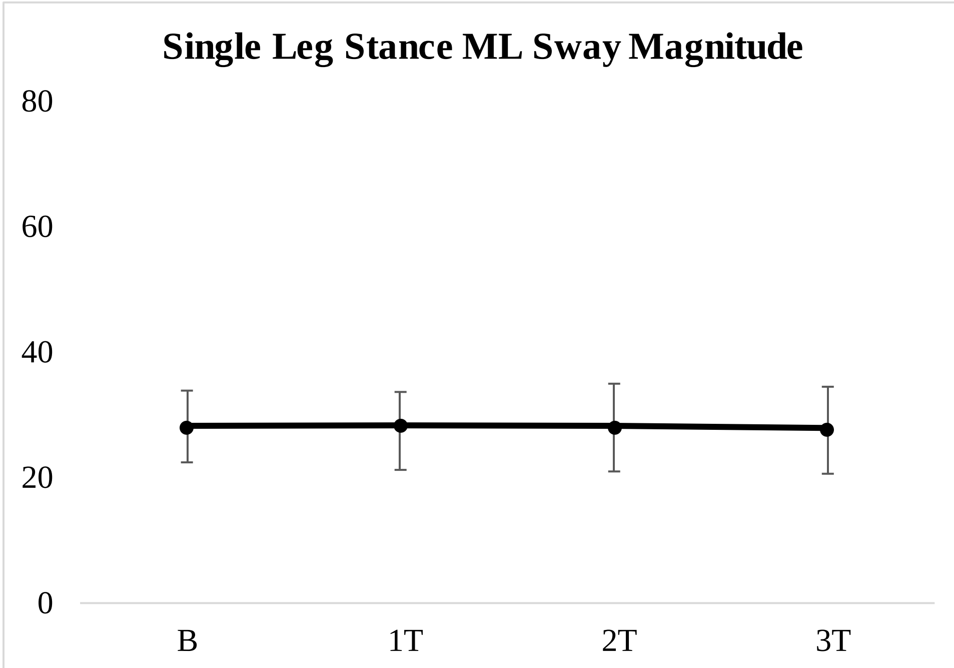

Figure 10. Single-limb stance mediolateral sway magnitude means and standard deviations across baseline (B), 1st trimester (1T), 2nd trimester (2T) and 3rd trimester (3T) visits.

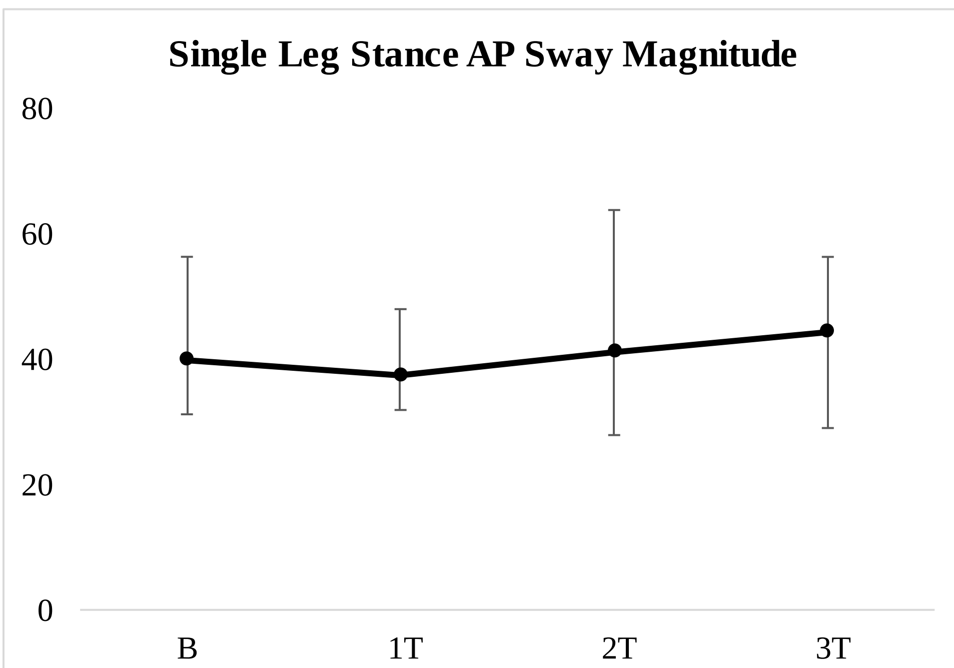

Figure 11. Single-limb stance anteroposterior sway magnitude medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and 3rd trimester (3T) visits.

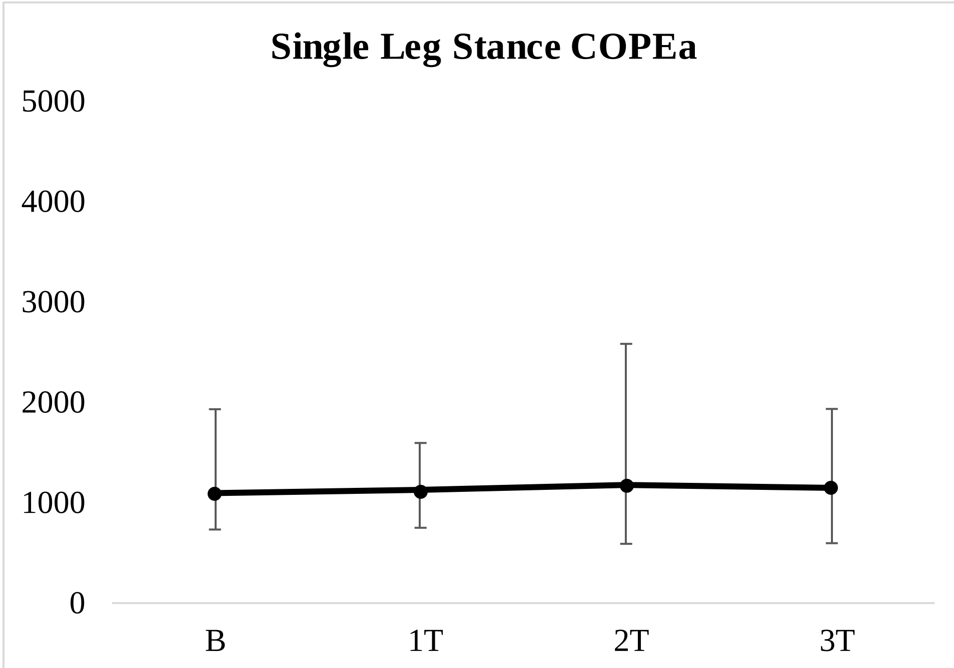

Figure 12. Single-limb stance center of pressure excursion area medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and 3rd trimester (3T) visits.

Discussion

The purpose of this study was to examine postural stability mechanics related to the mechanical influences of simulated gestational weight gain (GWG) on bilateral and single-limb balance. It was hypothesized that as anterior load increased at each visit (B, 1T, 2T, 3T), the overall center of pressure sway magnitude area (COPEa) and sway magnitude in the anteroposterior (AP) direction would increase in bilateral standing; yet conversely was expected to decrease in COPEa and sway magnitude in the single leg stance (SLS) task. The outcomes from this study partially supported the hypotheses as the bilateral tasks displayed general significant differences, however, significance was not specific across visits for all bilateral variables. Additionally, SLS performance remained consistent throughout each condition rather than decreasing as expected.

Previous research examining bilateral balance control in pregnant women generally determined that as pregnancy advances, postural stability declines (Butler, Colón, Druzen, & Rose, 2006; Carvalho et al., 2019; Danna-Dos-Santos et al., 2018; Jang, Hsiao, & Hsiao-Wecksler, 2008a). Butler and colleagues (2006) found that overall postural sway increased significantly in 2T and 3T with both eyes open (EO) and eyes closed (EC) compared to nulligravid controls, which was similar to findings from Carvalho and colleagues (2019) wherein pregnant women had significantly greater overall sway with EO and lower back pain. The outcomes from this study were not in agreement with previous findings as overall balance was not affected by increasing simulated GWG. However, the current study used a repeated measures approach with the same participants for all measurements, rather than comparisons to a control group. Interestingly, results from the current study aligned with findings from Opala-Berdzik and colleagues (2015) wherein overall sway remained statistically similar, but AP sway magnitude increased in the last two trimesters. Additionally, Jang and colleagues (2008) discovered that AP sway magnitude increased as pregnancy progressed, but ML sway magnitude remained similar, which they posited was correlated to women increasing their stance width in later trimesters. However, Opala-Berdzik and colleagues, despite having similar findings of increased stance width, detected only a weak correlation between increased stance width and AP sway magnitude. In the current study, stance width was controlled for across participants and condition which may explain only detecting significant differences in bilateral AP sway magnitude between B and 2T conditions. Furthermore, it is also possible that the participants in the current study utilized a pelvic tilt strategy to accommodate for the additional mass added to their torso, a strategy described by Danna-Dos Santos and colleagues (2018), in 3T women had increased pelvic tilt that relocated the COP closer to the longitudinal axis but still anterior to the ankle joint. This could explain the significance observed between B and 2T, but not observed between B and 3T, if participants altered their pelvic tilt; however, pelvic tilt was not measured in the current study.

Single limb balance measurements throughout pregnancy have not been researched as thoroughly as bilateral balance, despite SLS having clinical applications as a screening tool for balance disfunction (Balogun, Ajayl, & Alawale, 1997). Empirically, Bagwell and colleagues (2022) were the only researchers to examine SLS throughout pregnancy, determining that SLS sway decreased in 2T and 3T compared to nulligravida controls. The results from the current study did not align with these findings, wherein overall sway area and sway magnitude in AP and ML directions remained similar across conditions. The lack of similarity could be related to the current study using the same participants for all measurements rather than comparing a pregnant group to a nulligravida control group or the use of different, albeit related, COP measurement techniques to determine postural stability. Additionally, all participants had the same amount of mass added at each simulated trimester, whereas the actual pregnant women analyzed by Bagwell and colleagues had varying GWG.

To the authors' knowledge, this is the first study to use a repeated measures approach for simulated GWG that is based on recommendations by the IOM for normal BMI women. Previous literature examining balance in pregnant women did not account for variations in GWG among individual participants, however the reported average BMI for pregnant women remained within the recommended parameters in some cases (Bagwell et al., 2022; Jang, Hsiao, & Hsiao-Wecksler, 2008b; Opala-Berdzik et al., 2015). Interestingly, some authors reported no association between increasing GWG during pregnancy and postural behavior (Butler, Colón, Druzen, & Rose, 2006; Danna-Dos-Santos et al., 2018), which was in disagreement with some evidence that women with an overweight BMI exhibited decreased sway amplitudes, thus favorable postural behavior compared to normal BMI women (Błaszczyk, Janusz W., Cieślinska-Świder, Plewa, Zahorska-Markiewicz, & Markiewicz, 2009; Whitcome, Shapiro, & Lieberman, 2007). Therefore, the justification for using nulliparous women with simulated GWG was to control for the amount of GWG and to rule out other physiological factors related to pregnancy that may otherwise influence postural behavior measured by COP variables. While the findings from the current study have some similarities to previous pregnancy research, the only significant result was in bilateral AP sway magnitude between baseline and 2T, which may support the conclusion that increasing GWG may not influence postural behavior alone. On the other hand, it is possible that participants accommodated for the imbalanced anterior load in a manner not detected by COP measurements, such as tilting at the pelvis to create a postural correction similar to pregnant women (Whitcome, Shapiro, & Lieberman, 2007).

In conclusion, the findings from this study could suggest that increasing body mass alone may not be responsible for decreased static stability. It may also be viable that normal BMI women gaining within the recommended GWG parameters may exhibit favorable postural behavior. Future research should examine the kinematic strategies of the pelvis in congruence with COP measurements to determine if a postural strategy is being overlooked. While this study aimed to rule out confounding factors, some limitations to the study exist. The current study excluded pregnant women for comparison. Additionally, only acute effects of the added mass could be determined due to the nature of the study.

CHAPTER 3

Dynamic Stability During Gait Initiation in Nulliparous Women with Simulated Gestational Weight Gain

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Dynamic Stability During Gait Initiation in Nulliparous Women with Simulated Gestational Weight Gain

Significance of Chapter

This study will compare dynamic stability during gait initiation across simulated gestational weight gain (GWG).

Introduction

Gait initiation is the functional task of transitioning from static standing to steady gait, requiring forward momentum development while maintaining dynamic postural control (Yiou, Caderby, Delafontaine, Fourcade, & Honeine, 2017). Static standing requires postural control to achieve equilibrium, which refers to the ability to maintain the vertical projection of the body's center of mass (COM) within the bounds of the base of support (BOS) created when both feet are in contact with the ground (Hof, Gazendam, & Sinke, 2005). When a perturbation threatens the maintenance of balance, one of two postural control strategies are utilized as a response: anticipatory or compensatory (Duarte et al., 2022). In healthy adults, gait initiation creates a perturbation to static balance that is preemptively anticipated when the center of pressure (COP) shifts posteriorly and towards the swing-limb and the COM propels anteriorly and towards the stance-limb (Caderby et al., 2017), causing the anticipatory postural adjustments (APA) to respond by recruiting skeletal muscles to stabilize the whole-body movement during step execution (Yiou, Caderby, Delafontaine, Fourcade, & Honeine, 2017). Subsequently, lifting the foot of the swing-limb off the ground dramatically decreases the BOS area while the COM is being accelerated laterally, which could result in a gap between the COP and COM (Caderby et al., 2017). If the swing foot lands prior to the COM travelling too far laterally, then the BOS is instantly increased, balance can be restored, and the COM propels forward; however, if the swing foot does not land in sufficient time, or the COM travels significantly in the lateral direction, then a fall can occur (Yiou, Caderby, Delafontaine, Fourcade, & Honeine, 2017). Additionally, because COM position during gait initiation is mobile around a shifting BOS—as opposed to within a fixed BOS during static standing—measuring COP sway magnitudes no longer have the same implications (Yiou, Caderby, Delafontaine, Fourcade, & Honeine, 2017). Instead, the mediolateral gap created by the relationship between the COP and COM, relative to the altered BOS, can quantify the margin of stability (MOS) and reflect dynamic stability (Hof, Gazendam, & Sinke, 2005).

Gait initiation can assess dynamic postural control with the consideration of the central nervous system (CNS), because the CNS uses APA initiated by sensory feedback signals, thus making gait initiation a relevant tool to assess dynamic stability and predict fall risks in populations with pathological gait or movement disorders (Yiou, Caderby, Delafontaine, Fourcade, & Honeine, 2017). Gait initiation can present a notable challenge for the elderly (Azizah Mbourou, Lajoie, & Teasdale, 2003; Delval, Tard, & Defebvre, 2014; Mickelborough, Van Der Linden, Tallis, & Ennos, 2004), individuals with Parkinson's Disease (Amano et al., 2013; Delval, Tard, & Defebvre, 2014; Hass, Waddell, Fleming, Juncos, & Gregor, 2005), patients with chronic stroke (Gama, Celestino, Barela, & Barela, 2019; Sousa, Silva, & Santos, 2015), and children with cerebral palsy (Stackhouse et al., 2007; Wallard, Dietrich, Kerlirzin, & Bredin, 2014). In some instances, these groups performed similarly, wherein the elderly used similar movement patterns during gait initiation, displaying decreased movement time, amplitude, and velocity throughout the duration of the task (Dibble et al., 2004). Presumably, performance similarities in groups with movement pathologies are related to impairments to the visual, audio-vestibular, and somatosensory systems, which are responsible for maintaining postural stability (Peterka, 2018). For example, the elderly can experience age-related sensory degeneration, including: visual impairments (Peterka, 2018; Schneck & Haegerström-Portnoy, 2003), which can limit input of critical information for postural adaptations to different environments (Patla, 1997); vestibular dysfunction (Iwasaki & Yamasoba, 2015), which can hinder the anticipatory pre-step movements (Sasaki, Asawa, Katsuno, Usami, & Taguchi, 2001); and, proprioceptive deficits (Goble, Coxon, Wenderoth, Van Impe, & Swinnen, 2009), which can hinder the control of APAs and postural regulation (Ruget, Blouin, Coyle, & Mouchnino, 2010), all of which can greatly increase the risk of falls.

Surprisingly, gait initiation is not used extensively to measure dynamic stability in other groups with high fall risks, despite some speculation that it may predict fall risks in the elderly (Azizah Mbourou, Lajoie, & Teasdale, 2003). Pregnant women in particular could benefit from gait initiation measurements since their postural control mechanisms have similarities to those of the elderly (Sunaga, Takahashi, Anan, & Shinkoda, 2020). Both the elderly and pregnant women walk with similar “cautious” strategies to improve sense of stability, including walking with a reduced velocity, and increased double-limb support time (Gottschall, Sheehan, & Downs, 2013; Winter, Patla, Frank, & Walt, 1990). In addition, pregnancy-related adaptations can elicit or exacerbate dysfunction to the visual, vestibular, and somatosensory systems (Peterka, 2018), which can further affect postural stability. More importantly, pregnant women gain roughly twenty-five percent of their pre-pregnancy body mass unevenly and anteriorly about the torso (Hagan & Wong, 2010), which may greatly impact the shift in COP and propulsion of COM during gait initiation, and potentially increase the risk of falling while performing this task. Reportedly, the fall rates of pregnant women are even comparable to the elderly, with a high risk of requiring hospitalization post injury (Dunning, LeMasters, & Bhattacharya, 2010), making fall risk assessments imperative in the pregnant population.

Despite the potentially useful applications of measuring both gait initiation and margin of stability (MOS) in pregnant women, there remains a lack of research regarding both concepts. This gap may be related to the difficulty of measuring full-body COM (Catena, R. D., Connolly, McGeorge, & Campbell, 2018; Sunaga, Takahashi, Anan, & Shinkoda, 2020), which is ideal for quantifying APAs and MOS. Additionally, unlike other populations with pathological movement patterns, pregnant women can only be observed for a finite period of time, which can make longitudinal research in this population difficult. Thus, in order utilize a longitudinal study design, control for confounding hormonal factors, and minimize fall risks to pregnant women, the use of nulliparous women with simulated gestational weight gain (GWG) is being proposed in the current study. While findings from simulated GWG cannot translate outside of the mechanical adaptations of pregnancy, it can determine the acute impact that added mass may have on dynamic stability in women. Therefore, the purpose of this study is to examine dynamic balance mechanics during gait initiation associated with the mechanical influences of simulated GWG. It was hypothesized that with increasing simulated GWG, dynamic stability would decrease, characterized by an increase in duration of time spent in the APA phase, decreased MOS outcomes, and increased step width at gait initiation across visits.

Methods

Participants

An *a priori* power analysis was conducted on G*Power (version 3.1, Universität Kiel, Germany), with data from Qu and colleagues (2021). Based on a proposed effects size of 1.23, power of 0.95, and alpha (α) of 0.05, it was determined that 11 participants were required to achieve adequate statistical power. A total of seventeen nulliparous women were recruited for this study in anticipation of participant drop-out; six participants were removed from analyses due to drop-out or insufficient data. The eleven nulliparous women (25.45 ± 4.13 years; 1.65 ± 0.07 m; 60.25 ± 6.53 kg; 22.08 ± 1.73 kg/m^2) analyzed in this study were required to be between the ages of 18 and 34, within "normal" body mass index (BMI; 18.5 – 24.9), and free of lower limb injuries. Prior to completing any study-related tasks, written informed consent was obtained on institutional approved documentation (Protocol No: 1727598-9) and in accordance with 1964 Declaration of Helsinki.

Procedures

Data collections were conducted on four separate days, with a minimum of two days between visits to avoid fatigue as a confounding factor (Cheung, Hume, & Maxwell, 2003) and a maximum of four weeks between visits, to avoid potential fluctuation in body mass and loss of retention in participants. Baseline (B) measurements were obtained during the initial data collection, wherein age, height, mass, and body mass index (BMI) of participants were measured and recorded. At each subsequent collection, participants were fitted with an adjustable Velcro-secured, plate-loaded, weight-vest with an additional mass secured anteriorly to be equivalent to the mass at the terminal end of three months pregnant at the first trimester (1T); added mass equivalent to the end of six months pregnant at the second trimester (2T); and added mass equivalent to the end of nine months pregnant at the third trimester (3T). It is worth noting that baseline tasks were performed while wearing the weight-vest without additional mass, which added an additional three kilograms to participants' body mass on average, but did not interfere with the BMI classification of the participants. Additional mass magnitude for simulated GWG was determined by participants' baseline BMI in order to simulate body mass at the third, sixth, and ninth months of pregnancy based on the mass gain recommendations by the Institute of Medicine (IOM) (National Research Council, 2010) (for example, a woman with a height of 1.62m, body mass of 59kg, and BMI of 22.5 kg/m^2 would have the following mass amounts added upon separate visits: 2.0, 9.2, and 15.9 kg to simulate months 3, 6, and 9 of pregnancy, respectively).

Participants warmed up for 10 minutes at a self-selected, comfortable walking speed on a treadmill (Tracmaster TMX425, Newton, KS, USA) while wearing the weight-vest to acutely adapt to the additional anterior load. Then, a total of sixty-four retroreflective markers were adhered to the head, trunk, upper extremities, pelvis, lower extremities, and feet to track movements of the respective body sections (full marker set is detailed in **Appendix A)**. Five additional markers were placed on the weight plate to measure center of mass (COM) location on simulated pregnancy visits.

For gait initiation trials, participants were instructed to stand quietly with their arms at their sides, in the center of a force platform, with a self-selected width apart between feet. Participants were allowed to stand at their preferred stance width to determine if the increasing load would naturally alter their stance. Once cued with a verbal "walk" instruction, participants began walking forward continuously for five meters at a self-selected pace. Data collection began ten seconds prior to the cue to walk to capture anticipatory postural adjustments. After three practice trials, participants repeated the task from the same marked spot in the center of the first force plate for 12 trials. During all trials, marker trajectories were tracked using a 10-camera three-dimensional motion capture system (200 Hz, Vicon Motion Systems, Ltd., Oxford, UK). Although center of pressure and force data were collected via the force platforms, this data was not utilized for the current study.

Data Reduction

All raw kinematic variables were exported from Vicon Nexus and variables of interest were computed and filtered with low-pass Butterworth digital filters at cutoff frequencies of 6 Hz in Visual 3D software (C-Motion, Inc., Germantown, MD, USA). A thirteen-segment model was constructed from marker trajectories, including the head, trunk, pelvis, bilateral upper arm, forearm, thigh, leg, and foot segments. An additional segment was created by the weight plate during simulated pregnancy visits. Outcome variables included duration in three specific phases of the gait initiation movement, which were determined by launch velocity: 1) Anticipatory Postural Adjustments (APA), between gait onset defined by the instantaneous mediolateral acceleration of the COM and leading foot heel-off; 2) Foot Lift (LIFT), between leading limb heel-off and leading limb foot-off; and 3) Step Execution (STEP), between leading limb foot-off and leading limb heel contact. Additionally, dynamic stability was determined by the MOS at the time of heel contact of the leading foot (i.e. in STEP). MOS is the distance between the of the base of support (BOS) and the position of the extrapolated COM (xCOM) (Hof, Gazendam, & Sinke, 2005). To calculate xCOM, COM position was determined from the 13-segment model COM at baseline, and from the 13-segment model COM in combination with the plate COM, determined by the kinematics of the additional plate markers, for all simulated pregnancy visits. Additionally, COM velocity was determined by the first derivative of the COM position. Extrapolated COM (xCOM) was then calculated from the formula below:

$$xCOM = P_{COM} + \frac{V_{COM}}{\sqrt{g/l}}$$

Where P_{COM} and V_{COM} represent the position and velocity of the COM, respectively; g represents the gravitational constant, and l represents the leg length multiplied by 1.2 (leg length was determined by the distance between the greater trochanter and lateral malleolus in the calibration trial) (Hof, Gazendam, & Sinke, 2005; Young, Wilken, & Dingwell, 2012). MOS is then calculated as:

$$MOS = xCOM - BOS$$

An adaptation of the methods from Young and colleagues (2012) will be used to identify BOS boundaries as the mediolateral BOS (BOS_{ML}) and anteroposterior BOS (BOS_{AP}). BOS_{ML} will be determined by the fifth proximal metatarsal marker of the leading foot and BOS_{AP} will be determined by the first distal phalange marker on the leading foot (**Figure 13**).

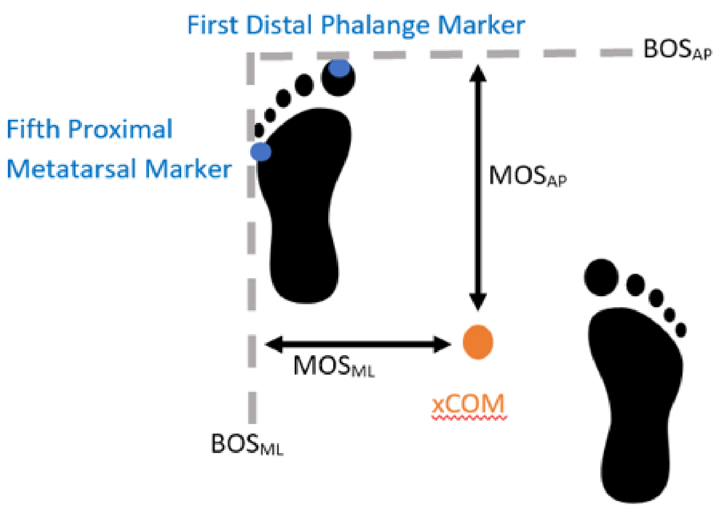

Figure 13. Margin of stability (MOS), adapted from Young and colleagues (2012), determined by extrapolated center of mass (xCOM) within the base of support (BOS)

A positive MOS outcome corresponds to the xCOM being within the bounds of the BOS, indicating dynamic stability; whereas a negative MOS outcome indicates dynamic instability and implies the need for an additional corrective step to maintain balance (Caderby, Yiou, Peyrot, Begon, & Dalleau, 2014).

In addition, both step length (SL) and step width (SW) were measured at the instance of the leading heel strike by the lateral distance between heel markers and by the anteroposterior distances between heel markers, respectively.

Statistical Analysis

Statistical analyses were conducted using SPSS 29 (IBM, NY). Mean (SD) values were calculated for all variables. An independent one-way analysis of variance (ANOVA; α=0.05) was conducted to compare subject mass with the additional mass (simulated pregnancy body mass) at each visit (B, 1T, 2T, 3T). If a significant difference was detected in the omnibus ANOVA test, pairwise comparisons were interpreted after applying the Sidak adjustment.

Separate repeated measures tests were utilized to compare all variables of interest, including: duration of time in gait initiation phases (APA, LIFT, STEP), margin of stability (MOS_{AP}, MOS_{ML}), and step characteristics (SL, SW). Data normality was evaluated for each variable using the Shapiro-Wilk test. Normally distributed data were assessed with individual one-way repeated-measures ANOVA tests. If Mauchly's test of sphericity was violated, then sphericity was assumed. If data were not normally distributed, individual Friedman's tests were used for comparisons. If statistical significance was detected, pairwise comparisons were performed for multiple comparisons between visits, with a Holm-Bonferroni method adjustment to the *p* values to control for family-wise error rates among multiple tests (Holm, 1979). (Abdi, 2010)

Results

One-way ANOVA results revealed significant differences in simulated pregnancy body mass among visits, $F_{(3, 40)} = 15.65$, $p < 0.001$, $\eta^2 = 0.54$. **Table 4** displays the means and standard deviations of simulated pregnancy body mass among visits.

Table 4. Means and standard deviation values for simulated pregnancy body mass.

	Baseline	First Trimester	Second Trimester	Third Trimester
Mass (kg)	63.71 ± 6.60 $^{\mathbf{a,\,b}}$	66.32 ± 6.54 $^{\mathbf{c}}$	74.95 ± 6.49 $^{\mathbf{a}}$	80.81 ± 6.78 $^{\mathbf{b,\,c}}$

Statistical significance is denoted by **a** $p < 0.05$, **b** $p < 0.001$, and **c** $p < 0.001$.

The Friedman tests revealed no statistical differences in gait initiation phases for APA ($\chi^2(3) = 6.16, p = 0.10$), LIFT ($\chi^2(3) = 5.84, p = 0.12$), nor STEP ($\chi^2(3) = 2.45, p = 0.48$) across visits. **Figure 14, Figure 15, & Figure 16** display the medians and ranges of APA, LIFT, and STEP across visits, respectively.

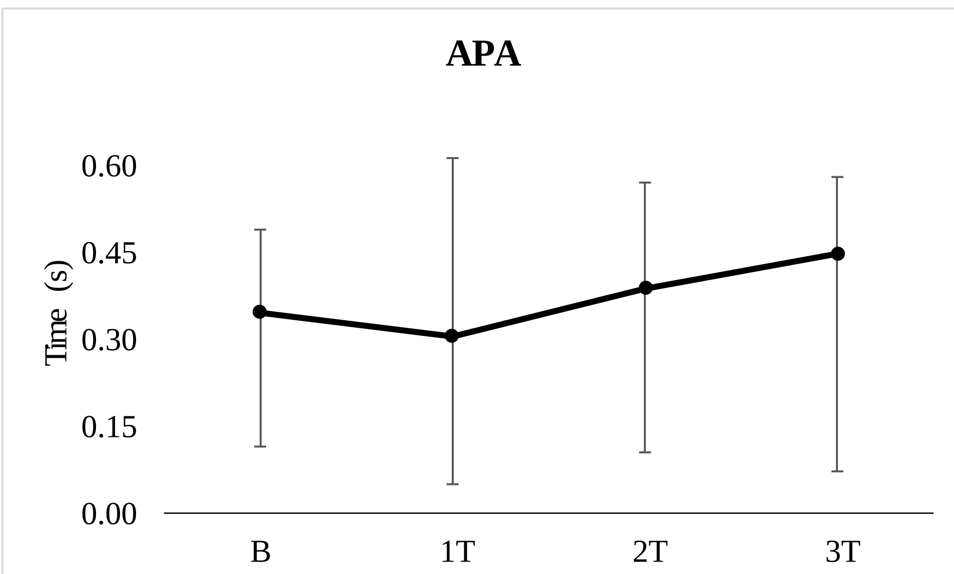

Figure 14. Anticipatory Postural Adjustment duration medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

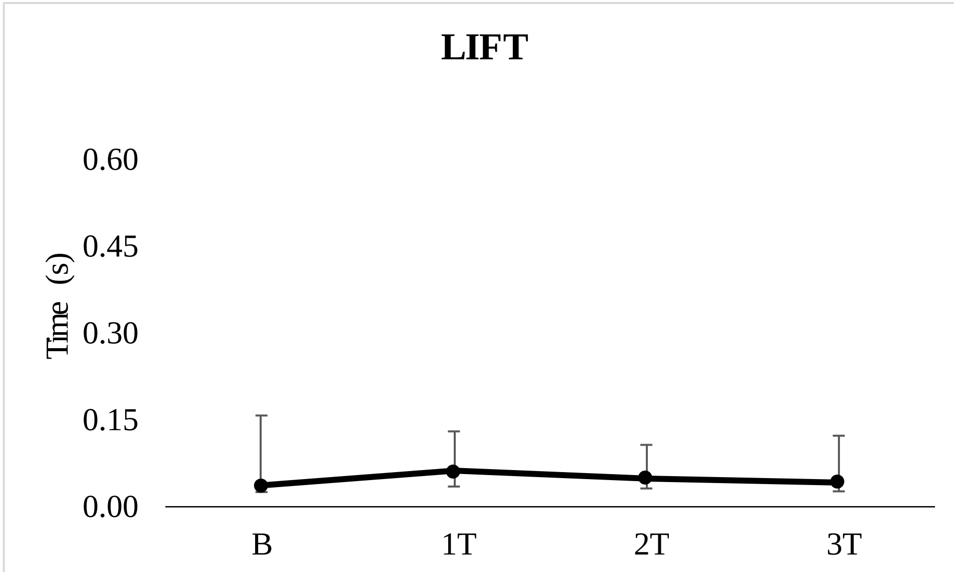

Figure 15. Foot lift duration medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

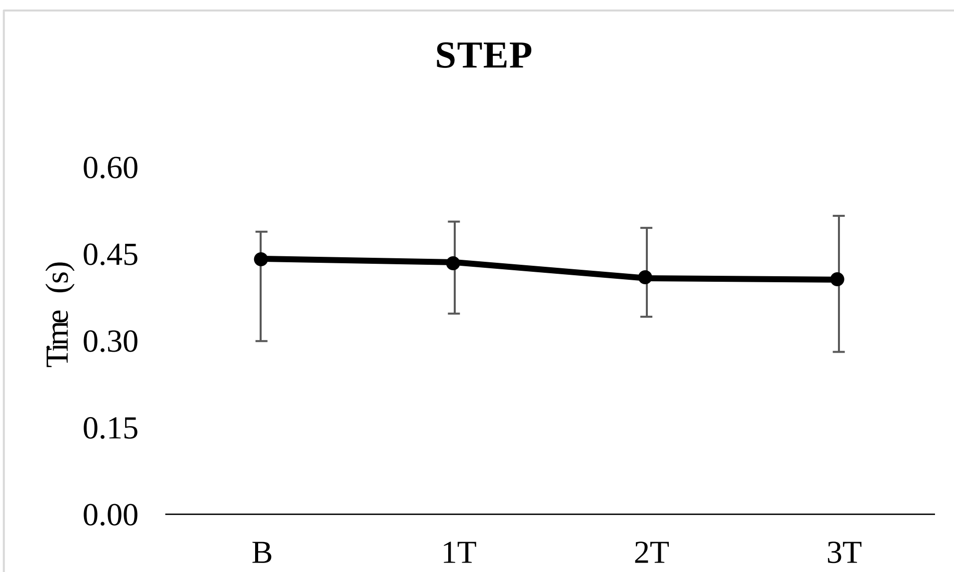

Figure 16. Step duration medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

Additionally, results revealed no statistical differences in MOS_{AP} ($\chi^2(3) = 4.09, p = 0.25$) and nor MOS_{ML} ($\chi^2(3) = 5.84, p = 0.12$) across simulated pregnancy visits. **Figure 17 & Figure 18** display the medians and ranges of MOS_{AP} and MOS_{ML}, respectively.

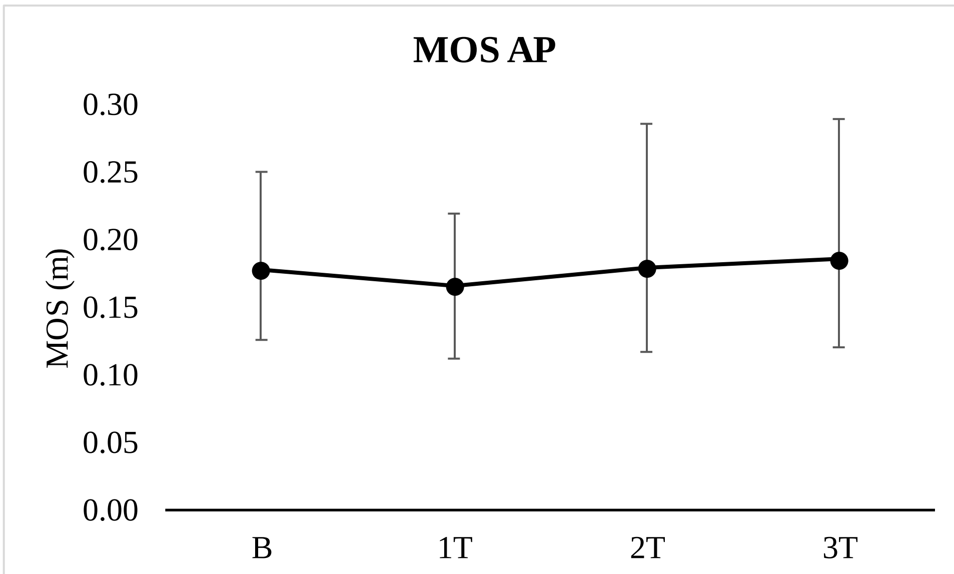

Figure 17. Anteroposterior margin of stability medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

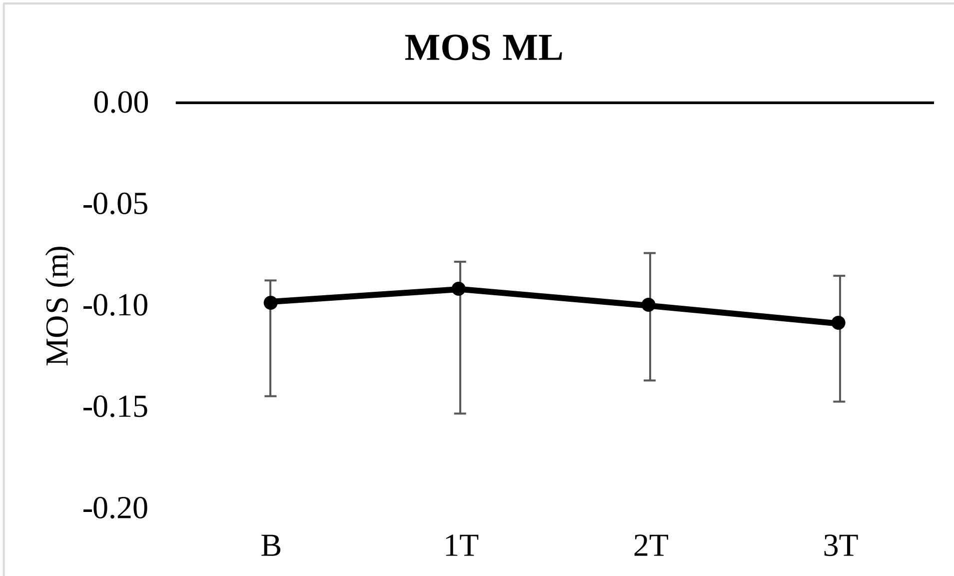

Figure 18. Mediolateral margin of stability medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

Lastly, results from the Friedman tests revealed no statistical differences in step length ($\chi^2(3) = 4.63, p = 0.20$) and nor step width ($\chi^2(3) = 5.73, p = 0.13$) across visits. **Figure 19 & Figure 20** display the medians and ranges of step length and step width, respectively.

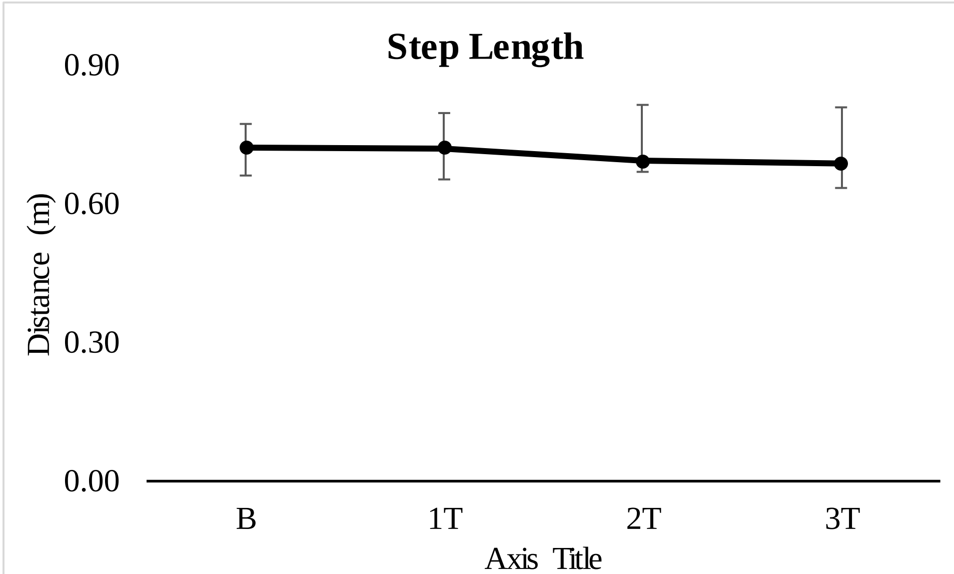

Figure 19. Step length medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

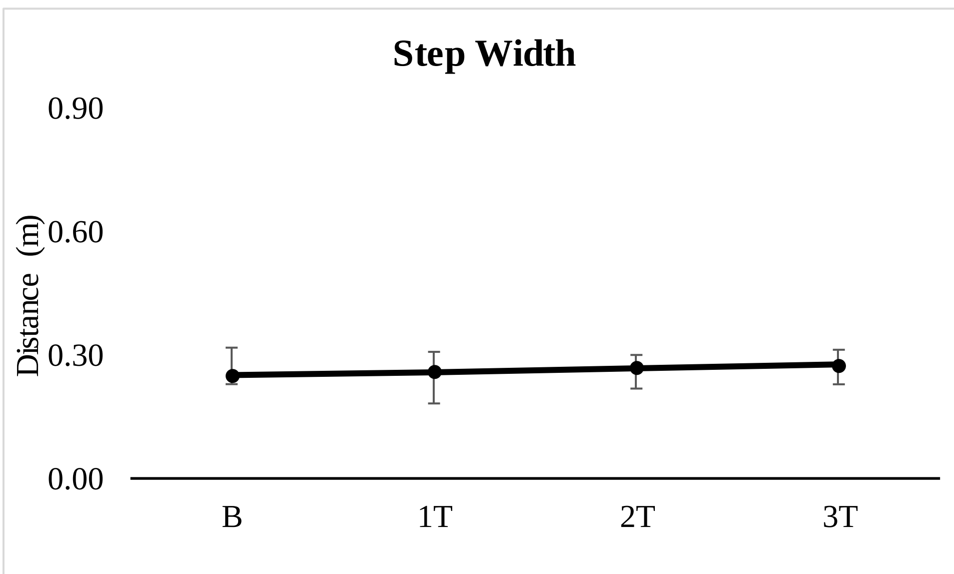

Figure 20. Step width medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

Discussion

The purpose of this study was to examine dynamic balance mechanics during gait initiation associated with the mechanical influences of simulated GWG. Dynamic stability outcomes were expected to differ in response to the increasing simulated GWG, wherein duration in APA phase would increase, MOS would decrease, and step width would increase; however, the results did not support this hypothesis.

The postural phase of gait initiation consists of the APA, which is responsible for postural stability and creating the propulsion into the execution phase (LIFT and STEP), followed by steady-state gait (Brenière, Cuong Do, & Bouisset, 1987; Yiou, Caderby, Delafontaine, Fourcade, & Honeine, 2017). Since APA duration plays a pivotal role in initiating gait and pre-determining dynamic stability (Yiou, Caderby, Delafontaine, Fourcade, & Honeine, 2017), the duration of time spent in the APA phase was expected to increase as a response to the increasing simulated GWG at each visit in the current study. In a previous study, Caderby and colleagues (2013) reported that an additional external load of fifteen percent of body mass, added symmetrically to a belt around the body, significantly increased the duration of time in the APA phase. Conversely, Qu and colleagues (2021) revealed that overweight and normal weight individuals performed gait initiation with similar APA durations, suggesting that the APA phase may not be affected until individuals are in the obese BMI category. Additionally, there has been some speculation that an external load, rather than an internal increased body mass, may be the cause for increased APA duration (Caderby et al., 2017). However, the results from the current study, wherein APA duration remained statistically similar despite the increasing and asymmetrical external loads about the torso, did not support either of these previous findings. The lack of significant alterations to APA duration could be due to the amount of external mass being standardized to participant BMI—the reason being was to follow the pregnancy weight gain recommendations from the CDC, which is determined by pre-pregnancy BMI, not body mass—rather than a percentage of individual body mass, creating some variability in both condition magnitude and strategy among participants. However, none of the participants exceeded overweight BMI when wearing the heaviest external load, which could also support the similar findings to Qu (2021). Furthermore, participants may have displayed an accommodation effect due to the mass increasing gradually at each visit, making the duration in APA statistically similar.

Dynamic stability, when quantified by MOS, is typically maintained in healthy populations (Hof, Gazendam, & Sinke, 2005) but hindered by pathological gait patterns (Azizah Mbourou, Lajoie, & Teasdale, 2003). Krkelijas (2018) established that MOS during gait was not altered by advancing pregnancy but rather influenced by increased step length; however, it was posited that recruitment of women with overweight pre-pregnancy BMIs may be responsible for these results and that normal BMI women would have displayed significant changes to gait pattern. However, the current study did not support this hypothesis, since normal BMI women were utilized yet MOS remained similar with increasing simulated GWG in both anteroposterior and mediolateral directions. Additionally, Krekelijas observed MOS during steady-state gait while the current study observed MOS at the initial heel contact of gait initiation, which could explain the negative (therefore unstable) values reported consistently across simulated trimesters. Gait initiation from quiet standing to steady-state gait remains underexplored in pregnancy research, however a recent study from Sunaga and colleagues (2020) determined that pregnant women had greater instability when transitioning from sit-to-stand to gait, requiring a change in direction to regain stability; albeit the assumption that pregnant women were unstable was based on vertical ground reaction force (vGRF) rather than MOS measurements.

Lastly, it was hypothesized that MOS and initiation phases would be affected by increasing simulated GWG, therefore step characteristics such as step length and/or step width would be increased as a stabilizing strategy; however, none of these hypotheses were supported by the results. Previous gait research with pregnant women suggested that pregnant women utilize an increased step width and decreased step length for stability purposes (Błaszczyk, Janusz, Opala-Berdzik, & Plewa, 2016), but this suggestion is controversial as others have determined that step width is a mechanical adaptation to pregnancy, rather than a functional mechanism to increase stability (Gilleard, 2018). While surprising that none of the variables measured in this study were affected by increasing simulated GWG, it could be due to a stabilizing strategy that was not measured, or the population being comprised of healthy, young women within the normal BMI classification.

In conclusion, dynamic stability during gait initiation was not affected by increasing GWG in normal BMI nulliparous women. These results may translate to women who are normal BMI pre-pregnancy, but could benefit from further exploration. Some limitations of the current study include the recruitment of nulliparous women as substitute for pregnant women and excluding overweight BMI women, however both of these decisions were based in controlling for confounding factors.

CHAPTER 4

Dynamic Stability During Slip Perturbation in Nulliparous Women with Simulated Gestational Weight Gain

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Dynamic Stability During Slip Perturbation in Nulliparous Women with Simulated Gestational Weight Gain

Significance of Chapter

This study will compare dynamic stability during slip perturbations across simulated trimesters. The aim of this study is to examine dynamic postural stability recovery mechanics influenced by gestational weight gain.

Introduction

The World Health Organization (2021) describes falls as a major public health concern to the general population, classifying falls as the second leading cause of unintentional injury-related deaths in the world. The elderly have the highest risk of fall-related mortalities; however, non-fatal fall-related hospitalization is a serious concern across all demographics, thus invoking extensive research in fall mechanisms and conditions (Chang, Leclercq, Lockhart, & Haslam, 2016; World Health Organization, 2021). Although not typically classified as a high fall-risk group, twenty-five percent of pregnant women fall during pregnancy, which is comparable to the fall rates of the elderly (Dunning, LeMasters, & Bhattacharya, 2010). Additionally, pregnant women who are hospitalized following a severe fall are of particular concern due to the risk of maternal and fetal complications associated with fall-related injuries (Schiff, 2008). Despite the breadth of fall research, falls in pregnant women continue to be understudied. Oftentimes, fall mechanisms in pregnant women are derived from indirect measures and are not directly studied, which may create an inaccurate understanding of what actually causes pregnant women to fall.

Across all demographics, falls most often occur due to loss of balance from either a trip or slip (Batterman & Batterman, 2005). A trip, wherein the foot of the swing limb inadequately clears the ground, typically results in a forward fall; whereas a slip, which occurs when the stepping limb has insufficient friction between the foot and the floor, typically results in a backward fall (Chang, Leclercq, Lockhart, & Haslam, 2016). Slips are roughly twice as likely to result in a fall when compared to trips (Redfern et al., 2001), which is thought to be related to the decreased margin of postural stability, and increased recruitment of muscle activity required to recover from a slip in the backward direction (Lee, Kim, & Seo, 2019). Whether from a trip or slip, falls are multicausal, involving intrinsic and environmental factors that may lead to loss of stability (Hsiao & Simeonov, 2001). Intrinsically, the visual, vestibular, and somatosensory systems contribute to preserving stability (Peterka, 2018); whereas environmental factors, such as navigating settings that are dimly lit, noisy, slippery/cluttered, may temporarily compromise one of the intrinsic systems responsible for maintaining postural stability (Boelens, Hekman, & Verkerke, 2013).

Generally, postural stability is maintained when the location of the body's center of mass (COM) remains within the of the base of support (Hof, Gazendam, & Sinke, 2005). When a perturbation threatens the maintenance of postural stability, either an anticipatory or compensatory postural control strategy is utilized in response (Duarte et al., 2022). For example, when transitioning from static to dynamic stability, such as gait initiation, the central nervous system (CNS) utilizes anticipatory postural adjustments (APAs) to prepare the postural muscles to control for the whole-body movement that will occur as the COM accelerates laterally and (if successful) forwards with the execution of the initial step (Yiou, Caderby, Delafontaine, Fourcade, & Honeine, 2017). Conversely, if a large-magnitude perturbation unexpectedly disturbs postural stability, such as during a slip, the CNS utilizes compensatory postural control to reactively produce a motor response to rapidly increase the BOS, shift the COM in the opposing direction of the perturbation, and take an additional compensatory step to recover balance, if needed (Martelli et al., 2017).

Consequently, dysfunction within the neuromuscular system can impair the intrinsic responses liable for preventing loss of balance and falls, making certain populations at higher risk for falls (Gauchard, Chau, Mur, & Perrin, 2001). In pregnant women, the onset of gestational weight gain (GWG), which can culminate to roughly twenty-five percent of pre-pregnancy body mass (Hagan & Wong, 2010), creates an uneven distribution of mass at the torso and can naturally shift whole-body COM anteriorly (Whitcome, Shapiro, & Lieberman, 2007). Given that both anticipatory and compensatory strategies require control of the COM, and rapid corrective motor abilities to maintain balance during a slip or trip (Martelli et al., 2017; Yiou, Caderby, Delafontaine, Fourcade, & Honeine, 2017), pregnant women may struggle to effectively perform the corrective task to avoid a resulting fall. Additionally, pregnancy-related hormonal fluctuations impact the neuromuscular systems, increasing ligament laxity, swelling, and weakness in the muscles and joints of the lower extremities (Talbot & Maclennan, 2016), likely adding to the difficulty of adapting to a sudden perturbation.

Analyzing dynamic stability during gait and measuring slip recovery following a perturbation are common in fall-risk research (Yang, F., Kim, & Yang, 2017; Yang, Feng, Bhatt, & Pai, 2013). Dynamic stability responses to slip perturbations have yet to be measured in pregnant women; however, it is not ethically justifiable to expose pregnant women to these high-risk conditions. Therefore, nulliparous women with simulated gestational weight gain (GWG) were recruited for the current longitudinal study design to eliminate safety risks. While these findings may not translate directly to pregnant women, the isolated effects of GWG and mechanical adaptations can be observed. Thus, the purpose of this study was to determine if the mechanical influences of simulated GWG during pregnancy affect dynamic stability during slip recovery. It was hypothesized that dynamic stability would decrease at the instance of the slip perturbation and during the recovery step, as anterior load increased across simulated gestational weight gain.

Methods

Participants

An *a priori* power analysis was conducted on G*Power (version 3.1, Universität Kiel, Germany), with data from Yang and colleagues (2017). Based on a proposed effects size of 0.94, power of 0.85, and alpha (α) of 0.05, it was determined that 10 participants were required to achieve adequate statistical power. A total of seventeen nulliparous women were recruited for this study in anticipation of participant drop-out; seven participants were removed from analyses due to drop-out or insufficient data. The ten nulliparous women (25.00 ± 4.06 years; 1.64 ± 0.06 m; 60.36 ± 6.87 kg; 22.42 ± 1.39 kg/m^2) analyzed in this study were required to be between the ages of 18 and 34, within "normal" body mass index (BMI; 18.5 – 24.9), and free of lower limb injuries. Prior to completing any study-related tasks, written informed consent was obtained on institutional approved documentation (Protocol No: 1727598-9) and in accordance with 1964 Declaration of Helsinki.

Procedures

Data collections were conducted on four separate days, with a minimum of two days between visits to avoid fatigue as a confounding factor (Cheung, Hume, & Maxwell, 2003) and a maximum of four weeks between visits, to avoid potential fluctuation in body mass and loss of retention in participants. Baseline (B) measurements were obtained during the initial data collection, wherein age, height, mass, and body mass index (BMI) of participants were measured and recorded. At each subsequent collection, participants were fitted with an adjustable Velcro-secured, plate-loaded, weight-vest with an additional mass secured anteriorly to be equivalent to the mass at the terminal end of three months pregnant at the first trimester (1T); added mass equivalent to the end of six months pregnant at the second trimester (2T); and added mass equivalent to the end of nine months pregnant at the third trimester (3T). It is worth noting that baseline tasks were performed while wearing the weight-vest without additional mass, which added an additional three kilograms to participants' body mass on average but did not interfere with the BMI classification of the participants. Additional mass magnitude for simulated GWG was determined by participants' baseline BMI in order to simulate body mass at the terminal stage of third, sixth, and ninth months of pregnancy based on the mass gain recommendations by the Institute of Medicine (IOM) (National Research Council, 2010) (for example, a woman with a height of 1.62m, body mass of 59kg, and BMI of 22.5 kg/m^2 would have the following mass amounts added upon separate visits: 2.0, 9.2, and 15.9 kg to simulate months three, six, and nine of pregnancy, respectively).

Participants warmed up for 10 minutes at a self-selected, comfortable walking speed on the ActiveStep treadmill (Simbex, NH) while wearing the weight-vest to acutely adapt to the additional anterior load. During the warm-up, participants were also secured with the full-body safety harness to adapt to walking while harnessed and to protect participants from potential falls. The safety harness connects by shock-absorbing ropes at the shoulders anchored to the ceiling. Afterward, a total of sixty-four retroreflective markers were adhered to the head, trunk, upper extremities, pelvis, lower extremities, and feet to track movements of the respective body sections **Appendix A**. Five additional markers were placed on the weight plate to measure center of mass (COM) location on simulated pregnancy visits.

For testing trials, participants returned to the ActiveStep treadmill and were informed that they may experience a "slip-like" movement on the treadmill. In contrast with a regular treadmill, the ActiveStep treadmill is a specialized device that aims to produce a “fall” in a secure environment via slip perturbations. Participants walked for ten trials, experiencing a slip each time, randomized by number of steps, within a range of five to twelve steps. The slip perturbations consisted of sudden forward acceleration of the treadmill belt, within 0.2s, with identical and standardized intensity for all participants. During all trials, marker trajectories were tracked using a 10-camera three-dimensional motion capture system (200 Hz, Vicon Motion Systems, Ltd., Oxford, UK).

Outcome variables were measured at three events, the heel strike before the slip (HS1), the first recovery heel strike (HS2), and the second recovery heel strike (HS3). Measurements of dynamic stability were measured from the described methodology from Yang and colleagues (2022). Dynamic stability was determined by the combined COM (i.e., of the body and the plate) anteroposterior position and velocity relative to the back of the BOS made by the leading heel.

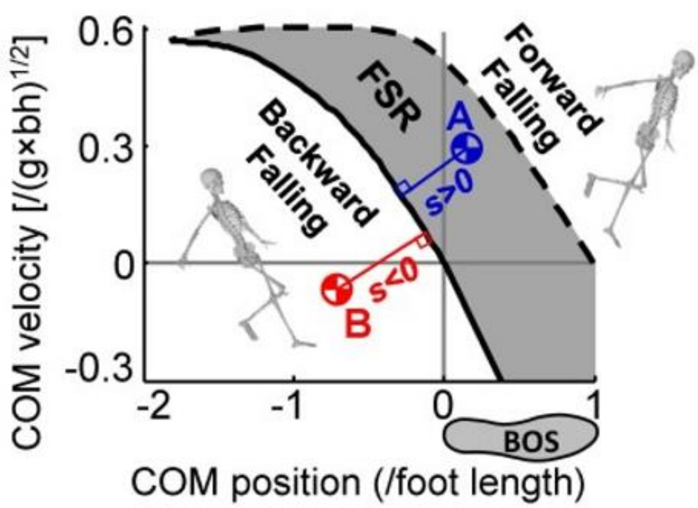

Figure 21. Feasible stability region (FSR) based on the center of mass (COM) motion relative to the base of support (BOS) (Yang et al., 2022).

Feasible Stability Region Theory (FSR) (**Figure 21**) at the time of each event (HS1, HS2, HS3) will be used to determine dynamic stability.

FSR is determined by two components of COM motion (anteroposterior position and velocity) relative to the base of support (BOS), calculated relative to the leading heel of the BOS. COM position was normalized by foot length and COM velocity was normalized by $\sqrt{g/bh}$, where g represents gravitational acceleration and bh represents body height recorded from the time of baseline anthropometric measurements (Yang, Feng, Ban, & Yang, 2022). Dynamic stability will be calculated as the shortest distance from COM motion to the threshold against backward balance loss. Negative dynamic stability outcomes indicate instability associated with backwards falling, such as during a slip.

Statistical Analysis

Statistical analyses were conducted using SPSS 29 (IBM, NY). Mean (SD) values were calculated for all variables. An independent one-way analysis of variance (ANOVA; α=0.05) was conducted to compare subject mass with the additional mass (simulated pregnancy body mass) at each visit (B, 1T, 2T, 3T). If a significant difference was detected in the omnibus ANOVA test, pairwise comparisons were interpreted after applying the Sidak adjustment.

Separate repeated measures tests were utilized to compare the dynamic stability calculated from the FSR at each heel strike (HS1, HS2, HS3). Data normality was evaluated for each variable using the Shapiro-Wilk test. Normally distributed data were assessed with individual one-way repeated-measures ANOVA tests. If Mauchly's test of sphericity was violated, then sphericity was assumed. If data were not normally distributed, individual Friedman's tests were used for comparisons. If statistical significance was detected, pairwise comparisons were performed for multiple comparisons between visits with a Holm-Bonferroni method adjustment to the *p* values to control for family-wise error rates among multiple tests (Holm, 1979).

Results

One-way ANOVA results revealed significant differences in simulated pregnancy body mass among visits, $F_{(3, 36)} = 12.79$, $p < 0.001$, $\eta^2 = 0.64$. **Table 5** displays the means and standard deviations of simulated pregnancy body mass among visits.

Table 5. Means and standard deviation values for simulated pregnancy body mass.

	Baseline	First Trimester	Second Trimester	Third Trimester
Mass (kg)	63.82 ± 6.94 **a, b**	66.48 ± 6.87 **c**	74.96 ± 6.84 **a**	80.94 ± 7.14 **b, c**

Statistical significance is denoted by **a** $p < 0.01$, **b** $p < 0.001$, and **c** $p < 0.001$

Results from the Friedman test revealed statistical differences in HS1 measures across visits $(\chi^2(3) = 16.56, p < 0.001)$. Initially, post hoc pairwise comparisons using the Wilcoxon test revealed that B was significantly different from 2T ($p < 0.001$), which was confirmed by the Holm-Bonferroni adjustment (adjusted $p = 0.0083$); B was significantly different from 3T ($p = 0.002$), confirmed by the Holm-Bonferroni adjustment (adjusted $p = 0.01$); and B was significantly different from 1T ($p = 0.038$), however, after applying the Holm-Bonferroni adjustment (adjusted $p = 0.0125$), the null hypothesis was not rejected. All other pairwise comparisons were not significantly different. **Figure 22** displays the median value and range of dynamic stability in HS1 across visits.

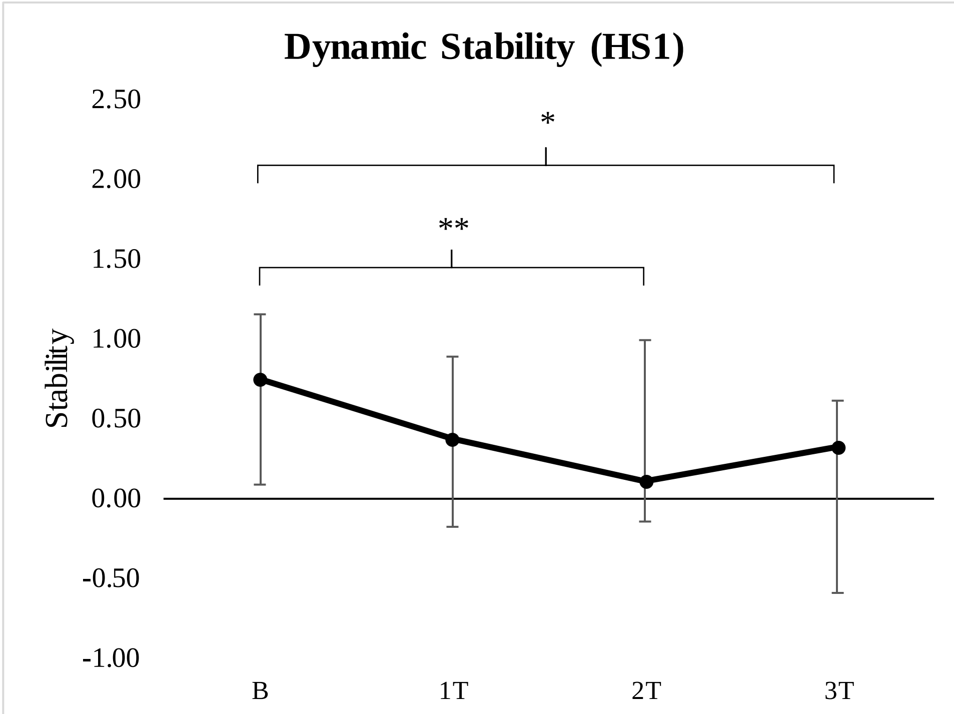

Figure 22. Dynamic stability in first heel strike medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

* denotes $p < 0.05$, ** denotes $p < 0.001$

The repeated measures ANOVA test determined that sphericity was not violated $(\chi^2(5) = 3.15, p = 0.68)$ and revealed significant differences in HS2 measures across visits, ($F_{(3, 27)} = 6.76$, $p = 0.002$, $\eta^2 = 0.43$). Pairwise comparisons revealed B was significantly different from 1T ($p = 0.007$), which was confirmed by the Holm-Bonferroni adjustment (adjusted $p = 0.0083$); and B was significantly different from 3T ($p = 0.029$), which was rejected by the Holm-Bonferroni adjustment (adjusted $p = 0.01$). All other comparisons were not significantly different. **Figure 23** displays the means and standard deviations of dynamic stability in HS2 across visits.

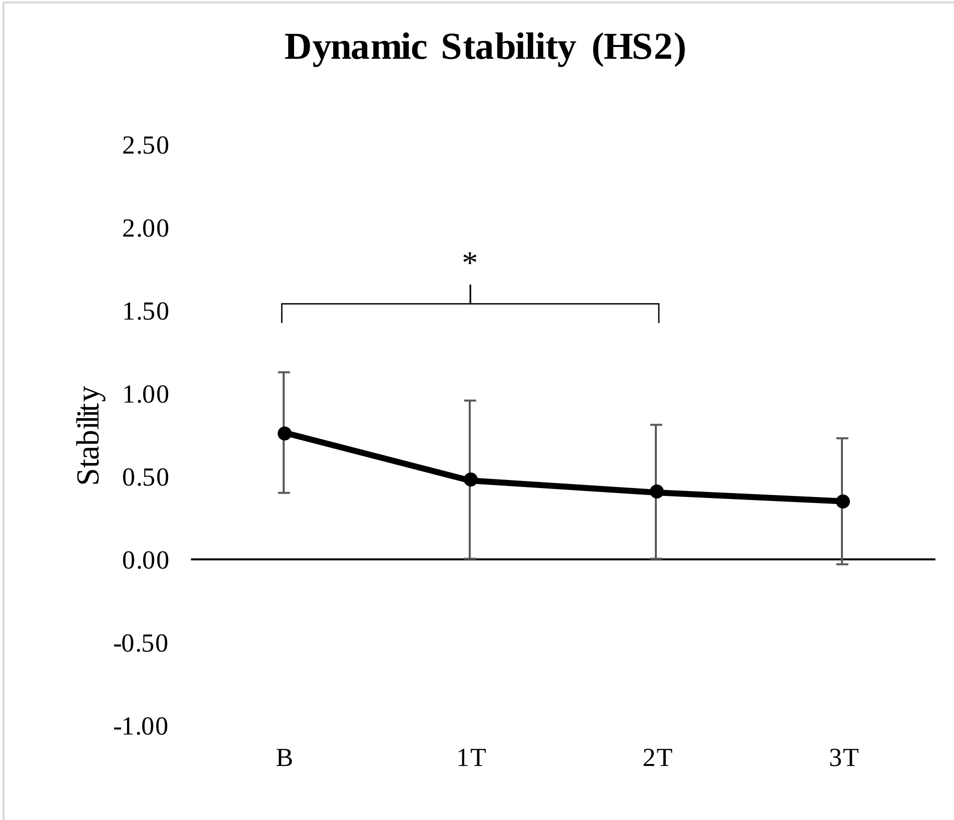

Figure 23. Dynamic stability in second heel strike means and standard deviations across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

* denotes $p < 0.01$

Results from the Friedman test revealed statistical differences in HS3 measures across visits ($\chi^2(3) = 14.04, p = 0.003$). Initially, post hoc pairwise comparisons using the Wilcoxon test revealed that B was significantly different from 2T ($p < 0.001$), which was confirmed by the Holm-Bonferroni adjustment (adjusted $p = 0.0083$); and B was significantly different from 3T ($p = 0.003$), confirmed by the Holm-Bonferroni adjustment (adjusted $p = 0.01$). All other pairwise comparisons were not significant. **Figure 24** displays the median value and range of dynamic stability in HS3 across visits.

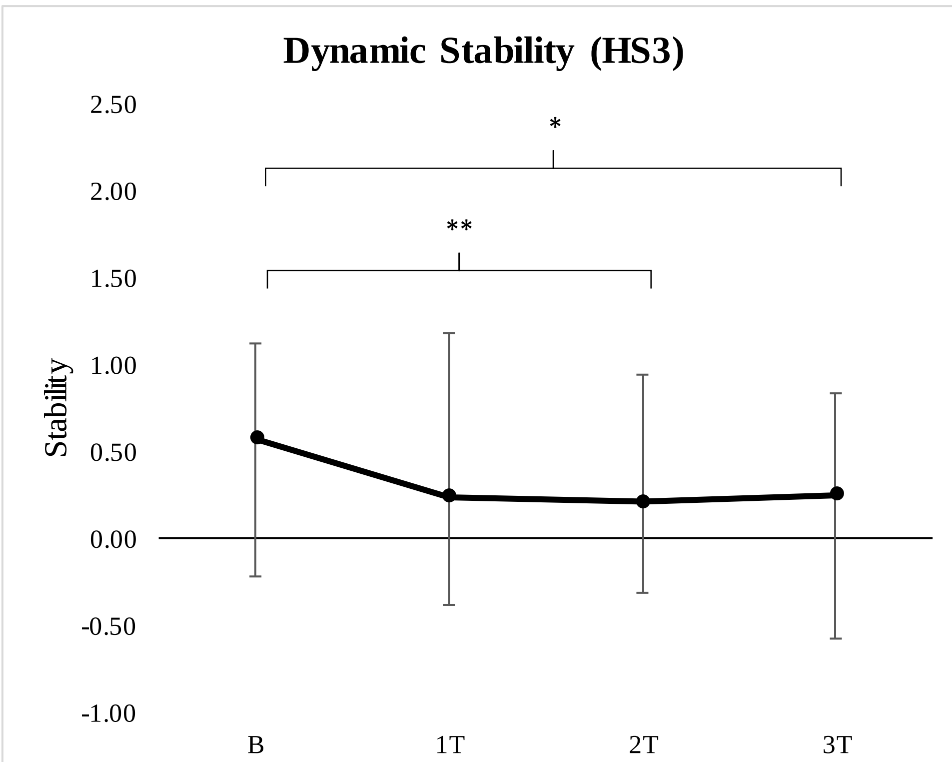

Figure 24. Dynamic stability in third heel strike medians and ranges across baseline (B), 1st trimester (1T), 2nd trimester (2T), and3rd trimester (3T) visits.

A single asterisk (*) denotes $p < 0.005$; double asterisks (**) denotes $p < 0.001$

Discussion

This study aimed to examine the potential mechanical influences of GWG during pregnancy on dynamic stability during slip recovery. It was hypothesized that dynamic stability would decrease at the instance of slip perturbation and during the recovery step, as simulated GWG increased. The results from the current study supported the hypothesis, wherein dynamic stability decreased significantly from baseline at each step.

During pregnancy, seventy-two percent of fallers reported falls on stairs and slippery surfaces, with the highest rate of falls occurring at seven months, or the beginning of the third trimester, with decreasing fall rates towards the end of the third trimester (Dunning, LeMasters, & Bhattacharya, 2010). The results from this study supported these findings, revealing significant decreases in dynamic stability during simulated GWG loads at the end of the second trimester; and to a lesser degree, significance into the third trimester during the step before the slip (HS1) and the step following the slip (HS3). Notably, no significant differences were found between baseline and the first trimester, which is a trend frequently cited in the literature (Butler, Colón, Druzen, & Rose, 2006; Inanir, Cakmak, Hisim, & Demirturk, 2014). The increased rate of falls near the end of the second trimester could be related to the rapid amount of weight gained during this period (Institute of Medicine, 2009), and would certainly explain the findings in the current study.

Interestingly, the instance at which participants were “catching” themselves from the backward fall, or the HS2 event, did not reveal to be as instable as during the recovery step forward, in HS3. In fact, the most unstable results were found to occur during HS3. This could be due to the slip occurring on a treadmill, wherein the legs are pulled across the belt during gait before the slip (Ahn, Simpkins, & Yang, 2022), but the transition from a sudden backwards step into a forward continuation of steps requires may more effort and adjustments of the body's COM than the slip itself. Additionally, the findings from the current study supported those from a similar study from Yang and colleagues (2022), wherein an anterior load of similar magnitude (roughly 10 and 20 percent of participant's body weight) held within the arms increased instability during slip perturbations. Interestingly, this trend did not continue further into the highest GWG load, wherein 2T and 3T were not significantly different, and B was not statistically different from 3T; however, there may have been a learning affect as participants performed the task repeatedly and were both cognitively and physically familiar with the task by the time they donned the 3T load.

Limitations of this study include the use of nulliparous women rather than pregnant women, however this was required due to the nature of the study and the risk involved. Participants were allowed to select their own speed at each data collection making the initial speed before the slip not standardized; however, the decision to not control for treadmill speed was made to observe natural accommodations to increasing load, since previous literature has suggested that pregnant women decrease gait speed into the later trimesters. Additionally, each trial contained a slip, so participants were aware that the slip would happen but were not told when it would occur, which may have altered their gait pattern.

Future research should consider measuring other strategies that may have affected stability during the slips. Observationally, many women leaned noticeably in the mediolateral direction when recovering from the slip. Including trunk inclination and step characteristics could have given more insights into the strategy involved. In conclusion, simulated GWG negatively affected the mechanical aspects of dynamic stability during slip recovery.

CHAPTER 5

Conclusion

The overarching purpose of this dissertation was to assess static and dynamic stability associated with the mechanical influences of pregnancy in nulliparous women while using simulated gestational mass at each "trimester" to ensure safety and control for contributing factors. Three specific aims were outlined within this overarching purpose: 1) to examine static postural stability; 2) to explore dynamic stability during gait initiation; and 3) to explore dynamic stability during slip recovery.

Throughout the study, the most noteworthy findings were revealed in the second simulated trimester, aligning with existing research on fall rates during pregnancy. Surprisingly, few significant findings were found in static balance and gait initiation tasks, yet significance was found in slip perturbations. Since the participants throughout the study were from the same dataset, it was surprising that the static balance tasks did not have more predictive insights on dynamic stability outcomes. These findings could suggest that relying solely on static postural stability assessments may not adequately predict fall risks in pregnant women. Moreover, the significant decreases to stability during slip perturbations implied that weight gain may play a pivotal role in the person's ability to respond to a sudden perturbation, albeit within the constraints of the simulated pregnancy conditions.

In conclusion, this study may offer valuable information on the mechanical influences of pregnancy on stability in nulliparous women with simulated gestational loads, further research is warranted to deepen our understanding and enhance fall risk assessment for pregnant women. Additionally, exploring alternative methods for assessing dynamic stability and fall risks, beyond traditional static postural stability assessments, could provide valuable insights in mitigating fall risks in pregnant women.

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APPENDIX

Appendix A. Full-body VICON Marker Placement

Segment	Bilateral Placement	Singular Placement
Head	Front of head	
	Back of head	
Torso/ Pelvis	Acromion process	Jugular notch
	Iliac crest	7th cervical vertebra
	Anterior superior iliac spine	Xiphoid process
	Posterior superior iliac spine	10th thoracic vertebra
	Greater trochanter	Sacrum
Upper Arm	Upper arm cluster (4)	
	Lateral humeral epicondyle	
	Medial humeral epicondyle	
Forearm	Forearm cluster (4)	
	Radial styloid process	
	Ulnar styloid process	
Leg	Thigh cluster (4)	
	Lateral femoral epicondyle	
	Medial femoral epicondyle	
Shank	Shank cluster (4)	
	Lateral malleolus	
	Medial malleolus	
Foot	1st distal phalanx	
	1st metatarsal	
	5th metatarsal	
	Calcaneus (3)	

VITA

Heather R. Vanderhoof holds a Bachelor of Science degree in both Nutrition Sciences and Kinesiology, as well as a Master of Science degree in Kinesiology. Heather has dedicated her academic career to understanding the science of human movement and physical well-being.

With a passion for both teaching and research, Heather served as a Lecturer for the undergraduate course of Introduction to Biomechanics (KIN4313) at the University of Texas at El Paso (UTEP) for three consecutive semesters. This role offered her the opportunity to inspire and guide students through the fundamental principles of biomechanics. Prior to her role as Lecturer, Heather honed her teaching skills as a Laboratory Teaching Assistant for KIN 4313 for two years, providing hands-on instruction and support to students in the supplemental laboratory setting.

In addition to her teaching experience, Heather has contributed to the biomechanical research field as a Graduate Research Associate for four years. Her dedication to advancing knowledge within the field is evident through her involvement in various research projects aimed at understanding human movement patterns, including publications in the following journals: (insert paper names and journals here???) Journal of Biomechanics, MDPI Biomechanics, Perceptual & Motor Skills, and Human Movement Science.

Heather's main research endeavors are fueled by a commitment to representing women at every stage in life, with a particular focus on pregnancy. Heather is dedicated to improving the quality of life for pregnant individuals and reducing fall risks in this underrepresented group. Her work aims to empower women with evidence-based findings to maintain physical well-being throughout pregnancy, ultimately contributing to advantageous outcomes for both mothers and their families.

Outside of academia, Heather is a Certified Strength and Conditioning Specialist, bringing her expertise in exercise science and training to individuals seeking to enrich their physical fitness and improve their quality of life. With a commitment to lifelong learning and a passion for empowering women to lead healthy and active live at every stage, Heather continues to make meaningful contributions to her field through her teaching, research, and professional practice.