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Faculdade de Motricidade Humana

Biomechanical models of the lower limb and pelvis, for female human gait in regular and overload conditions related to pregnancy

Tese elaborada com vista à obtenção do Grau de Doutor em Motricidade Humana, na especialidade de Biomecânica

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Título

Modelação biomecânica do membro inferior e pelvis na marcha da mulher em condição de sobrecarga relacionada com a gravidez

Sumário

A gravidez é uma fase especial da vida , considerando as adaptações morfológicas, fisiológicas, biomecânicas e hormonais vivenciadas pelas mulheres durante cerca de 40 semanas e no período pós-parto, podendo modificar o padrão de marcha e contribuir para uma sobrecarga no sistema músculo-esquelético, causando dor nos membros inferiores, bacia e zona lombar. Os objetivos do presente trabalho foram: 1) analisar a marcha de mulheres grávidas no segundo trimestre; 2) comparar as adaptações biomecânicas da marcha, entre as mulheres grávidas no segundo trimestre, mulheres não grávidas e mulheres com condições de sobrecarga artificiais; 3) analisar modelos biomecânicos com quatro *set ups* diferentes de análise; e, 4) analisar um modelo de contacto que determina a força vertical de reação do apoio. Os resultados demonstraram que as mulheres grávidas têm uma padrão de marcha similar ao normal. Observou-se que o ganho do peso no tronco aumenta o tempo das fases de apoio e de duplo apoio, quer nas mulheres grávidas quer nas mulheres com carga adicional. A resposta ao momento externo flexor da anca está relacionada com maior atividade dos extensores para suportar a carga anterior do tronco na direção da translação do centro de massa. Nas mulheres grávidas, o modelo universal-revolução-esférica afetou mais as variáveis cinemáticas quando comparado com o modelo de juntas com seis graus de liberdade. O modelo de contacto entre o pé e o solo, sobrestimou as forças verticais de reação. O aumento da massa do pé, devido ao inchaço consequente da gravidez, reduz a rigidez durante a fase de apoio. Os resultados do presente trabalho serão úteis para promover a investigação biomecânica do padrão de marcha durante a gravidez.

Palavras chave: gravidez, marcha, modelos biomecânicos, artefacto gerado pelo tecido mole, simulação.

Title

Biomechanical models of the lower limb and pelvis, for female human gait in regular and overload conditions related to pregnancy

Abstract

Pregnancy is a special phase of life, considering the morphologic, physiological, biomechanical and hormonal experienced by women during about 40 weeks and in the post-partum period. Such changes can modify the gait pattern and contribute to an overload on the musculoskeletal system, causing lower limbs, hip and lower back pain. The purposes of the present thesis were to: 1) analyze the second trimester pregnant women gait; 2) compare the biomechanical adaptations of gait, between second trimester pregnant women, non-pregnant women and women with artificial overload conditions; 3) analyze biomechanical models with four different set ups of joints; 4) to establish a contact model to predict vertical ground reaction forces. Results showed that pregnant women have a similar walking pattern to the normal gait. It was possible to observe that the trunk weight gain increases both stance phase duration and double limb support time in both load carrying and pregnant subjects. The body's response to the external hip flexor moment is related to a higher extensor activity to support the anterior additional mass of the trunk and to the forward translation of the center or mass. In pregnant women, the universal-revolute-spherical model affects more the kinematic variables when compared with the six degrees of freedom model. The foot contact model overestimated vertical ground reaction force. The foot swelling is related to pregnancy, increasing the foot mass and reducing the stiffness during the contact. The results of the present work will be useful to further biomechanical research regarding gait during pregnancy.

Keywords: pregnancy, gait, biomechanical models, soft tissue artifact, simulation.

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List of Abbreviations

- 1T – First trimester
- 2T – Second trimester
- 3-D - Three dimensional
- 3T – Third trimester
- 6DOF - Six degrees of freedom
- ASIS - Anterior superior iliac spine
- BMI - Body Mass Index
- DLST - Double limb support time
- DOF - Degrees of freedom
- F1 – Loading response peak force
- F2 – Mid stance response peak force
- F3 – Terminal stance response peak force
- GCS - Global Coordinate System
- GOM - Global Optimization Method
- GRF - Ground Reaction Forces
- IOM - Institute of Medicine
- ISAK - International Society for the Advancement of Kinanthropometry
- LCG - Load carrying group
- SCS - Segment Coordinate System
- LCS - Laboratory Coordinate System
- LHJC - Left hip joint center
- Me – Median
- M_f – Moments of force
- NPG_LCG – Comparison between non-pregnant and load carrying groups
- NPG_PG - Comparison between non-pregnant and pregnant groups
- NPG – Non-pregnant group
- NRC - American Research Council
- PD - Proportional-derivate
- P – Pregnancy
- PG_LCG – Comparison between pregnant and load carrying groups
- PG - Pregnant group
- PP – Post-partum
- PSIS - Posterior superior iliac spine
- PW - Pregnant women
- RHJC - Right hip joint center
- RHS - Right heel strike
- ROM - Range of motion
- RTO - Right toe off
- SCS - Segment coordinate system
- SP - Stance phase
- SRS - Spherical-Revolute-Spherical
- SSS - Spherical-spherical-spherical
- STA - Soft tissue artifact
- SW – Swing phase
- URS - Universal-Revolute-Spherical

Chapter 1

1 Introduction

1.1 Rationale for the Investigation

One of the most studied human movements is gait, once it is the basic mean of locomotion and it is used throughout the life. Thus, since childhood to elderly, walking is the functional way of moving from one place to another. With growing, the human body will change anatomically, going through different stages of growth and in adulthood there is a possible stabilization of its development.

The musculoskeletal system is constantly subjected to mechanical loading which is a prerequisite for morphological and functional adaptations of biological materials. Bones, cartilages, ligaments and tendons have some stress and strain limits and once they are exceeded, micro and macro damages can occur. Pain and injuries have relation with overloading conditions, and as a bionegative response, this situation can cause loss of strength and function. On the other side, biological structures need mechanical stimulus to maintain or increase their function capacity and morphological structure (Brüggemann, 2005).

The human body weight is an anthropometric variable crucial to the amount of load that the body supports static and dynamically. Excessive body weight is related to several health problems and gait related injuries (Must & Strauss, 1999). Pregnancy is a special phase of life and also an overweight condition, with some specific associated health problems caused by the morphological, physiological and hormonal changes (Nicholls & Grieve, 1992). When walking the human body can adjust to different circumstances thereby changing the gait pattern. Muscle function can be reorganized to provide body acceleration and therefore modify the gait pattern, which may add an overload to the musculoskeletal system (Foti, Davis & Bagley, 2000). The assessment of biomechanical loading in the musculoskeletal system of the pregnant women is of particular interest since it is important to better understand the gait adaptations during pregnancy and few biomechanical studies can be found in literature

Moreover, it is important to quantify the biomechanical load and the musculoskeletal adaptations that occur during gait and identify which ones are more related with the increased trunk mass, or which ones can be more associated to physiological and hormonal changes. This analysis can provide more information about joint loads and kinematics and it may be useful to exercise prescription in order to reduce the risk of discomfort or injure associated to overload.

The changes associated to pregnancy as weight gain, skinfolds increase, increased ligamentous laxity, decreased neuromuscular control and coordination and swelling of the arms and legs (Pitkin, 1976; Taggart, Holliday, Billewicz, Hytten & Thomson, 1967; Butte, Ellis, Wong, Hopkinson & Smith, 2003; Block, Hess, Timpano & Serlo, 1985; Brett & Baxendale, 2001; Calganeri, Bird & Wright, 1982; deGroot, Adam & Hornstra, 2003; Dumas, Reid, Wolfe, Griffin & McGrath, 1995; Dunning et al., 2003; Hainline, 1994; McNitt-Gray, 1991; Snijders, Seroo, Snidjer & Hoedt, 1976) may contribute to the increase of the soft tissue artifact (STA). This can be a concern for biomechanical modeling, where the efficacy of gait analysis information may be restricted by sources of error like the soft tissue displacement relative to bone. So, there is the need to obtain accurate data on gait analysis, particularly regarding overweight conditions related to pregnancy. The global optimization method proposed by Lu and O'Connor (1999) estimates the general pose for each frame, by minimizing the differences between the measured coordinates in the static position and the measured coordinates during motion in order to find an optimal position for the set of segments. To minimize the STA effect in pregnant women, the GOM can be applied, where joint constraints are added to the model in order to minimize the sensor noise and estimate an optimal pose of a multi-link model that best matches the motion capture data in terms of global criterion (Lu & O'Connor, 1999).

Typically, the gait cycle is defined as the amount of time from the initial foot contact with the ground to the next instant when the same foot starts the following initial contact (Peterson & Bronzino, 2008). As a cyclical locomotor task, gait provides several impacts on the ground. Ground reaction forces produces joint reaction and moments of force, and some internal forces as bone-on-bone, ligament and muscle forces (Winter, 2009). To analyze human gait, ground reaction forces (GRF) and motion data are measured. Both parameters are used to calculate other variables, as joint moments which are calculated by the inverse dynamics method and add more information about the stress imposed in the joints and the necessary muscle control (Perry, 1992). Also, the GRF are a common indicator for external biomechanical load. When GRF are unobtainable, simulation models can predict these data and can add more information for the understanding of interactions of the musculoskeletal system with the physical environment, such as the foot floor contact. In a clinical context, applied to pregnant population, this methodology can be important to provide more important biomechanical information that could not be collected in another way and to monitoring the foot contact during the pregnancy.

1.2 General Research Questions

Walking is the primary mean of human locomotion and gait analysis is useful to quantitative measurement and assessment (Peterson & Bronzino, 2008). But throughout life, human body can adjust to different conditions and the gait pattern may change. During pregnancy, women are subject to morphological, physiological and hormonal changes, which can lead to adaptations in gait. It is important to quantify the internal and external biomechanical load related to pregnancy condition and to clarify which gait adaptations are more related with the increased trunk mass, or which can be more associated to physiological and hormonal changes.

Also, considering the pregnancy condition, there is a possibility of a higher STA and to minimize the associated error there is the need to use the GOM to obtain a more accurate three-dimensional (3-D) multibody biomechanical gait model that best represents the pregnant women.

In order to obtain all the variables, two solutions will were used: inverse and forward dynamics.

1.3 Purposes of the Thesis

The purposes of the present thesis were to:

1) Develop a 3-D biomechanical model of the lower limb, based on a rigid multibody system;
2) Describe the angular kinematics of the lower limb of young and pregnant women and women carrying an extra load;
3) Calculate net moments of force in ankle, knee and hip joints, using inverse dynamics;
4) Describe peak magnitudes of the joint moments of force of young and pregnant women and women carrying an extra load;
5) Determine the temporal parameters of the gait cycle of young and pregnant women and women carrying an extra load;
6) To identify potential differences between the three groups on the biomechanical parameters, concerning gait cycle;
7) To use three different constraining sets in the lower limb joints, in pregnant and non-pregnant women biomechanical models. The ankle, knee and hip joints

were modeled respectively with the following characteristics: 1) universal-revolute-spherical, 2) spherical-revolute-spherical and 3) spherical-spherical-spherical;

8) To use simulation technique to predict vertical ground reaction forces generated during gait in pregnant women.

1.4 Thesis Overview

Following the present introduction (chapter 1), this thesis embraces a compilation of four papers, as follows.

The aim of the first study (chapter 2) was to describe the gait pattern during the second trimester of pregnancy and give an orientation for biomechanical modeling of the lower limb of pregnant women. This model construction revealed to be appropriate to describe gait during second trimester of pregnancy.

The aim of the second study (chapter 3) was to analyze the effect of the increased mass in the trunk associated to pregnancy on the lower limb and pelvis, during walking task, on temporal-distance parameters, joint range of motion (ROM) and moments of force (M_f), by comparing second trimester pregnant women group to non-pregnant women group.

The aim of the third study (chapter 4) was to compare three different approaches on joint modelling of the ankle, knee and hip, as follows: 1) universal-revolute-spherical (URS); 2) spherical-revolute-spherical (SRS), and; 3) spherical-spherical-spherical (SSS), in regard of the gait of two groups of participants: 1) second trimester pregnant women, and; 2) non pregnant women.

The aim of the fourth study (chapter 5) was to predict vertical ground reaction forces (GRF) generated during gait, measured in the three trimesters of pregnancy and post-partum, by means of a contact model (simulation).

Finally, in chapter 6, a general discussion and main conclusions of the thesis are presented, as well as few recommendations for future research.

The study was approved by ethics committee of Faculty of Human Kinetics, University of Lisbon.

1.5 References

Block, R.A., Hess, L.A., Timpano, E.V. & Serlo, C., (1985). Physiological changes in the foot in pregnancy. Journal of the American Podiatric Medical Association, 75, 297–299.

Brett, M. & Baxendale, S., (2001). Motherhood and memory: a review. Psychoneur-oendocrinology 26, 339–362.

Brüggemann, G.P. (2005). Mechanical Loading of Biological Structures and Tissue Response. In Rodrigues, H; Cerrolaza, M; Doblaré, M; Ambrósio, J & Viceconti, M (Eds), Proceedings of the ICCB 2005 – II International Conference on Computational Bioengineering, 1-2,25-26.

Butte, N., Ellis, K., Wong, W., Hopkinson, J. & Smith, E. (2003). Composition of gestational weight gain impacts maternal fat retention and infant birth weight. American Journal of Obstetrics and Gynecology, 189(5), 1423-1432.

Calganeri, M., Bird, H.A. & Wright, V. (1982). Changes in joint laxity occurring during pregnancy. Annals of the Rheumatic Diseases, 41, 126–128.

de Groot, R.H., Adam, J.J. & Hornstra, G. (2003). Selective attention deficits during human pregnancy. Neuroscience Letters, 340, 21–24.

Dumas, G.A., Reid, J.G., Wolfe, L.A., Griffin, M.P. & McGrath, M.J. (1995). Exercise, posture, and back pain during pregnancy. Clinical Biomechanics, 10, 98–103.

Dunning, K., LeMasters, G., Levin, L., Bhattacharya, A., Alterman, T. & Lordo, K. (2003). Falls in workers during pregnancy: risk factors, job hazards, and high risk occupations. American Journal of Industrial Medicine, 44, 664–672.

Foti, T. ; Davids, J. & Bagley, A. (2000). A Biomechanical Analysis of Gait During Pregnancy. The Journal of Bone and Joint Suergery, Incorporated, 82-A(5), 625-632.

Hainline, B. (1994). Low back pain in pregnancy. Advances in Neurology, 64, 65–76.

Lu, T.W. & O'Connor, J.J.(1999). Bone position estimation from skin marker coordinates using global optimization with joint constraints. Journal of Biomechanics, 32, 129-134.

McNitt-Gray, J.L. (1991). Biomechanics related to exercise in pregnancy, Exercise in pregnancy 2nd edition Williams & Wilkins, Baltimore 133-140.

Must, A. & Strauss, R. (1999). International Journal of Obesity and Related Metabolic Disorders, 23,S2-11.

Nicholls, J.A. & Grieve D.W. (1992). Performance of Physical Tasks in Pregnancy. Ergonomics, 35,301-11.

Perry, J. (1992) Gait Analysis. Normal and Pathological Function. Thorofare, New Jersey: SLACK Inc.

Peterson, D.R. & Bronzino J.D. (2008). Biomechanics: Principles and Applications (2nd ed.). Boca Raton: CRC Press.

Pitkin, R. (1976). Nutritional support in obstetrics and gynecology. Clinical Obstetrics and Gynecology, 19(3), 489 – 513.

Snijders, C.J., Seroo, J.M., Snijder, J.G. & Hoedt, H.T. (1976). Change in form of the spine as a consequency of pregnancy. In: Proceedings for the 11th International Conference on Medical and Biological Engineering. 670-671. Ottawa, ONT, CAN.

Taggart, N., Holliday, R., Billewicz, W., Hytten, F. & Thomson, A. (1967). Changes in skinfolds during pregnancy. British Journal of Nutrition, 21(2), 439-451.

Winter, D.A. (2009). Biomechanics and Motor Control of Human Movement (4th ed.). New Jersey: John Wiley & Sons.

Chapter 2

2 Biomechanical Model for Kinetic and Kinematic Description of Gait During Second Trimester of Pregnancy to Study the Effects of Biomechanical Load on the Musculoskeletal System [1]

[1] Published as: Aguiar L, Santos-Rocha R, Branco M, Vieira F, Veloso AP (2014). Biomechanical model for kinetic and kinematic description of gait during second trimester of pregnancy to study the effects of biomechanical load on the musculoskeletal system. Journal of Mechanics in Medicine and Biology (JMMB), 14, 1450004. DOI: 10.1142/S0219519414500043 (Appendix 2)

2.1 Abstract

Walking is daily physical activity and a common way of exercise during pregnancy, but morphological changes can modify the gait pattern. Biomechanical models can help in evaluating joint mechanical loads, kinetics and kinematics during gait and provide the knowledge about the pattern of movement. This study aims to describe the gait pattern during the second trimester of pregnancy and give an orientation for biomechanical modeling of the lower limb of pregnant women. An inverse dynamics three-dimensional approach was used. Ankle and hip joint seem to be more overloaded, mainly in the sagittal and frontal plane, respectively. Results showed that pregnant women have a similar walking pattern to the normal gait. This model construction revealed to be appropriate to describe gait during the second trimester of pregnancy.

Keywords: Modelling, gait, pregnancy, kinematics, kinetics.

2.2 Introduction

Locomotor system models are an increasingly common way of studying human gait and a variety have been developed to perform specific functions (Zajac, Neptune & Kautz, 2002; 2003) and specific conditions. Human body can adjust to different conditions and the gait pattern may change. Muscle function may be reorganized to provide body acceleration and therefore modify the gait pattern, which contribute to an overload on the musculoskeletal system, causing discomfort and even pain in lower back, hip and leg (Foti et al., 2000).

During pregnancy, women are subject to morphological, physiological and hormonal changes, which can lead to adaptations in gait. These changes can include weight gain, swelling of the arms and legs, increased ligamentous laxity, decreased neuromuscular control and coordination, decreased abdominal muscle strength, increased spinal lordosis, altered biomechanical parameters, an anterior displacement in the location of the center of mass, and changes in mechanical loading and joint kinetics (Block et al., 1985; Brett & Baxendale, 2001; Calganeri et al., 1982; deGroot et al., 2003; Dumas et al., 1995; Dunning et al., 2003; Hainline, 1994; McNitt-Gray, 1991; Snijders et al., 1976). Other studies demonstrated that more than 50% of women reported swelling of the foot, ankle, and leg, unsteady gait, increased foot width and hip pain (Ponnapula & Boberg, 2010). From week 20th changes due to pregnancy, as weight gain and skinfolds increase, are already well established and are clearly visible (Pitkin, 1976; Taggart et al., 1967; Butte et al., 2003). In what concerns weight gain, according to the Institute of Medicine (IOM) and the National Research Council (NRC) (2009) recommendations, the body mass of a woman with avarage body mass index (BMI) may increase between 11.5 and 16 kg.

Foti et al. (2000), used three-dimensional (3-D) motion data of 15 women during the end of the third trimester, and again one year after delivery. Kinematic and kinetic parameters were compared between the two conditions. It was found that the gait pattern was remarkably changed during pregnancy. Concerning kinetic parameters, it was observed an increase in the hip moment and power in the frontal and sagittal planes. The general pattern of gait kinematics in pregnant women is similar to that of no pregnant women, however, at faster speed, pregnant women have difficulty in coordinating between the thorax and pelvis. Pregnant women prefer to walk with slower speed, but this fact cannot be explained as energy economy mode, since walk more slowly than the comfort speed, requires more energy (Wu et al., 2004).

Walking is daily physical activity and a common way of exercise during pregnancy, but morphological changes can modify the gait pattern. Biomechanical models can help in evaluating joint mechanical loads, kinetics and kinematics during gait and provide the knowledge of the movement patterns. However, such modifications may become a concern in biomechanical models construction. Joint centers are calculated based on the marker's locations and the swelling of the segments may increase the error associated with the marker placement. In the pelvis, marker locations have special attention due the mass distribution in the anterior trunk moving forward the position of the center of mass. Another concern is to minimize soft tissue artifact produced by segments mass vibration.

2.3 Objectives

This study aims to describe the gait pattern during the second trimester of pregnancy and give an orientation for 3-D biomechanical modeling of the lower limb of pregnant women.

2.4 Materials and Methods

Some methodological problems may be associated with the kinematic and kinetic variables estimations due to anthropometric characteristics as marker locations for three-dimensional segments reconstruction and markers associated to skin movement. The aim of the present is to contribute to add more information to biomechanical gait analysis regarding the specific conditions of pregnancy, provided by a 3-D biomechanical model constructed based on two main processes: inverse kinematics and inverse dynamics methods.

2.4.1 Subjects

Eighteen pregnant women (PW) with 27.3 weeks (second trimester) mean gestational age, mean chronological age of 32.6 years, 68.2 kg of weight, 1.60 m of height and mean BMI of 26.3 kg/m^2 (table 1), participated in this study.

Table 1 - Characterization of sample group (mean ± standard deviation): age, weight, height, body mass index (BMI) and gestational age.

	Age (years)	Weight (kg)	Height (m)	BMI (kg/m^2)	Gestational age (weeks)
PW	32.6±2.7	68.2±7.3	1.6±0.1	26.3±2.6	27.3±3.0

2.4.2 Data Collection and Processing

Anthropometric data (weight and height) were collected following ISAK recommendations, to calculate de body segments masses and inertia moments. Reflective spherical markers were placed on anatomical points according to the defined marker setup protocol (figure 1). Motion capture was performed with an optoelectronic system of twelve cameras Qualisys (Oqus-300) operating at a frame rate of 200 Hz, synchronized with two force platforms (Kistler AG, Winterthur, Switzerland) and one AMTI (Advanced Mechanical Technology, Inc, Watertown, MA), which collected ground reaction force (GRF) data. The participants walking during three non-consecutive minutes at a comfortable speed, with a break of thirty seconds between each trial.

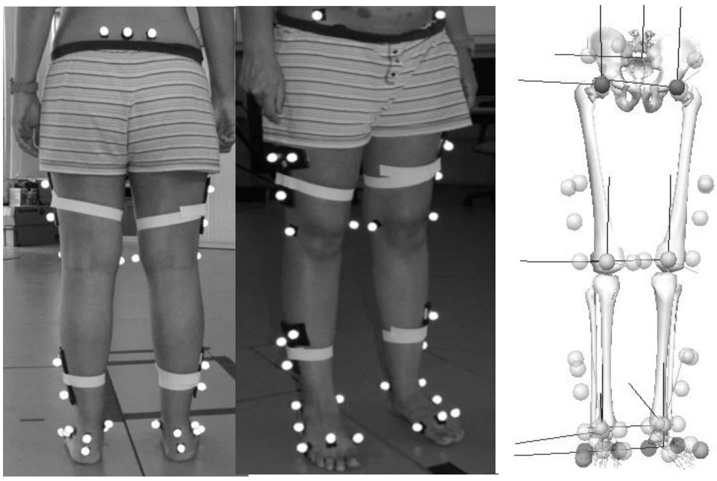

Figure 1 - Marker setup used for motion captures and lower limb and pelvis constructions: five markers for pelvis, five markers for thigh, seven markers for shank an eight markers for foot a) Posterior setup. b) Anterior setup. c) Reconstructed biomechanical model in Visual 3-D.

Gait cycles were selected based on the force platforms and on the quality of both GRF. Cycle started at the right heel strike (RHS) and finished at the next RHS.

A static trial was performed (figures 1 and 2) which enables segments reconstruction and allows scaling the model, defining of the joint centers and the segment coordinate system (SCS) (figure 2).

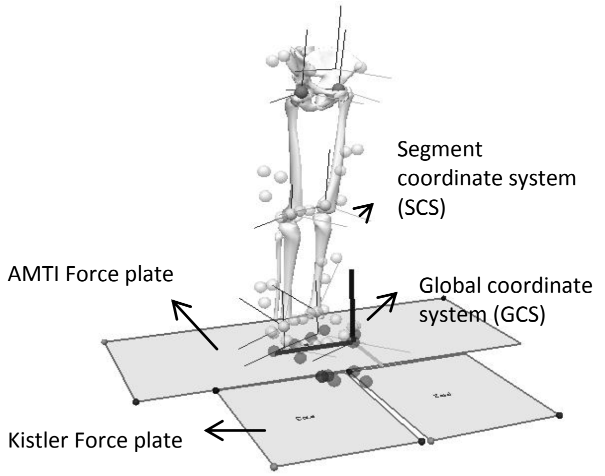

Figure 2 - Static trial to set all the segments as a reference position and to model construction based on the marker setup, in Visual 3-D. Subject was located near to the Laboratory or Global Coordinate System (LCS), on the calibrated volume.

To create the local SCS it was necessary to establish the frontal plane using border targets: medial-distal, medial-proximal, lateral-distal and lateral-proximal. The SCS was created using the midpoints between the medial and lateral targets to define the distal and proximal end points. The SCS Z-axis was determined by the unit vector directed from the distal segment end point to the proximal segment end. The SCS Y-axis was determined by the unit vector that is perpendicular to both the frontal plane and the Z axis. Finally, the SCS X-axis was determined by the application of the vector product and as consequence using the right hand rule. The SCS Z-axis is directed from distal to proximal, the SCS Y-axis is directed from posterior to anterior, and the SCS X-axis is medial-lateral in orientation. For the pelvis, the (x-y) plane of the segment coordinate system is defined as the plane passing through the right and left ASIS (anterior superior iliac spine) markers, and the mid-point of the right and left PSIS (posterior superior iliac spine) markers.

Segments were defined using a proximal and a distal end point, measured by the middle distance between lateral and medial markers. Foot segment was defined by first and fifth metatarsals lateral and medial tibia malleolus. For the shank it was used markers placed in tibia malleolus and in femur condyles. Pelvis was built based on CODA model (figure 3a), defined using the anatomical locations of the right and left ASIS and the right and left PSIS. This model allows the estimation of the right and left hip joint center (LHJC; RHJC) using the regression equation (equations 1 and 2) proposed by Bell, Pedersen and Brand (1989, 1990).

$$\mathrm{RHJC} = (0.36 * \mathrm{ASIS_{Distance}}, -0.19 * \mathrm{ASIS_{Distance}}, -0.3 * \mathrm{ASIS_{Distance}}) \quad \text{(Eq. 1)}$$

$$\mathrm{LHJC} = (-0.36 * \mathrm{ASIS_{Distance}}, -0.19 * \mathrm{ASIS_{Distance}}, -0.3 * \mathrm{ASIS_{Distance}}) \quad \text{(Eq.2)}$$

Tracking markers (figure 3b) were also used allowing the segment to follow their coordinates to replicate the performed motion. In the foot, markers were placed in the second metatarsal, lateral side and in the heel, and a three marker cluster was strapped around the leg and another one around the thigh. In the pelvis, it was added a marker placed in the middle point between the PSIS markers to ensure at least three tracking markers.

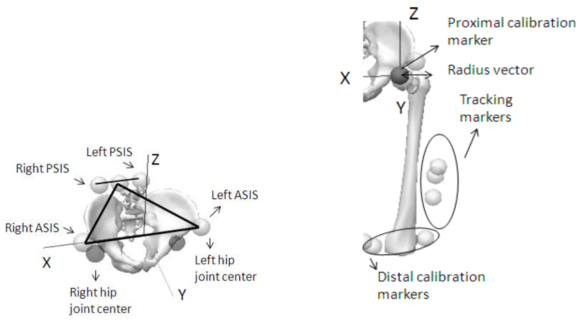

Figure 3 - a) CODA model and for pelvis construction in Visual 3-D: ASIS and PSIS are necessary to determine pelvis SCS. b) Setup for body segments construction, using the distal and proximal markers and tracking markers, which let the segment follow its coordinates.

The position and orientation of the local SCS is described with respect to the laboratory coordinate system (LCS) (figure 4). The location of one point in the LCS is given by equation 3, where T is the rotation matrix from the SCS to the LCS and O is the translation between coordinate systems.

$$P = TA + O \qquad \text{(Eq. 3)}$$

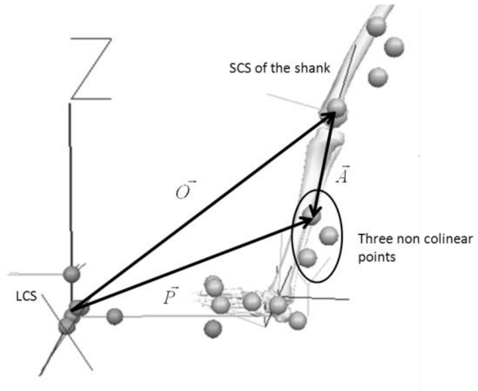

Figure 4 - Determination of A point from measurement of P in SCS regarding the transformation matrix T and the translation O. LCS is the laboratory coordinate system or global coordinate system (GCS). Model reconstructed in Visual3-D.

Thus, A is a point determined from measurement of P at this position (equation 4). The matrix T and translation vector O, are computed in all instants, provided that for at least three noncolinear points and can be found by minimizing the sum of squares error (equation 5), under the orthonormal condition (equation 6), where m is equal to the number of targets on the segment. Langrange multipliers are used, to find the maxima and minima of the function subject to constraints, given by equation 7 (Spoor & Veldpaus, 1980).

$$A = T^{-1}(P - O) \qquad \text{(Eq. 4)}$$

$$\sum_{i=1}^{m}((P_i - TA_i) - O)^2 \qquad \text{(Eq. 5)}$$

$$T^{-1}T = T^{T}T = I \qquad \text{(Eq. 6)}$$

$$g(T) = T^{T}T - I = 0 \qquad \text{(Eq. 7)}$$

All the markers coordinate data were interpolated using third degree polynomial to fulfill gap displacements. To reduce the noise the motion data was filtered, using a low pass Butterworth filter, with a cutoff frequency of 15 Hz (Winter, 1991). Each marker location was recognized in the static collection, which is as a reference position and must be set in each frame.

Due to the skin movement, the markers move relatively to the bone, modifying the initial configuration and therefore reproducing noisy motion data. Moreover, the segments are considered linked rigid bodies and joints have six degrees of freedom (6DOF), making them independent from each other and allowing more movement. Joint degrees of freedom were manipulated through the Inverse Kinematics tool in order to achieve the kinematic that can reproduce better the real motion. The inverse kinematics is based on the method of global optimization (Lu & O'Connor, 1999) and allows finding an optimal position for the set of segments, which constitute the model on each frame, the differences between the measured coordinates in the static position and the motion measured coordinates are minimized by the method of least squares. Thus, in the ankle, rotations in sagittal and frontal plane were allowed (universal joint), in the knee only the sagittal rotation was unrestricted (revolute joint) and in the hip all the rotations were used (spherical joint). Translations were locked for the three joints. These were the default available degrees or freedom provided by the software Visual 3-D C-Motion, Inc. Frame by frame, the recorded motion reproduces the model joint angles set in a configuration that best represents the experimental kinematics.

Given a set of measured marker coordinates *P* on a data frame, the global optimization is to find a set of generalized coordinates ξ so that the following error function (equation 9) is minimized, where W is a positive-definite weighting matrix, P'(ξ) is the corresponding set of marker coordinates calculated by the transformation given by equation 8, from the segment frame to laboratory frame.

$$\mathrm{P}'(\xi) = \mathrm{T}(\xi)\mathrm{P}^{*} \quad \text{(Eq. 8)}$$

$\mathrm{T}(\xi)$ is the transformation matrix from the SCS to LCS. This condition also maintains the rigid body condition and the integrity of the model once joint constraints are part of the model.

$$f(\xi) = [P - P'(\xi)]^{T} W [P - P'(\xi)] \qquad \text{(Eq. 9)}$$

W is a ($3m_i$ x $3m_i$) weighting matrix (equation 10) assigned to the each segment to reflect the error distribution among the m_i markers. To each segment is given different weighting factor reflecting its average degree of soft tissue artifacts. In this study it was used 1 for all segments weights.

$$W = \begin{bmatrix} W0 & 0 & 0 & 0 \\ 0 & W1 & 0 & 0 \\ 0 & 0 & \ddots & 0 \\ 0 & 0 & 0 & Wn \end{bmatrix} \quad \text{(Eq. 10)}$$

For each segment, the equation 11 is solved to find the minimum value, under the orthogonal constrain $R^TR = I$, where x_i and y_i are position vectors of marker i at the reference and global positions, respectively. R is the rotation matrix, v is the translation vector and m is the number of markers.

$$f = \sum_{i=1}^{m}(Rx_i + v - y_i)^T(Rx_i + v - y_i) \qquad \text{(Eq. 11)}$$

The minimum value of f is the segmental residual error (e), a measure of the markers distortion which is mainly due to soft tissue artifacts. The weighting matrix W_i is defined as equation 12.

$$W_i = \frac{1}{e_i}I \qquad \text{(Eq. 12)}$$

This calculation is proposed for the kinematics of a rigid body between two sequential positions, where the first one works as the reference (Cappello, Francesco, Palombara & Leardini, 1996; Challis, 1995; Spoor & Veldpaus, 1980). The segments velocities and accelerations were obtained by derivation of the new position equations.

The inverse dynamics method $\tau = ID(model, q, \dot{q}, \ddot{q})$ (figure 5) was used to calculate the forces and moments produced by the muscular action, requiring the estimated mass and inertia values, body segments linear and angular accelerations, as the data from external forces measured by the force plates. The weights and locations of the centers of mass for each body segments considered were calculated using the regression equations of Dempster and inertia moments using inertial properties based on their shape (Hanavan, 1964). To determine the joints net moments and forces, the equations of motion was iteratively solved, considering the equilibrium dynamic and boundary conditions.

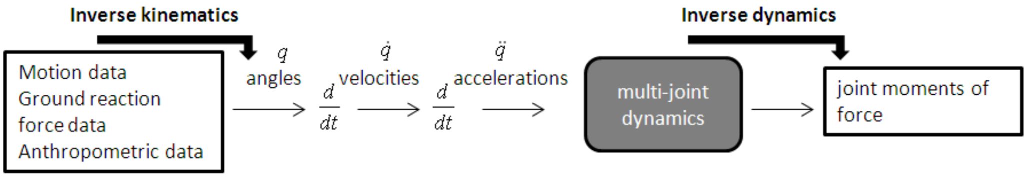

Figure 5 - Workflow of the inverse problem using inverse kinematic and dynamic solutions.

The joint torques calculation was performed during the right foot stance phase from the heel strike (heel touches the ground) to toe off (foot leaves of the ground), completing approximately 60% of gait cycle. The cycle starts with the left foot contact with the platform and ends in the next heel contact of the same foot with the ground.

As figure 6 shows, the first step is defining the location of the center of mass of each segment to identify the position vectors relatively to the SCS given by the vectorial summation (equation 13).

$$R_i = r_i + r_i^{'} + r_{i-1} \qquad \text{(Eq. 13)}$$

Then, we must identify the proximal forces and torques due to inertia and joints and finally include external forces and couples (figure 6).

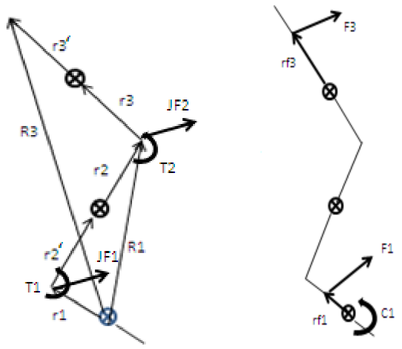

Figure 6 - a) Definition of vectors and location of center of mass, proximal forces and torques. b) Identification of external forces and couples.

The proximal joint reaction force is computed in the global coordinate system (GCS), using an iterative algorithm, which allows any applied external force on segments (equation 14), where m_i is the mass of segment i, a_i is the acceleration of segment i, n the number of distal segments connected in chain, q the number of external forces and F_q, the applied external forces.

$$F_{proximal} = \sum_{i=1}^{n} m_i(a_i + g) + \sum_{j=1}^{q} F_q \qquad \text{(Eq. 14)}$$

The proximal moment due to the inertial forces and applied moments at the joint is calculated by equation 15, where P_j is the vector from application of the external force to the proximal joint, p is the number of external couples (τ), R_i is the distance from center of gravity of each distal segment to proximal joint, $A_i = m_i(a_i + g)$ and $C_i = T^{-1}C_i^{'}$. T is the matrix that transforms inertial torque form SCS to LCS based on the motion data.

$$M_{proximal} = \sum_{i=1}^{n}(C_i + R_i\, A_i) + \sum_{j=1}^{q}\left(P_j F_q\right) + \sum_{k=1}^{p} \tau_k \qquad \text{(Eq. 15)}$$

The couple acting on a segment due the inertia is determined by equation 16.

$$M_i = C_i + C_{i-1} + r_i \times F_i + r_i^{'} \times F_{i-1} \quad \text{(Eq. 16)}$$

Is important to remember that the T matrix, transforms the inertial torque $C_i^{'}$ from the SCS into C_i in the LCS. So, the proximal couple is computed at the proximal end of a segment in SCS is given by equation 17, where I_i is the inertia moment of the segment, α is the angular acceleration and ω is the angular velocity for the angular moment calculation $I_i\omega_i^{'}$ (Siegel, Kepple & Stanhope, 2004).

$$C_i^{'} = I_i\alpha_i^{'} + \omega_i^{'}(I_i\omega_i^{'}) \quad \text{(Eq. 17)}$$

Joint angles and moments of force (M_f) were calculated in order to a SCS, using a Cardan sequence of rotations x-y-z, for the ankle, knee and hip (Cole, Nigg, Ronsky, & Yeadon, 1993). However, as Baker (2001) suggested, pelvis angles were computed using z-y-x rotation to descript the movement relatively to the laboratory, representing axial rotation, obliquity and tilt (figure 7).

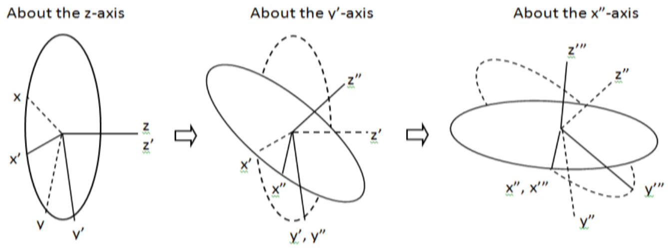

Figure 7 - Cardan sequence for pelvis angles processing, Z-Y'-Z'' (adapted from Ying & Kim, 2002).

For ankle, shank and thigh segments x rotation was around the medial-lateral axis, y was related with anterior- posterior axis and z was around the vertical axis. Based on this sequential rotations, T matrix is (equation 18):

$$T = \begin{bmatrix} cos\beta cos\gamma & cos\gamma sin\beta sin\alpha + sin\gamma cos\alpha & sin\alpha sin\gamma - cos\alpha sin\beta cos\gamma \\ -sin\gamma cos\beta & cos\alpha cos\gamma - sin\alpha sin\beta sin\gamma & sin\gamma sin\beta cos\alpha + cos\gamma sin\alpha \\ sin\beta & -cos\beta sin\alpha & cos\alpha cos\beta \end{bmatrix} \quad \text{(Eq. 18)}$$

The T matrix as a final representation based on motion data is shown in equation 19:

$$T = \begin{bmatrix} T_{11} & T_{12} & T_{13} \\ T_{21} & T_{22} & T_{23} \\ T_{31} & T_{32} & T_{33} \end{bmatrix} \quad \text{(Eq. 19)}$$

Using both matrices (equations 18 and 19) it was possible to obtain the equations 20 to 22:

$$\beta = sin^{-1}(T_{31}) \quad \text{(Eq. 20)}$$

$$\propto = -sin^{-1}(\frac{T_{32}}{cos\beta}) \quad \text{(Eq. 21)}$$

$$\gamma = -sin^{-1}(\frac{T_{21}}{cos\beta}) \quad \text{(Eq. 22)}$$

To calculate ankle, knee and hip joint angles and M_f, three-dimensional biomechanical models were constructed, using the software Visual 3-D C-Motion, Inc.

2.4.3 Statistical Analysis

The results are based on five cycles per subject. Each cycle starts in the Right Heel Strike (RHS) and finish in the nest RHS. RHO and Right Toe Off (RTO) were created events to establish temporal-distance parameters. The average joint angles and moments curves were obtained for each participant as the mean curve for the group. Both angular displacement and M_f data were normalized to time cycle. The M_f were normalized to body mass.

2.5 Results

2.5.1 Time and Distance Parameters

In what concerns to the temporal-distance variables, speed has a mean value of 1.16±0.12 m/s. Stride width was 0.10±0.02 m. Left and right step lengths were 0.62±0.05 with a duration of 0.54±0.03 s. Total cycle time was 1.07±0.06 s while left and right stance time were 0.65±0.04 s. Double limb support time was 0.22±0.03 s.

2.5.2 Kinematic and Kinetic Data

The range of motion (ROM) and M_f were analyzed according to the allowed degrees of freedom. Maximum dorsiflexion and ankle plantar flexion angles were 12.16±3.05 and -14.34±5.10 deg, respectively. Sagittal ankle plane ROM was 26.50±3.65 deg. Peak ankle dorsiflexion M_f was 1.21±0.16 N.m/kg and the peak plantar flexion M_f was -0.52±0.32 N.m/kg. In the frontal plane, maximum ankle eversion was -6.71±3.06 deg and maximum inversion was 7.18±3.49 deg; the correspondent ROM was 12.02±2.04 deg. Ankle eversion peak M_f was -0.05±0.02 N.m/kg and the inversion peak was 0.10±0.04 N.m/kg (figure 8).

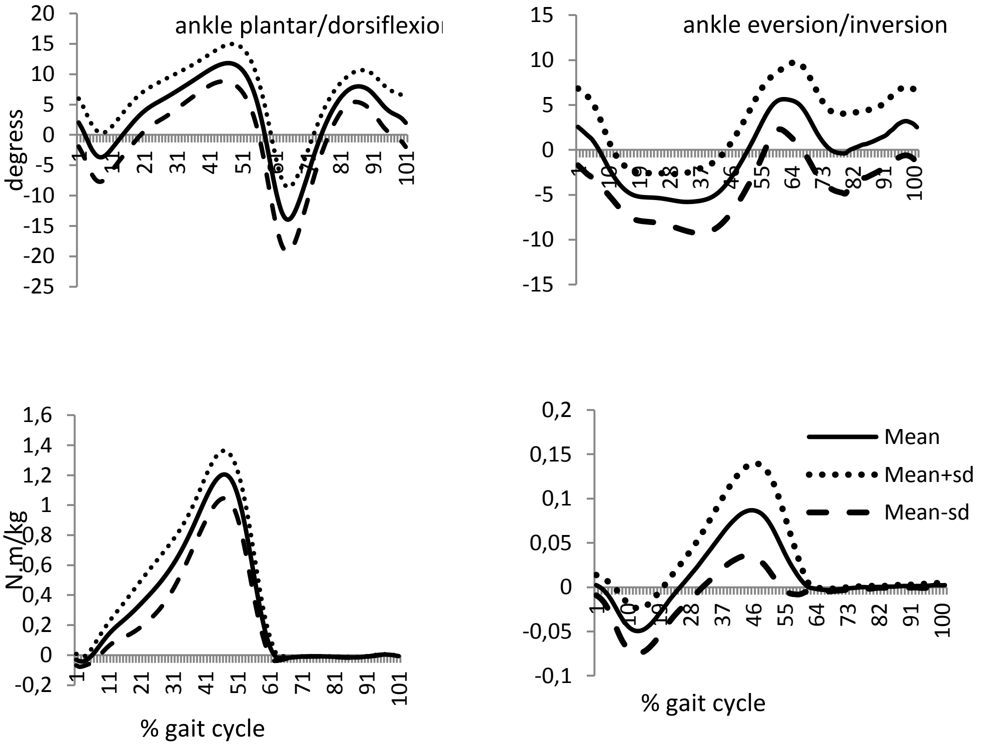

Figure 8 - Mean and standard deviations curves of ankle range of motion - ROM (above figures) in sagittal (left side) and frontal (right side) planes and the respective M_f (lower figures).

In the knee joint, flexion maximum angle was 63.62±4.87 deg while the extension was -0.44±3.90 deg. In sagittal plane knee ROM was 64.06±5.39 deg. The peak flexion knee M_f was 0.53±0.20 N.m/kg and the peak extension M_f was -0.17±0.09 N.m/kg (figure 9).

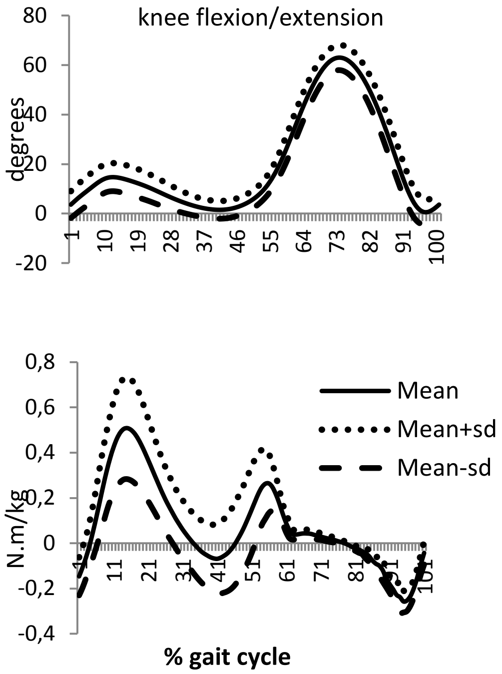

Figure 9 - Mean and standard deviations curves of knee ankle range of motion - ROM (above figure) in sagittal plane and the respective M_f (lower figure).

Maximum hip flexion reached 39.55±5.00 deg and maximum hip extension was -4.43±5.15 deg. Flexion/ extension hip ROM was 43.98±3.08 deg. Peak flexion and extension M_f were 0.43±0.09 N.m/kg and -0.79±0.18 N.m/kg, respectively. The mean value of 11.92±2.38 deg was observed for hip maximum adduction and 7.77±4.95 deg for maximum abduction. The adduction/abduction ROM was 16.69±4.34 deg. Peak adduction M_f was 0.76±0.11 N.m/kg and peak abduction M_f (valley) was 0.52±0.09 N.m/kg. In the z axis, hip maximum internal rotation was 5.30±5.57 deg and maximum external rotation was -11.46±7.96 deg; the correspondent ROM was 16.76±5.36 deg. Peak internal rotation M_f was -0.12±0.05 N.m/kg and external peak M_f reached 0.08±0.04 N.m/kg (figure 10).

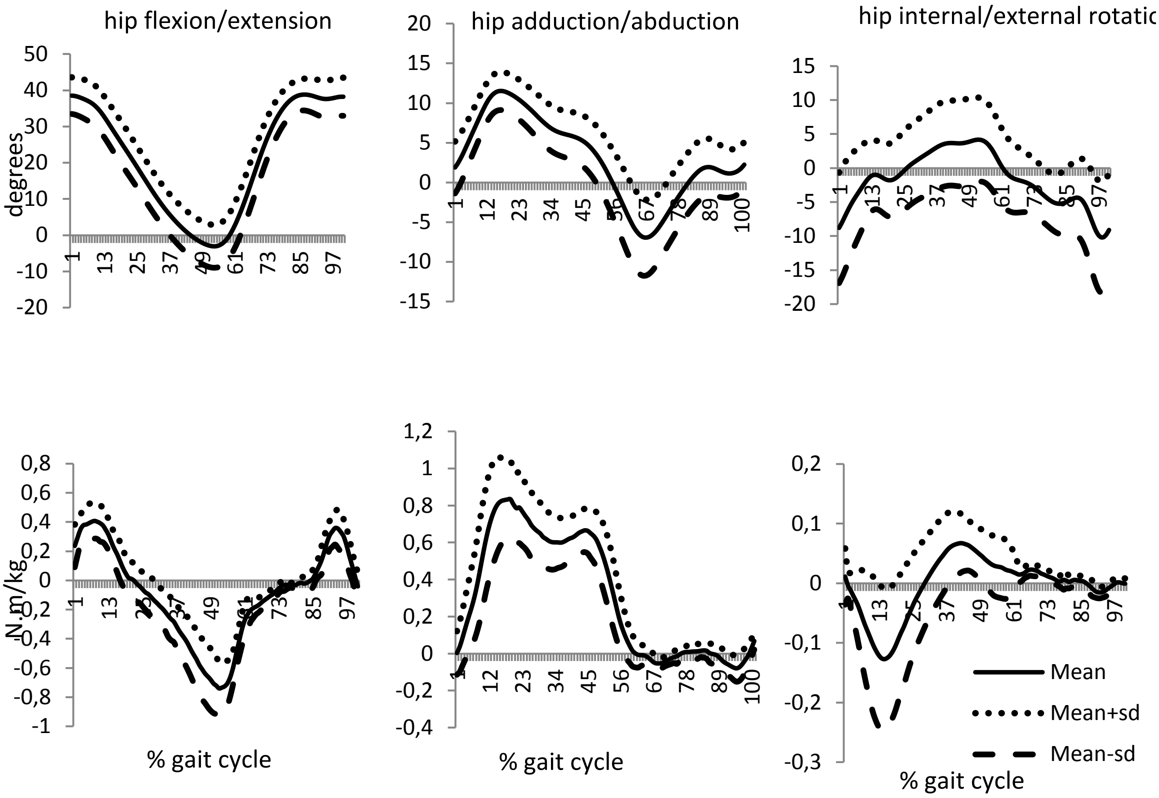

Figure 10 - Mean and standard deviations curves of hip range of motion - ROM (above figures) in sagittal (left side), frontal (middle) and transversal (right side) planes and the respective M_f (lower figures).

Maximum anterior pelvic tilt was 16.88±4.13 deg while maximum posterior pelvic tilt was 14.07±4.36 deg. Pelvic tilt ROM was 2.81±0.99 deg. In the frontal plane, maximum pelvic left and right obliquities were 7.13±2.15 deg and -6.39±3.16 deg, respectively. Pelvic obliquity ROM was 13.52±3.49 deg. Maximum transversal left rotation was 5.91±2.88 deg and right maximum rotation was -7.15±3.12 deg. Pelvic rotation ROM was 13.05±4.30 deg (figure 11).

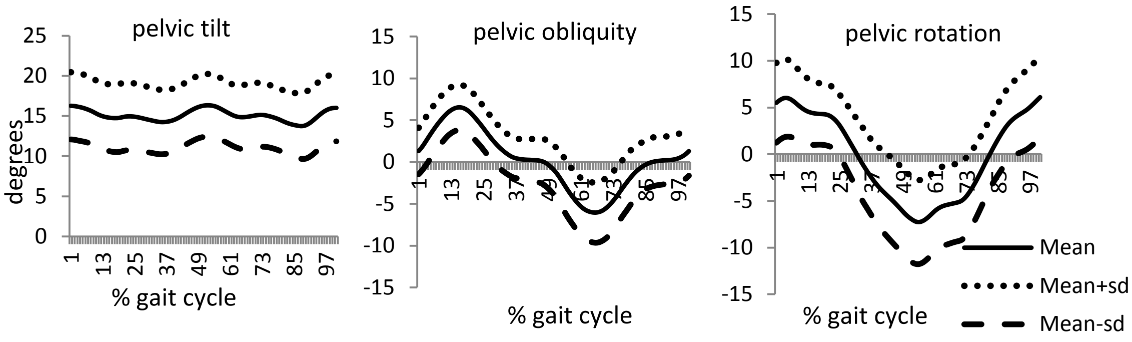

Figure 11 - Mean and standard deviations curves of pelvis range of motion - ROM in sagittal (left side), frontal (middle) and transversal (right side) planes.

2.6 Discussion

It was possible to observe similar angles and joint moment's curves patterns as the maximum ROM and peak M_f values, compared with other studies related to pregnancy (Foti et al., 2000). Concerning total gait cycle, stance phase was around 60% (Vaughan, Davis & O'Connor, 1992; Perry, 1992), similar to the normal gait pattern.

In the ankle joint, the plantar flexion peak moment occurred during the early stance phase (figure 8). The dorsiflexor moment increased as the body moves in the forward direction. Peak inversion M_f increases in late stance phase.

Immediately after the first foot contact with the ground, there was a reaction vector anterior to the knee and hip joints, producing during the stance initial phase an extensor moment, which decreased with the alignment of this vector with the leg and thigh segments (Perry, 1992).

The hip peak flexion moment occurred in the beginning of the stance phase (while the extensor moment of the same joint, was recorded on the final stance phase. Adduction peak M_f revealed an higher value relatively to the other planes during the earlier stance phase (figure 10). This can be related a larger base of gait during walking, which can improve locomotor stability (Foti et al., 2000; Bird, Menz & Hyde, 1999). Abduction peak moment was recorded in the middle of the stance phase. Internal and external peaks M_f registered lower values (figure 10).

Pelvic anterior tilt was maximum in the end of the stance phase, while left maximum obliquity was maximum in the earlier stance phase while right obliquity was maximum immediately after the RTO. As pelvic obliquity, transversal rotation starts for the left side, reaching the maximum angle in the end of the stance phase and rotating to the right side (figure 11).

The speed wasn't imposed so the obtained mean value may reflect the comfort walking speed for a second trimester pregnant woman. Wu et al. (2004) found that pregnant prefer walking with lower velocities. This can be related with fatigue associated to the weight gain and its distribution, hormonal changes and a consequent less efficient gait mechanics.

2.7 Conclusions

Regarding kinetics and when compared to other studies, ankle and hip joint seem to be more overloaded, mainly in the sagittal and frontal plane, respectively. The results showed that pregnant women have a similar walking pattern to the normal gait, when comparing to other studies. Using this methodology and data from non-pregnant women could bring more useful information's about gait abnormalities in this population and concerning different stages of pregnancy. This study aimed to describe joint ROM and M_f during second trimester of pregnancy using a 3-D biomechanical model. There were some concerns about the marker placements due the pregnant anthropometric characteristics, mainly in the pelvis. However, this model construction revealed to be appropriate to describe gait during second trimester of pregnancy.

2.8 References

Baker, R. (2001). Pelvic angles: a mathematically rigorous definition which is consistent with a conventional clinical understanding of the terms. Gait and Posture 13, 1-6.

Bell, A.L., Pedersen, D.R. & Brand, R.A. (1990). A Comparison of the Accuracy of Several hip Center Location Prediction Methods. Journal of Biomechanics. 23, 617-621.

Bell, A.L., Pederson, D.R., & Brand, R.A. (1989). Prediction of hip joint center location from external landmarks. Human Movement Science. 8, 3-16.

Bird, A., Menz, H. & Hyde, C. (1999). The effect of pregnancy on footprint parameters. A prospective investigation. Journal of the American Podiatric Medical Association, 89(8): 405-409.

Block, R.A., Hess, L.A., Timpano, E.V. & Serlo, C., (1985). Physiological changes in the foot in pregnancy. Journal of the American Podiatric Medical Association, 75, 297–299.

Brett, M. & Baxendale, S., (2001). Motherhood and memory: a review. Psychoneuroendocrinology 26, 339–362.

Butte, N., Ellis, K., Wong, W., Hopkinson, J. & Smith, E. (2003). Composition of gestational weight gain impacts maternal fat retention and infant birth weight. American Journal of Obstetrics and Gynecology, 189(5), 1423-1432.

Calganeri, M., Bird, H.A. & Wright, V. (1982). Changes in joint laxity occurring during pregnancy. Annals of the Rheumatic Diseases, 41, 126–128.

Cappello, A., Francesco, P., Palombara, L. & Leardini, A. (1996). Optimization and smoothing techniques in movement analysis. International Journal of Biomedical Computation, 41, 137-151.

Challis, J.H. (1995). A procedure for determining rigid body transformation parameters. Journal of Biomechanics, 28, 733-737.

Cole, G.K., Nigg, B.M., Ronsky, J.L. & Yeadon, M.R. (1993). Application of the Joint Coordinate System to Three-Dimensional Joint Attitude and Movement Representation: A Standardization Proposal. Journal of Biomechanical Engineering, 115, 344-9.

de Groot, R.H., Adam, J.J. & Hornstra, G. (2003). Selective attention deficits during human pregnancy. Neuroscience Letters, 340, 21–24.

Dumas, G.A., Reid, J.G., Wolfe, L.A., Griffin, M.P. & McGrath, M.J. (1995). Exercise, posture, and back pain during pregnancy. Clinical Biomechanics, 10, 98–103.

Dunning, K., LeMasters, G., Levin, L., Bhattacharya, A., Alterman, T. & Lordo, K. (2003). Falls in workers during pregnancy: risk factors, job hazards, and high risk occupations. American Journal of Industrial Medicine, 44, 664–672.

Foti, T. ; Davids, J. & Bagley, A. (2000). A Biomechanical Analysis of Gait During Pregnancy. The Journal of Bone and Joint Suergery, Incorporated, 82-A(5), 625-632.

Hainline, B. (1994). Low back pain in pregnancy. Advances in Neurology, 64, 65–76.

Hanavan, E. (1964). A Mathematical Model for the Human Body. Technical Report, Wright-Patterson Air Force Base.

IOM & NRC. (2009). Weight gain during pregnancy. *Reexamining the guidelines.* Washington, D.C.: The National Academies Press.

Lu, T.W. and O'Connor, J.J.(1999). Bone position estimation from skin marker coordinates using global optimization with joint constraints. Journal of Biomechanics, 32, 129-134.

McNitt-Gray, J.L. (1991). Biomechanics related to exercise in pregnancy, Exercise in pregnancy 2nd edition Williams & Wilkins, Baltimore 133-140.

Perry, J. (1992) Gait Analysis. Normal and Pathological Function. Thorofare, New Jersey: SLACK Inc.

Pitkin, R. (1976). Nutritional support in obstetrics and gynecology. Clinical Obstetrics and Gynecology, 19(3), 489 – 513.

Ponnapula, P. & Boberg, J. (2010). Lower Extremity Changes Experienced During Pregnancy. The Journal of Foot & Ankle Surgery, 49, 452–458.

Siegel, K.L., Kepple T.M. & Stanhope S.J. (2004). Joint moment control of mechanical energy flow during normal gait. Gait and Posture, 19, 69-75.

Snijders, C.J., Seroo, J.M., Snijder, J.G. & Hoedt, H.T. (1976). Change in form of the spine as a consequency of pregnancy. In: Proceedings for the 11th International Conference on Medical and Biological Engineering. 670-671. Ottawa, ONT, CAN.

Spoor, C.W. & Veldpaus, F.E. (1980). Rigid body motion calculated from spatial coordinates of markers. Journal of Biomechanics, 13(4), 391- 393.

Taggart, N., Holliday, R., Billewicz, W., Hytten, F. & Thomson, A. (1967). Changes in skinfolds during pregnancy. British Journal of Nutrition, 21(2), 439-451.

Vaughan, C.L., Davis, B.L. & O'Connor, J.C. (1992). Dynamics of Human Gait. Champaign, IL: Human Kinetics.

Winter, D. (1991) The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological. Waterloo, Ontario: University of Waterloo Press.

Wu, W., Meijer, O., Lamoth, C., Uegaki, K., van Dieën, J., Wuisman, P., de Vries, J. & Beek, P.(2004). Gait coordination in pregnancy: transverse pelvic and thoracic rotations and their relative phase. Clinical Biomechanics, 19, 480-488.

Zajac, F.E., Neptune, R.R. & Kautz, S.A. (2002). Biomechanics and muscle coordination of human walking. Part I: Introduction to concepts, power transfer, dynamics and simulations, Gait Posture, 16, 215–232.

Zajac, F.E., Neptune, R.R. & Kautz, S.A. (2003). Biomechanics and muscle coordination of human walking. Part II: Lessons from dynamical simulations and clinical implications, Gait Posture, 17, 1–17.

Chapter 3

3 Comparison Between Overweight Due to Pregnancy and Due to Added Weight to Simulate Body Mass Distribution in Pregnancy [2]

[2] Under review: Aguiar L, Santos-Rocha R, Andrade C, Branco M, Vieira F, Veloso AP (2014). Comparison between overweight due to pregnancy and due to added weight to simulate body mass distribution during pregnancy. Gait & Posture.

3.1 Abstract

The assessment of biomechanical loading in the musculoskeletal system of the pregnant women is particularly interesting since they are subject to morphological, physiological and hormonal changes, which may lead to adaptations in gait. The purpose of this study was to analyze the effect of the increased mass in the trunk associated to pregnancy on the lower limb and pelvis, during walking, on temporal-distance parameters, joint range of motion and moments of force, by comparing a pregnant women group to a non-pregnant group, and to this group while carrying a 5kg additional load located in the abdomen and breasts during walking, to understand which gait adaptations may be more related with the increased trunk mass, or if may be more associated to physiological and hormonal changes. The subjects performed a previous twelve minute training adaption to the added load. To calculate ankle, knee and hip joint angles and moments of force, the three-dimensional biomechanical models were developed. The inverse dynamics method was used to estimate net joint moments of force. The anterior increased mass of the trunk associated with second trimester of pregnancy may influence some gait variables such as the left step time, left and right stance times, double limb support time, maximum hip extension, maximum pelvic right obliquity, pelvic obliquity range of motion, maximum transversal left rotation and peak hip flexion moments of force.

Keywords: Gait, trunk overweight, range of motion, moments of force, inverse dynamics, inverse kinematics.

3.2 Introduction

When walking the human body can adjust to different circumstances thereby changing the gait pattern. Muscle function can be reorganized to provide body acceleration and therefore modify the gait pattern, which may add an overload to the musculoskeletal system (Foti et al., 2000). The assessment of biomechanical loading in the musculoskeletal system of the pregnant women is of particular interest since it may be useful to exercise prescription in order to reduce the risk of discomfort or injure associated to overload. Also, as few biomechanical studies can be found in literature it is important to better understand the gait adaptations during pregnancy.

During pregnancy, women are subject to morphological, physiological and hormonal changes, which can lead to adaptations in gait. These changes include weight gain, anterior displacement of the location of the center of mass, increased ligamentous laxity, decreased neuromuscular control and coordination, swelling of the arms and legs, altered biomechanical parameters, changes in mechanical loading and joint kinetics, decreased of abdominal muscle strength and increased spinal lordosis (Block et al., 1985; Brett & Baxendale, 2001; Calganeri et al., 1982; Dumas et al., 1995; Dunning et al., 2003; McNitt-Gray, 1991; Snijders et al., 1976). In a previous study, more than 50% of the women reported swelling of the foot, ankle, and leg, unsteady gait, increased foot width and hip pain (Ponnapula & Boberg, 2010).

Pregnancy causes a gradual increase in body mass during the entire gestational period. According to the Institute of Medicine (IOM) and the American Research Council (NRC) recommendations, the body mass increase for a woman with a normal pre-pregnancy body mass index (BMI) may be, on average, between 11.5 and 16 kg. Its distribution depends on different components as the fetus growth, placenta, amniotic fluid, uterus, mammary gland, blood and adipose tissue (IOM, 2009).

When comparing pregnant and non-pregnant women, while walking on the treadmill at predefined speeds from 0.17 to 1.72 m/s, it was found that in pregnancy self-selected velocity was significantly smaller, while pelvis and thorax rotation amplitudes were slightly reduced. Pregnant women prefer to walk at lower speeds, but this fact cannot be explained as energy economy mode, since walking more slowly than the comfort speed, requires more energy (Wu et al., 2004). Gilleard (2013) found that sagittal plane range of motion for thoracic, pelvic and thoracolumbar spine, showed no linear trends with advancing pregnancy, during walking. But in post-birth, the thoracic segment range of motion increase and pelvic range of motion decrease in comparison to late pregnancy.

From the 20th week onward, changes due to pregnancy are already well established and are clearly visible (Pitkin, 1976; Butte et al., 2003). Foti et al. (2000) used three-dimensional (3-D) motion data of 15 women during the end of the third trimester, and again one year after delivery, to compare kinematic and kinetic parameters between the two conditions. They reported an increase in the following variables: hip moment of force (M_f), power in the frontal and sagittal planes, maximum ankle plantar flexion M_f, and maximum ankle plantar flexion power absorption. An increase was also observed in the use of the abductor and extensor muscles of the thigh and in the use of the ankle plantar flexor muscles. It was also found that stride width increased.

The kinetic changes during pregnancy may lead to compensation in order to maintain a normal gait pattern. This increased use of hip muscles may contribute to the pain in lower back, pelvic and hip. Stride width increase results in a larger base of support during walking, possibly to improve locomotor stability (Bird, Menz & Hyde, 1999; Foti et al., 2000).

When comparing the effect of externally distributed load carriage with the influence of excessive body mass, a greater hip range of motion (ROM) was found in the former, which suggests that the external load carriage requires greater energy expenditure than unloaded walking (Smith, Roan & Lee, 2010).

3.3 Objectives

The purpose of this study was to analyze the effect of the increased mass in the trunk associated to pregnancy on the lower limb and pelvis, during walking task, on temporal-distance parameters, joint ROM and M_f, by comparing a second trimester pregnant women group to a non-pregnant women group. This last group had to carry a 5 kg additional load located in the abdomen and breasts during walking. This condition helps to clarify which gait adaptations are more related with the increased trunk mass, or which can be more associated to physiological and hormonal changes. This is an alternative to the studies that use longitudinal approach to better characterize the gait changes along the pregnancy period (Foti et al., 2000; Hagan & Wong, 2010). This study, also intended to characterize the differences between gait in young non-pregnant and pregnant women may have an important role because the first condition can be considered similar to the pre-partum period and useful to better understand the process of recovery of normal gait.

3.4 Materials and Methods

3.4.1 Subjects

After being fully informed about the aims and procedures all the participants gave informed consent to participate voluntarily in the study.

The inclusion criteria of participants were: to had no history of lower limb complaints of either orthopedic or neurological conditions, to be aged between 18 and 35 years old, to be in second trimester of an healthy pregnancy, young women who have never been pregnant, to have body mass index between 18 and 26 kg/m^2 (in pregnant women, BMI was considered at the beginning of the pregnancy). Thus, the sample consisted of two groups:

1) Eighteen pregnant women, twelve primiparas and six multiparas, with 27.3±3 weeks of gestational age (second trimester), mean age of 32.6±2.7 years, body mass of 68.2±7.3 kg, height of 1.60±0.1 m and BMI of 26.3±2.6 kg/m^2.
2) Eighteen non-pregnant women with mean age of 20.4±1.5 years, body mass of 58.9±8.4 kg, height 1.60±0.1 m and BMI of 21.9±2.7 kg/m^2.

To understand the effect of the increased mass in the trunk associated to the second trimester of pregnancy, during gait, an extra load was added in the abdomen and breasts of the non-pregnant women, providing a representation of this condition and taking into account only this anthropometric characteristic. A strong large strap and adjustable to the abdominal area was constructed in order to load sandbags with 0.5, 1 and 2 kg weights. The sand allowed adjusting the volume of the extra load to the morphological characteristics of each subject, being tight at the waist with Velcro.

The non-pregnant group (NPG) performed two trials of gait, one of them carrying 5 kg extra load. This value was calculated based on Institute of Medicine recommendations for weight gain during pregnancy (IOM, 2009), which was 0.42 kg/week; we assumed that the load distribution was 34.3% located in the lower trunk (Hytten & Chamberlain, 1991). resulting in 4 kg and 0.5 kg in each breast, which value was based on (Jensen, Doucet & Treitz, 1996) and in the mass distribution for the upper trunk (Hytten &

Chamberlain, 1991). In this condition the group was called load carrying (LCG), with average values of 64.5 kg of weight, 24 kg/m^2 of BMI and 92.6 cm of abdominal girth.

A pilot study was drawn in, where the trunk was included on the biomechanical model. The sub sample consisted of two groups:

1) Three pregnant women two primiparas and one multiparas, with mean gestational age of 30.8±5.1 weeks (second trimester), mean age of 34.7±0.6 years, body mass of 66.5±12 kg of, height of 1.7 m and BMI of 23.4±3.0 kg/m^2.
2) Three non-pregnant women with mean age of 27.3±4.5 years, body mass of 58.8±1.5 kg, height of 1.60 m of and BMI of 22.1±0.4 kg/m^2.

The NPG also performed the trials using an extra load was added in the abdomen and breasts, to recreate the morphological aspect of the pregnant women.

A questionnaire was also applied to characterize the physical activity in what is concerned to volume and type of physical activity level. Regarding the pregnant group (PG), 55.5% performed some physical activity, 3 times per week, with 54 minutes per session. Non-pregnant women were all physically active, exercising 3 times per week during 60 minutes per session.

3.4.2 Data Collection and Processing

Reflective spherical markers were placed on anatomical landmarks of the lower limbs according to the defined marker setup protocol suggested by Capozzo, Capello, Della Croce and Pensalfini (1997).

Motion capture was performed with an optoelectronic system of twelve cameras Qualisys (Oqus-300) operating at a frame rate of 200 Hz, synchronized with two force platforms (Kistler AG, Winterthur, Switzerland) and one AMTI (Advanced Mechanical Technology, Inc, Watertown, MA), to collect ground reaction force (GRF) data. The participants performed three one-minute trials of barefoot walking at a self-selected speed, with a break of thirty seconds between each trial, making a total of about 20 cycles. The subjects were not informed about the platforms location.

The NPG performed unloaded and loaded barefoot walking. For load adaptation, subjects performed a twelve minutes predefined route with walking and climbing/descending stairs, before data collection.

Gait cycles were selected from registered events based on force plate data. Each cycle started at the right heel strike (RHS) and finished at the next RHS.

To reduce noise, the motion data were filtered, using a low pass Butterworth filter, with a cutoff frequency of 15 Hz (Winter, 1991).

The segments were defined using a proximal and a distal end point, measured by the middle distance between lateral and medial markers. Tracking markers were also used allowing the segment follow their coordinates to replicate the performed motion. To build the pelvis segment the Charnwood Dynamics model was used (Charnwood Dynamics Ltd, Leicestershire, UK), which is defined by the anatomical locations of the Anterior Superior Iliac Spine (ASIS) and the Posterior Superior Iliac Spine (PSIS). Thus, the 3-D lower body model included seven body segments: feet, shanks, thighs and pelvis. In the pilot study, was added to the biomechanical model the trunk segment, apart from the other segments above mentioned. Trunk was defined as a unique rigid body delimited by acromiums and ASIS bony landmarks.

In order to reduce the effect of soft tissue artifact, a global optimization on the data processing algorithm was performed (Lu & O'Connor, 1999). It was assume an healthy joint kinematics of all the participants, not taking into account the effect of excessive joint movements. The model also assumed a universal joint to the ankle, a revolute joint to the knee and a spherical joint to the hip.

The inverse dynamics method was used to estimate net joint M_f. To calculate ankle, knee and hip joint angles and M_f, the 3-D biomechanical models were developed with the software Visual 3-D C-Motion, Inc. The weights and locations of the centers of mass for each body segments considered were calculated using the regression equations of Dempster and inertia moments using inertial properties based on their shape (Dempster, 1995). The foot segment was defined by the first and fifth metatarsals, lateral and medial tibia malleolus, to better characterize the frontal movement of the foot. Thus, in this case the zero ankle angle (neutral position) is approximately 70 degrees, but not changing the ankle ROM.

The results were based on five representative cycles per subject, selected based on the stability of gait. The average joint angles and moment curves were obtained for each participant as well as the mean curve for the three groups. Both angular displacement and M_f data were normalized to time cycle, and M_f was normalized to body mass.

3.4.3 Statistical Analysis

For descriptive statistics, continuous data are presented as mean and standard deviations. After verifying the normality distribution of the different parameters, through Shapiro-Wilk test, the comparison between NPG and pregnant group (PG) were performed using the Student t-test. The Mann-Whitney U-test was used when normal distribution was not verified. The comparisons between the NPG and LCG, were carried out by Paired-Samples t-test and for variables did not present a normal distribution, the Wilcoxon test was performed. Statistical tests were performed using the program PAWS Statistics 19.0. The value of $p<0.05$ was used to denote statistical significance.

3.5 Results

Concerning the temporal distance parameters (table 2), right and left stance phase (SP) time, and double limb support time (DLST), PG had higher values when compared to NPG, but not affecting the walking speed. The right SP represented 60.2% and 59.6% of the gait cycle on PG and NPG, respectively. The left step time increased in PG. Stride width was wider in PG.

Compared with NPG, there were no significant differences in PG, regarding ankle joint ROM and maximum dorsi and plantarflexion angles. However, peak eversion angle was 2 degrees higher in pregnant women as the maximum inversion, which also increased in almost 4 degrees.

Consequently, the longitudinal foot rotation ROM increased 6 degrees in PG (table 3). The maximum knee extension angle, occurred at the end of the gait cycle, was lower in PG when compared with NPG and the respective ROM was significantly higher (table 4).

Table 2 - Comparison of temporal distance parameters (mean ± standard deviation) between non-pregnancy and pregnancy groups (NPG_PG), non-pregnancy and load carrying conditions groups (NPG_LCG) and between pregnant and load carrying conditions groups (PG_LCG).

Variable	NPG	PG	LCG	p value
Speed (m/s)	1.24±0.13	1.16±0.12	1.19±0.16	NPG_PG) 0.054 NPG_LCG) <0.001[b] PG_LCG) 0.516
Stride Width (m)	0.08±0.02	0.10±0.02	0.08±0.02	NPG_PG) 0.025[a] NPG_LCG) 0.580 PG_LCG) 0.040[a]
Left Step Length (m)	0.64±0.06	0.62±0.05	0.63±0.06	NPG_PG) 0.315 NPG_LCG) 0.001[a] PG_LCG) 0.976
Right Step Length (m)	0.65±0.06	0.62±0.05	0.63±0.07	NPG_PG) 0.133 NPG_LCG) <0.001[b] PG_LCG) 0.726
Cycle Time (s)	1.04±0.05	1.07±0.06	1.06±0.06	NPG_PG) 0.057 NPG_LCG) 0.094 PG_LCG) 0.334
Left Step Time (s)	0.52±0.02	0.54±0.03	0.53±0.03	NPG_PG) 0.036[a] NPG_LCG) 0.023[a] PG_LCG) 0.286
Right Step Time (s)	0.52±0.03	0.54±0.03	0.53±0.03	NPG_PG) 0.134 NPG_LCG) 0.309 PG_LCG) 0.397
Left Stance Time (s)	0.62±0.04	0.65±0.04	0.64±0.05	NPG_PG) 0.030[a] NPG_LCG) 0.006[a] PG_LCG) 0.279
Right Stance Time (s)	0.62±0.03	0.65±0.04	0.63±0.05	NPG_PG) 0.005[a] NPG_LCG) 0.006[a] PG_LCG) 0.167
Double Limb Support Time (s)	0.19±0.03	0.22±0.03	0.21±0.03	NPG_PG) 0.002[a] NPG_LCG) <0.001[b] PG_LCG) 0.673

Table 3 - Comparison of ankle, knee and hip range of motion - ROM (mean ± standard deviation) between non-pregnancy and pregnancy groups (NPG_PG), non-pregnancy and load carrying conditions groups (NPG_LCG) and between pregnant and load carrying conditions groups (PG_LCG).

Maximum angle and ROM (degrees)	NPG	PG	LCG	p value
ankle dorsiflexion SP **ankle dorsiflexion SW***	85.25±9.70 77.85±8.33	86.51±3.37 79.50±2.16	88.14±3.72 79.66±2.97	NPG_PG) 0.874 NPG_LCG) 0.619 PG_LCG) 0.178 NPG_PG*) 0.569 NPG_LCG*) 0.906 PG_LCG*) 0.856
ankle plantarflexion SP **ankle plantarflexion SW***	58.92±12.70 53.18±13.77	63.53±4.67 55.53±5.14	64.82±3.83 56.93±4.68	NPG_PG) 0.094 NPG_LCG) 0.005 PG_LCG) 0.368 NPG_PG*) 0.899 NPG_LCG*) 0.619 PG_LCG*) 0.398

plantarflexion/dorsiflexion ankle ROM	32.07±5.45	30.98±4.47	31.21±4.92	NPG_PG) 0.518 NPG_LCG) 0.424 PG_LCG) 0.886
ankle eversion	6.54±10.69	4.41±4.50	7.93±3.78	NPG_PG)0.011[a] NPG_LCG) 0.085 PG_LCG) 0.075
ankle inversion	19.84±11.44	23.48±7.99	21.58±6.11	NPG_PG) 0.206 NPG_LCG) 0.396 PG_LCG) 0.428
eversion/invesion ankle ROM	13.30±4.19	19.07±5.11	13.64±3.83	NPG_PG) 0.001[a] NPG_LCG) 0.717 PG_LCG) 0.001[a]
knee flexion SP **knee flexion SW***	23.92±4.61 63.98±4.70	20.37±6.09 63.48±4.83	26.83±6.87 65.64±4.83	NPG_PG) 0.057 NPG_PG) 0.008[a] PG_LCG) 0.195 NPG_PG*) 0.751 NPG_LCG*) 0.082 PGG_LCG*) 0.182
knee extension SP **knee extension SW***	11.2±4.23 9.04±4.72	6.18±4.21 3.37±5.32	13.06±4.86 11.19±4.76	NPG_PG) 0.002[a] NPG_LCG) 0.003[a] PG_LCG) <0.001[b] NPG_PG*) 0.001[a] NPG_LCG*) 0.010[a] PG_LCG*) <0.001[b]
flexion/extension knee ROM	54.94±5.11	60.11±5.22	55.45±5.41	NPG_PG) 0.006[a] NPG_LCG) 0.560 PG_LCG) 0.004[a]
hip flexion SP **hip flexion SW***	34.07±3.94 34.16±3.89	39.91±5.22 40.07±5.26	37.93±5.82 37.58±5.79	NPG_PG) 0.001[a] NPG_LCG) <0.001[b] PG_LCG) 0.291 NPG_PG*) 0.001[a] NPG_LCG*) <0.001[b] PG_LCG*) 0.184
hip extension	-8.28±4.25	-3.35±5.27	-5.67±5.54	NPG_PG) 0.004[a] NPG_LCG) 0.001[a] PG_LCG) 0.207
flexion/extension hip ROM	42.81±3.79	43.77±3.17	43.97±4.22	NPG_PG) 0.414 NPG_LCG) 0.031[a] PG_LCG) 0.877
hip adduction	10.61±3.88	9.18±2.53	8.71±3.47	NPG_PG) 0.201 NPG_LCG) 0.009[a] PG_LCG) 0.642
hip abduction	-10.11±3.28	-9.39±4.87	-8.82±3.81	NPG_PG) 0.609 NPG_LCG) 0.051 PG_LCG) 0.695
adduction/abduction hip ROM	20.72±5.55	18.58±4.31	17.52±4.62	NPG_PG) 0.205 NPG_LCG) 0.003[a] PG_LCG) 0.484
hip internal rotation	2.71±5.14	-0.24±4.71	1.44±6.48	NPG_PG) 0.082 NPG_LCG) 0.185 PG_LCG) 0.380
hip external rotation	-17.79±6.51	-20.58±4.72	-14.54±7.78	NPG_LCG) 0.149 NPG_LCG) 0.003[a] PG_LCG) 0.008[a]
hip internal/external rotation ROM	20.49±5.33	20.34±5.02	15.98±5.12	NPG_PG) 0.931 NPG_LCG) <0.001[b] PG_LCG) 0.014[a]
anterior pelvic tilt	10.23±2.73	16.61±4.49	11.32±4.07	NPG_PG) <0.001[b] NPG_LCG) 0.080 PG_LCG) 0.001[a]

posterior pelvic tilt	6.67±2.56	13.01±4.47	7.46±3.88	NPG_PG) <0.001[b] NPG_LCG) 0.169 PG_LCG) <0.001[b]
pelvic tilt ROM	3.56±1.24	3.60±1.03	3.86±0.92	NPG_PG) 0.923 NPG_LCG) 0.222 PG_LCG) 0.122
pelvic left obliquity	8.25±4.01	6.83±2.51	5.16±3.43	NPG_PG) 0.211 NPG_LCG) <0.001[b] PG_LCG) 0.105
pelvic right obliquity	-8.64±2.45	-5.78±3.48	-7.63±2.61	NPG_PG) 0.007[a] NPG_LCG) 0.018[a] PG_LCG) 0.079
pelvic obliquity ROM	16.89±5.20	12.61±3.62	12.79±3.97	NPG_PG) 0.007[b] NPG_LCG) 0.003[a] PG_LCG) 0.885
pelvic transversal left rotation	8.44±2.72	6.07±3.05	6.20±2.65	NPG_PG) 0.019[a] NPG_LCG) 0.001[a] PG_LCG) 0.893
pelvic transversal right rotation	-7.25±3.06	-7.63±3.43	-5.11±3.23	NPG_PG) 0.731 NPG_LCG) 0.005[a] PG_LCG) 0.030[a]
pelvic rotation ROM	15.69±4.01	13.70±4.95	11.31±3.07	NPG_PG) 0.193 NPG_LCG) <0.001[b] PG_LCG) 0.091

SP – stance phase; SW – swing phase; [a] p values were considered statistical significant; [b] p values <0.001 were considered highly statistical significant.

Table 4 - Comparison of ankle, knee and hip peaks of moments of force - M_f (mean ± standard deviation) between non pregnancy and pregnancy groups (NPG_PG), non-pregnancy and load carrying conditions groups (NPG_LCG) and between pregnant and load carrying conditions groups (PG_LCG).

Normalized peak M_f (N.m/kg)	NPG	PG	LCG	P value
ankle dorsiflexion	1.07±0.21	1.50±0.26	1.14±0.22	NPG_PG) <0.001[b] NPG_LCG) 0.087 PG_LCG) <0.001[b]
ankle plantarflexion	-0.06±0.37	-0.06±0.41	-0.07±0.04	NPG_PG) 0.622 NPG_LCG) 0.711 PG_LCG) 0.376
ankle eversion	-0.02±0.03	-0.01±0.02	-0.01±0.02	NPG_PG) 0.062 NPG_LCG) 0.327 PG_LCG) 0.195
ankle inversion	0.13±0.05	0.19±0.07	0.13±0.08	NPG_PG) 0.002[a] NPG_LCG) 0.659 PG_LCG) 0.018[a]
knee flexion	0.65±0.17	0.66±0.28	0.68±0.30	NPG_PG) 0.752 NPG_LCG) 0.811 PG_LCG) 0.899
knee extension	-0.14±0.07	-0.19±0.11	-0.15±0.07	NPG_PG) 0.137 NPG_LCG) 0.133 PG_LCG) 0.343
hip flexion	0.41±0.13	0.53±0.11	0.49±0.16	NPG_PG) 0.008[a] NPG_LCG) <0.001[b] PG_LCG) 0.490
hip extension	-0.89±0.14	-0.93±0.24	-0.81±0.14	NPG_PG) 0.950 NPG_LCG) 0.004[a] PG_LCG) 0.137

hip adduction	0.67±0.19	0.93±0.16	0.77±0.21	NPG_PG) <0.001[b] NPG_LCG) 0.006[a] PG_LCG) 0.014[a]
hip abduction	0.02±0.56	0.05±0.04	0.14±0.07	NPG_PG) <0.001[b] NPG_LCG) 0.010[a] PG_LCG) 0.095
hip external rotation	0.13±0.05	0.15±0.06	0.18±0.08	NPG_PG) 0.849 NPG_LCG) <0.001[b] PG_LCG) 0.164
hip internal rotation	-0.06±0.02	-0.10±0.05	-0.05±0.02	NPG_PG) 0.023[a] NPG_LCG) 0.149 PG_LCG) 0.002[a]

[a] p values were considered statistical significant; [b] p values <0.001 were considered highly statistical significant.

Table 5 - Comparison of trunk range of motion - ROM (mean ± standard deviation) between non-pregnancy and pregnancy groups (NPG_PG), non-pregnancy and load carrying conditions groups (NPG_LCG) and between pregnant and load carrying conditions groups (PG_LCG).

Trunk ROM (degrees)	NPG	PG	LCG	p value
Sagittal	1.98±0.66	1.78±0.22	1.60±0.47	NPG_PG) 0.513 NPG_LCG) 0.285 PG_LCG) 0.513
Frontal	3.13±1.30	2.11±0.91	2.23±1.02	NPG_PG) 0.275 NPG_LCG) 0.285 PG_LCG) 0.827
Transverse	11.31±5.90	7.06±2.35	7.81±4.10	NPG_PG) 0.275 NPG_LCG) 0.109 PG_LCG) 0.827

[a] p values were considered statistical significant; [b] p values <0.001 were considered highly statistical significant.

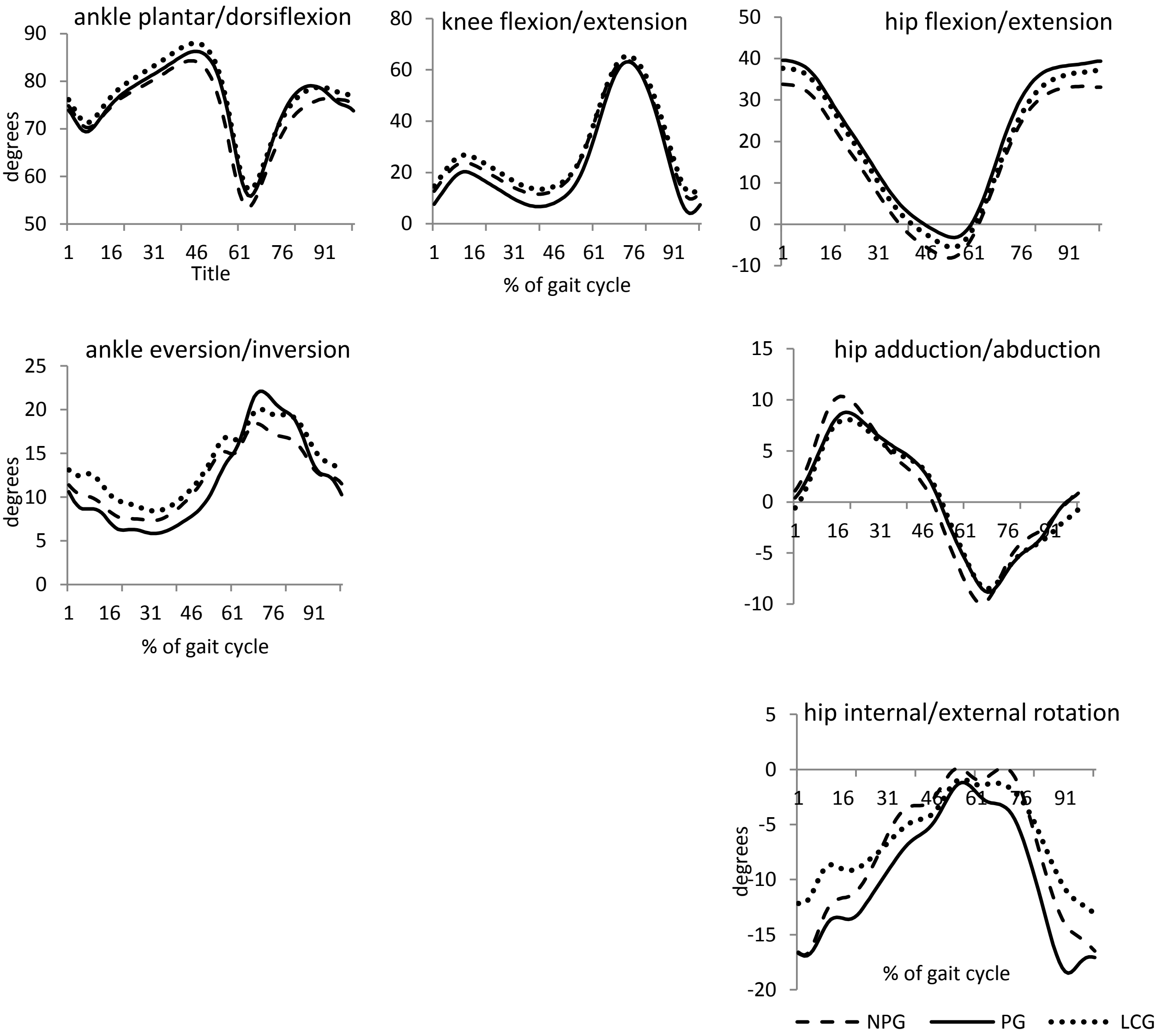

Figure 12 - Ankle, knee and hip joints range of motion (ROM) in sagittal (first row), frontal (second row) and transversal planes (third row) for the non-pregnancy (NPG), pregnancy (PG) and load carrying (LCG) groups.

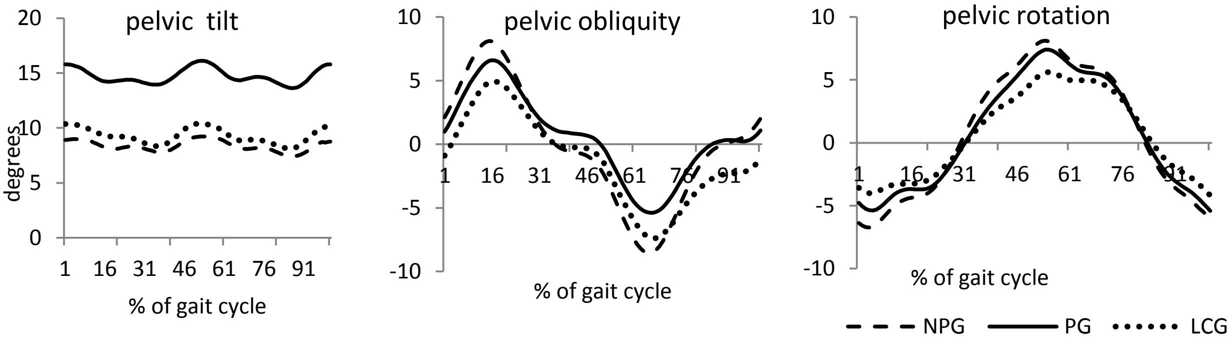

Figure 13 - Pelvis range of motion (ROM) in sagittal (first row), frontal (second row) and transversal planes (third row) for the non-pregnancy (NPG), pregnancy (PG) and load carrying (LCG) groups.

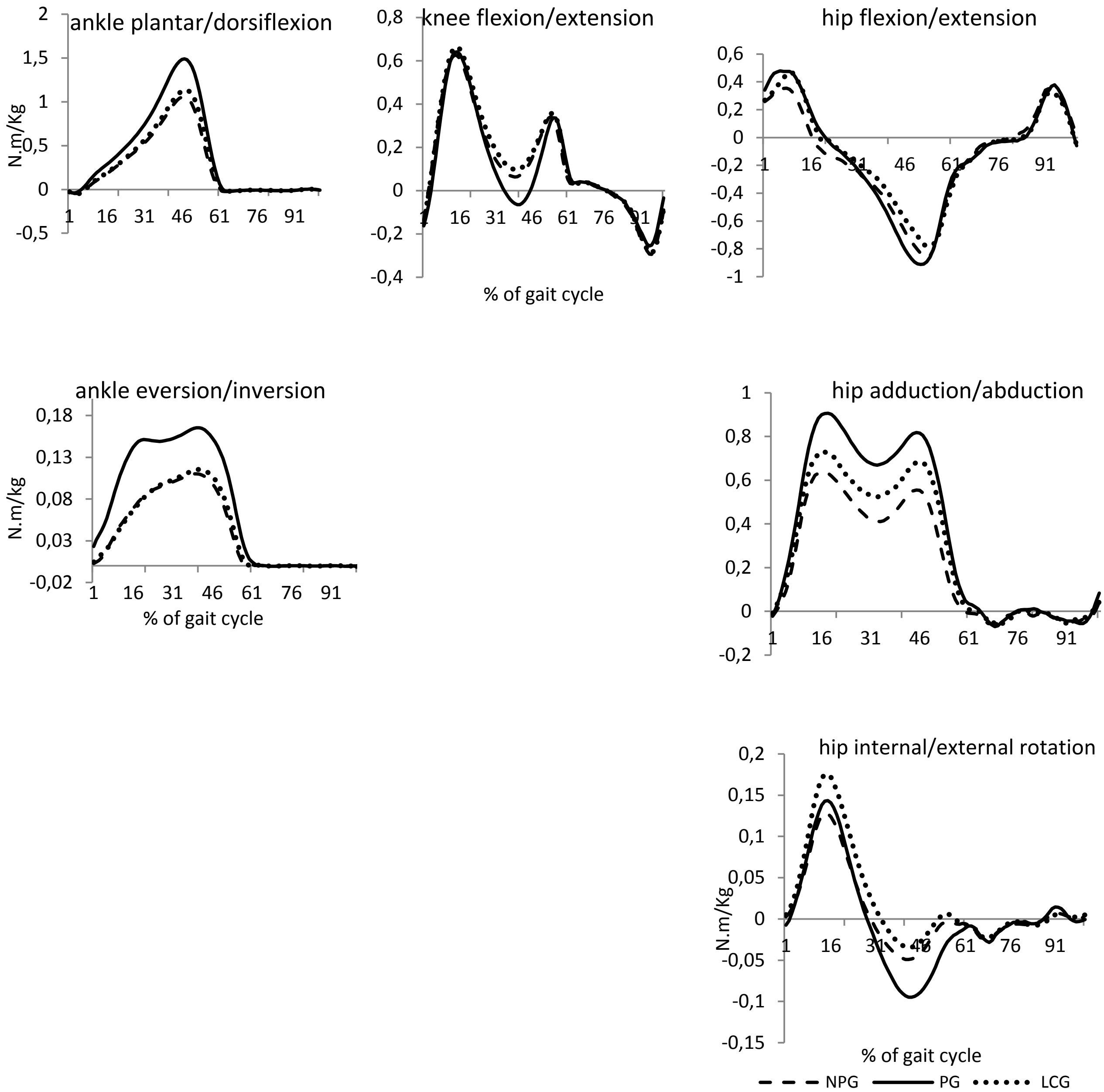

Figure 14 – Ankle, knee and hip joints moments of force (M_f) in sagittal (first row), frontal (second row) and transversal planes (third row) for the non-pregnancy (NPG), pregnancy (PG) and load carrying (LCG) groups.

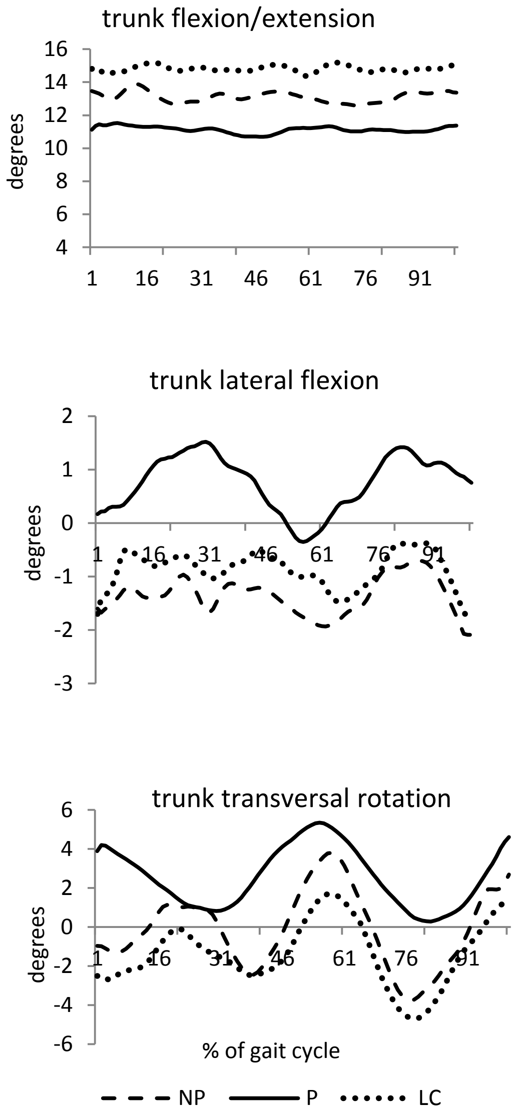

Figure 15 – Ankle, knee and hip joints moments of force (M_f) in sagittal (first row), frontal (second row) and transversal planes (third row) for the non-pregnancy (NPG), pregnancy (PG) and load carrying (LCG) groups.

Concerning the kinetic variables, in the ankle joint in PG showed higher peak dorsiflexion moment at the end of the SP. Also inversion peak moment was 0.06 N.m/kg higher in PG. In the hip joint, PG had higher hip peak flexion moment. Regarding the adduction moment, it was also higher in PG as the abduction peak moment which had a significant higher value mid stance (33%) of the gait cycle (figure 12). Also the internal peak moment was significantly higher in the pregnancy condition.

Few significant differences were also found between LCG and NPG. In LCG, double limb support time, right and left stance times, were significantly higher and it was observed slower walking speed. While carrying the extra load right and left step lengths were shorter. Left step time was higher in LCG; SP was 59.4% of total gait cycle time (table 2).

At the end of the SP, LCG increased the maximum plantarflexion ankle angle in 6 degrees, when compared with NPG. The maximum knee extension was higher in LCG. At the hip, the maximum flexion angle was higher in PG and the extension peak value was lower. During the SP, the maximum knee flexion and extension angles were 3 and 2 degrees higher, respectively, in LCG. The hip maximum flexion angle, increased in LCG, in both instances, after initial contact and in the end of the swing phase (SW). In what concerns the maximum adduction angle, the value in LCG was lower than in NPG, which reduces adduction/abduction ROM. The hip maximum external rotation was significantly lower in LCG as the internal/external rotation ROM. No significant differences were found in pelvic tilt. However maximum pelvic right and left obliquity angles were lower for LCG reducing pelvic obliquity ROM. Right and left transverse pelvic rotations and ROM were also lower in LCG (table 3).

Flexion and adduction hip maximum M_f were higher for LCG at the same point of the SP. The external rotation peak moment was higher in the earlier SP (16% of gait cycle) as well as the abduction peak moment that had significant higher values around 33% of gait cycle (figure 13).

Comparing PG with LCG, stride remained wider in the first group. Some kinematic differences were observed: in the frontal plane, the ankle eversion/inversion ROM increased 5 degrees in PG. During SP the knee extension in PG reached values close to 6 degrees, while higher values were observed in LCG (table 4). The maximum knee extension angle occurred during the terminal swing, where P group presented a more extended knee. Thus, knee ROM remained higher for PG. No significant differences were found between PG and LCG, in hip sagittal ROM. Maximum hip external rotation was 6 degrees higher in PG, also increasing the respective ROM. Maximum anterior and posterior pelvic tilt values in PG changed for higher values, not affecting the ROM. On transversal plane, the maximum right rotation angle was also higher in PG, and in this case, didn´t affect the corresponding ROM.

Regarding M_f, dorsiflexion and inversion peak moments remained the highest in PG. In the loading response, hip adduction and internal rotation peak M_f were also higher in PG, as the external peak moment that occurred in the end of the SP (table 4).

About the trunk ROM, the results obtained from the pilot study allows to confirm that no significant differences were found between groups.

3.6 Discussion

The purpose of this study was to find gait biomechanical variables that were affected by the increased trunk mass, which is a characteristic of second trimester of pregnancy. Some findings were obtained as the changes in the support time, in the hip and pelvis kinematics and kinetics in different planes. To achieve these findings, comparisons between the three groups tested were done. Ideal conditions would have been to establish comparisons between the same woman when non-pregnant and during pregnancy. Due to circumstances of timing this was not possible, so NPG and PG are two different groups of women. Conclusions from comparing these groups in terms of significant differences in variables have therefore to be made with care. To understand the isolated effect of the added trunk mass, comparisons between NPG and LCG were performed. In these groups the subjects are the same, so significant differences in a variable between them is relevant. If there is a significant change in a variable value when the same subject has a weight applied to her we could confirm that increased weight will be relevant for the change in this variable. In the comparisons between NPG and PG, we use this fact to give strength to the changes related to the real increased trunk mass associated to pregnancy. In this way we establish three groups where conclusions can be drawn. From here on, comparison between NPG and PG is designated by (NPG_PG), NPG and LCG by (NPG_LCG) and PG and LCG by (PG_LCG). From these comparisons, three sets of particular variables were designed.

In set I, there are significant differences in a variable in (NPG_PG) and (NPG_LCG), and no difference in (PG_LCG). From the difference in (NP_LC), as mentioned above, we have clear relation between the extra weight and changes on the variable. Differences in (NPG_LCG) would make us expect differences in (NPG_PG), which is the case. No changes in (PG_LCG) reinforce the fact that weight is influent on the change in the variable.

In the set I we find the variables: left step time, left stance time, right stance time, double limb support time, maximum hip extension, maximum pelvic right obliquity, pelvic obliquity ROM, maximum transversal left rotation and peak hip flexion M_f. The trunk weight gain increases both SP duration and DLST in both LCG and PG (Lai, Leun and Li, 2009). The body’s response to the external hip flexor moment is related to a higher extensor activity to support the anterior additional mass of the trunk and to the forward translation of the center or mass.

The pelvis is linked to the thigh and trunk, and supports the belly weight. So its amplitude on the frontal and transversal planes may be reduced due to the surrounding constraints. The hip joint and pelvis are adjacent to the trunk and can be more affected by the weight gain and its distribution.

In set II, we have differences in (NPG_PG), (NPG_LCG) and (PG_LCG). This is similar to group I, but the fact that the variable is significantly different in (PG_LCG) weakens the connection between the extra load and the change in the variable. This difference could be explained by the fact that weight is distributed differently in LC and P, and in P the subject had several months of adaptation to a gradual weight increase, and hormonal and physiological changes that adapt the body to extra load, whereas in LC the subject is not changed apart from the extra weight, adaptation to it consisting of no more than 12 minutes of physical activity.

In relation to this set, three variables were found: maximum knee extension during the SP, maximum knee extension during the swing phase and peak hip adduction M_f. Pregnant women extend more the knee, during stance and swing phases. Furthermore, hip peak adduction M_f increases, which may be a consequence of a wider stride. The same happens to hip peak adduction M_f in LCG, although not so much as in the PG.

In set III, there are differences in (NPG_PG) and (PG_LCG), and no differences in (NPG_LCG). The fact that a variable is not different in (NPG_LCG) shows some independence of it to weight. Therefore, the differences in (NPG_PG) can only be attributable to other factors, such as hormonal or physiological changes, or the fact that the subjects are different. The difference in (PG_LCG) is consistent with this situation. In conclusion, the variables in group I show a statistically significant dependence on weight, those in III on other factors, those in group II being in between.

Thus, in set III we have the following variables: stride width, maximum ankle eversion, eversion/inversion ankle ROM, flexion/extension knee ROM, maximum anterior pelvic tilt, maximum posterior pelvic tilt, peak ankle dorsiflexion M_f, peak ankle inversion M_f and peak hip external rotation M_f. Stride width is higher in pregnancy, thus creating a larger base of support and thereby providing more stability (Foti et al., 2000; Bird et al., 1999). Ankle maximum dorsiflexion and inversion angles decrease during the second trimester of pregnancy (Hagan & Wong, 2010). Pregnant women do more eversion and increase eversion/inversion ankle ROM, which might be related to joint laxity associated to the pregnancy (Bird et al., 1999; Ponnapula & Boberg, 2010). During the loading response, in the sagittal plane, maximum plantarflexion M_f reduces to zero during the earlier SP. From this moment on, dorsiflexion torque increases, reaching higher peak values in P, and the effect of body weight is shown in the acceleration of the downward fall (Lu & O'Connor, 1999). The decrease in the maximum dorsiflexion ankle angle and increase in peak ankle dorsiflexion M_f, both on PG, suggest that plantarflexor muscles are more active during the second trimester of pregnancy. Also the peak inversion M_f increases at the subtalar joint, which means that the everter muscles of this joint (anteriorly: the extensor digitorum longus and peroneus tertius; posteriorly: gastrocnemius, peroneus longus and peroneus brevis) are more active in order to control the ankle (Perry, 1992), once that pregnant women present more eversion/inversion mobility. Immediately before heel strike, pregnant women extend more the knee and the respective ROM, while hip flexion increases (Block et al., 1985; Bird et al., 1999), which in turn may be a consequence of an increased anterior pelvic tilt (Foti et al., 2000; Smith et al., 2010). Other studies indicate that increased pelvic tilt is typically related with increased lumbar lordosis (Foti et al., 2000). Increased laxity during pregnancy can be responsible for these kinematical changes. The same happens in the terminal swing where the knee is again more extended and the hip more flexed in PG.

This study led us to conclude that the anterior increased mass of the trunk associated with second trimester of pregnancy may influence some gait biomechanical variables gait as the left step time, left and right stance times, double limb support time, maximum hip extension, maximum pelvic right obliquity, pelvic obliquity ROM, maximum transversal left rotation and peak hip flexion M_f. However the additional load should not be understood has a simulation of the pregnancy condition, once that we found several biomechanical variables, which changes can have a strong relation with the specific physiological and hormonal adaptation of pregnancy.

The main goal of this research was focused on the lower limbs and pelvis. Although the trunk and pelvis movements have a secondary function on the lower limbs (Perry, 1992), it was included a pilot study to better understand the trunk movement. The results were not conclusive possibly due to the small number of studied subjects.

One factor to be taken into consideration for further analysis is the adaptation to the 5 kg extra load, which can be modified in terms of distribution, duration, or the activity route use for the accommodation to the increased load.

The use of extra load is not a simulation of pregnancy condition, and this issue needs more attention, particularly concerning the distribution and the adaptation to it. The relation between anterior trunk extra weight and biomechanical gait modifications needs further research since weight gain during pregnancy provides higher biomechanical joint loads during walking which is a common way of physical exercise during pregnancy and a daily physical activity during almost forty weeks, that can cause injuries by overuse of certain specific muscles.

3.7 References

Bird, A., Menz, H. & Hyde C. (1999). The effect of pregnancy on footprint parameters. A prospective investigation. Journal of the American Podiatric Medical Association, 89, 405-409.

Block, R.A., Hess, L.A., Timpano, E.V. & Serlo, C.(1985). Physiological changes in the foot in pregnancy. Journal of the American Podiatric Medical Association, 75, 297–299.

Brett, M. & Baxendale, S. (2001). Motherhood and memory: a review. Psychoneuroendocrinology 26, 339–362.

Butte, N., Ellis, K., Wong, W., Hopkinson, J. & Smith, E. (2003). Composition of gestational weight gain impacts maternal fat retention and infant birth weight. American Journal of Obstetrics and Gynecology 189(5), 1423-1432.

Calganeri, M., Bird, H.A. & Wright, V. (1982). Changes in joint laxity occurring during pregnancy. Annals of the Rheumatic Diseases, 41, 126–128.

Capozzo, A., Cappello, A., Della Croce, U. & Pensalfini, F. (1997). Surface-Marker Cluster Design for 3-D Bone Movement Reconstruction. IEEE Transactions on Biomedical Engineering, 44(12), 1165-1174.

Dempster, W.T. (1995). Space requirements of the Seated Operator: Geometrical, Kinematic, and Mechanical Aspects of the Body with Special Reference to the Limbs. WADC Technical Report 55-149. Wright-Patterson Air Force Base, OH.

Dumas, G.A., Reid, J.G., Wolfe, L.A., Griffin, M.P. & McGrath, M.J. (1995). Exercise, posture, and back pain during pregnancy. Clinical Biomechanics, 10, 98–103.

Dunning, K., LeMasters, G., Levin, L., Bhattacharya, A., Alterman, T. & Lordo, K. (2003). Falls in workers during pregnancy: risk factors, job hazards, and high risk occupations. American Journal of Industrial Medicine, 44, 664–672.

Foti, T., Davids, J. & Bagley, A. (2000). A Biomechanical Analysis of Gait During Pregnancy. The Journal of Bone and Joint Suergery, Incorporated, 82-A(5), 625-632.

Gilleard, W. (2013). Trunk motion and gait characteristics of pregnant women when walking: report of a longitudinal study with a control group. BMC Pregnancy and Childbirth, 13 (71).

Hagan, L. & Wong, C. (2010). Gait in Pregnant Women: spinal and lower extremity changes from pre- to postpartum. Journal of Women's Health Physical Therapy, 34(2), 46-56.

Hytten, F. & Chamberlain, G. (1991) Clinical Physiology in Obstetrics. Oxford: Blackwell Scientific Publications.

Jensen, R.K., Doucet, S. & Treitz T. (1996). Changes in segment mass and mass distribution during pregnancy. Journal of Biomechanics, 29, 251–256.

IOM, NRC. (2009). Weight gain during pregnancy. Reexamining the guidelines. Washington, D.C.: The National Academies Press.

Lai, P., Leung., A. , Li, A. and Zhang, M. (2009). Three-dimensional gait analysis of obese adults. Clinical Biomechanics, 23, S2-S6.

Lu, T.W. & O'Connor, J.J.(1999). Bone position estimation from skin marker coordinates using global optimization with joint constraints. Journal of Biomechanics, 32, 129-134.

McNitt-Gray, J.L.(1991). Biomechanics related to exercise in pregnancy, Exercise in pregnancy 2nd edition Williams & Wilkins, Baltimore, 133-140.

Paisley, J. & Mellion ,M. (1988) Exercise during pregnancy. American Family Physician, 38(5), 143 – 150.

Perry, J. (1992) Gait Analysis. Normal and Pathological Function. Thorofare, New Jersey: SLACK Inc.

Pitkin, R. (1976). Nutritional support in obstetrics and gynecology. Clinical Obstetrics and Gynecology, 19(3), 489 – 513.

Ponnapula, P. & Boberg, J. (2010). Lower Extremity Changes Experienced During Pregnancy. The Journal of Foot & Ankle Surgery, 49, 452–458.

Smith, B., Roan, M. & Lee, M. (2010). The effect of evenly distributed load carrying on lower gait dynamics for normal weight and overweight subjects, 32, 176-180.

Snijders, C.J., Seroo, J.M., Snijder, J.G. & Hoedt, H.T. (1976). Change in form of the spine as a consequency of pregnancy. In: Proceedings for the 11th International Conference on Medical and Biological Engineering. 670-671. Ottawa, ONT, CAN.

Winter, D.A. (1991). The biomechanics and motor control of human gait: normal, elderly and pathological. Waterloo, Ontario: University of Waterloo Press.

Wu, W., Meijer, O., Lamoth, C., Uegaki, K., van Dieën, J., Wuisman, P., de Vries, J. & Beek, P.(2004). Gait coordination in pregnancy: transverse pelvic and thoracic rotations and their relative phase. Clinical Biomechanics, 19, 480-488.

Chapter 4

4 Global Optimization Method Applied to the Kinematics of Gait in Pregnant and in Non-Pregnant Women [3]

[3] Accepted for publication as: Aguiar L, Andrade C, Branco M, Santos-Rocha R, Vieira F, Veloso AP (2014). Global optimization method applied to the kinematics of gait in pregnant and in non-pregnant women. Journal of Mechanics in Medicine and Biology.

4.1 Abstract

Morphological changes are associated to the pregnancy, such as weight gain and increased volume of the trunk. The soft tissue artifact can also increase with these characteristics. The main objective of this study was to understand the effect of using three different constraining sets in the lower limb joints, in the amount of soft tissue artifact of pregnant and non-pregnant women. The ankle, knee and hip joints were modeled respectively with the following characteristics: 1) universal-revolute-spherical, 2) spherical-revolute-spherical and 3) spherical-spherical-spherical. The six degrees of freedom model was used as the basis for comparison, considering that is the approach with highest error associated to the soft tissue artifact. The results show that in non-pregnant women, the use of three spherical joints is the method that revealed more differences with the six degree of freedom model. The ankle was the most affected segment when modifying the constraints, particularly ankle plantar/dorsiflexion and adduction/abduction movements in spherical-revolute-spherical and three spherical joint models. In pregnant women, the universal-revolute-spherical model seems to affect more the kinematic variables when compared with the six degrees of freedom model. Assuming that the kinematic error associated with pregnant women is increased due to the soft tissue artifact, the universal-revolute-spherical model may be affecting more the angular kinematics of the knee joint. Regarding the morphological changes during pregnancy, it is important to understand how the manipulation of the joints' degrees of freedom influences the kinematic variables, even of the adjacent joints, considering that the developed biomechanical is an assembly of rigid bodies acting as a kinematic chain.

Keywords: Gait, pregnancy, kinematics, global optimization.

4.2 Introduction

Kinematic analysis is widely used in the study of human gait, particularly in situations of disabilities, rehabilitation processes or exercise effects, so its improvement is extremely important. Several sources of error affect joint kinematics, such as instrumental inaccuracies, anatomical landmark mislocation and soft tissue artifact (STA). Inertial effects, skin deformation and sliding, gravity and muscle contraction contribute independently for STA (Capello, Stagni, Fantozzi & Leardini, 2005). Leardini, Chiari, Della Croce and Capozzo (2005) and Ceratti et al. (2006) found that STA have a frequency component in the same region as the bone motion, and is subject and task dependent. Holden et al. (1997) verified that the amount of STA depends on the physical characteristics of the individuals and Fuller, Liu, Murphy and Mann (1997) demonstrated that it could also depend on the nature of the movement task.

During pregnancy, the growth of the fetus and of the surrounding areas causes anthropometric and inertial changes in the women trunk and limbs, mainly in the lower trunk mass and in the principal moments of inertia (Jensen, Doucet & Treitz, 1996). Pregnancy causes morphological adaptations, like weight gain and skinfolds increase, which are well established from week 20th (Pitkin, 1976; Taggart et al., 1967; Butte et al., 2003). More changes can include increased ligamentous laxity, decreased neuromuscular control and coordination and swelling of the arms and legs (Block et al., 1985; Brett & Baxendale, 2001; Calganeri et al., 1982; deGroot et al., 2003; Dumas et al., 1995; Dunning et al., 2003; Hainline, 1994; McNitt-Gray, 1991; Snijders et al., 1976). It was also demonstrated that more than 50% of women reported swelling of the foot, ankle and leg, unsteady gait and increased foot width (Ponnapula & Boberg, 2010). These changes may also contribute to the increase of STA.

It is common to model the musculoskeletal system as a chain of multilinked rigid bodies. For a three dimension (3-D) representation, at least three markers are needed to build a segment embedded reference frame which is associated to a pose. Due to the skin movement, the markers move relatively to the underlying bone and the array of markers shape also changes. To minimize this STA effect, a global optimization method (GOM) can be applied, where joint constraints are added to the model in order to minimize the sensor noise and STA. An optimal pose is estimated of a multi-link model that best matches the motion capture data in terms of global criterion (Lu & O'Connor, 1999). This method computes realistic motion representing healthy joint kinematics and not allowing the effect of excessive joint movements, which can happen in pregnant women.

Foti et al. (2000) used a 3-D biomechanical model representation of women during the later stage of the third trimester of pregnancy, and again one year after delivery. Motion data were collected with six video cameras, performing at 60 Hz. It was used spherical markers in specific bony landmarks. So, the 3-D gait analysis of pregnant women has already been studied but the STA issue has never been reported.

Researchers emphasize that STA is subject- and task-dependent (Leardini et al., 2005; Cereatti et al., 2006; Andersen, Benoit, Damsgaard, Ramsy & Rasmussen, 2010), but with respect to pregnant women the STA measurement can be challenging once that invasive methodologies are not appropriated to this population, which is particularly vulnerable to medical imaging exposure causing deleterious effects (Goodman & Amurao, 2012). Based on the anthropometric changes associated with pregnancy this study raises the hypothesis that the STA is increased in pregnant women, when compared with non-pregnant women. To understand this phenomenon and how different joint constraint models affect the kinematics of gait in pregnant women, a GOM was used with three different sets of joint constraints.

The ankle, knee and hip joints were modelled respectively with the following characteristics:

1) Universal-revolute-spherical (URS);
2) Spherical-revolute-spherical (SRS), and;
3) Spherical-spherical-spherical (SSS).

The gait of two groups was analyzed in the following groups:

1) Second trimester pregnant women, and;
2) Non-pregnant women.

4.3 Objectives

The objectives of the study were to compare three different approaches on joint modelling of the ankle, knee and hip, as follows: 1) universal-revolute-spherical (URS); 2) spherical-revolute-spherical (SRS), and; 3) spherical-spherical-spherical (SSS), in regard of the gait of two groups of participants: 1) second trimester pregnant women, and; 2) non pregnant women.

4.4 Materials and Methods

4.4.1 Subjects

The sample was composed of 36 subjects divided in two groups: a pregnant women group (PG) and a non-pregnant women group (NPG). The PG consisted of eighteen women with 32.6±2.8 years of chronological age, 68.2±7.3 kg of weight, 1.60±0.10 m of height, 26.4±2.6 kg/m^2 of body mass index (BMI) and 27.4±3.0 weeks (second trimester) of gestation. The NPG consisted of eighteen women with 20.5±1.5 years of chronological age, 59.1±8.4 kg of weight, 1.60±0.10 m of height and 22.1±2.7 kg/m^2 of BMI. The PG included twelve primiparas and six multiparas.

Pregnant subjects were recruited via direct contact and flyers placed in gym and health centers. NPG subjects were recruited among faculty students. All subjects gave informed written consent to participate voluntarily in the study. The subjects had no history of foot, ankle, knee, musculoskeletal, neuromuscular trauma or disease. None of the subjects had contraindication to physical exercise.

4.4.2 Data Collection and Processing

The study was approved by the ethical committee of the faculty and data were collected at the Laboratory of Biomechanics and Functional Morphology.

Anthropometric data (weight and height) were collected by certified kinanthropometrists by ISAK, to calculate de body segments masses and inertia moments. Twenty three reflective markers and four marker clusters (each one containing three reflective markers), were placed on predefined anatomical points (figure 16): Segments were defined using a proximal and a distal end point, measured by the middle distance between lateral and medial markers. Foot segment was defined by first and fifth metatarsals lateral and medial tibia malleolus. For the shank it was used markers placed in tibia malleolus and in femur condyles. Reflective spherical markers were placed on anatomical landmarks of the lower limbs according to the defined marker setup protocol suggested by Capozzo et al. (1997). Pelvis was built based on CODA model, defined using the anatomical locations of the right and left anterior superior iliac spine and the right and left posterior superior iliac spine. This model allows the estimation of the right and left hip joint center using the regression equation proposed by Bell et al. (1989; 1990).

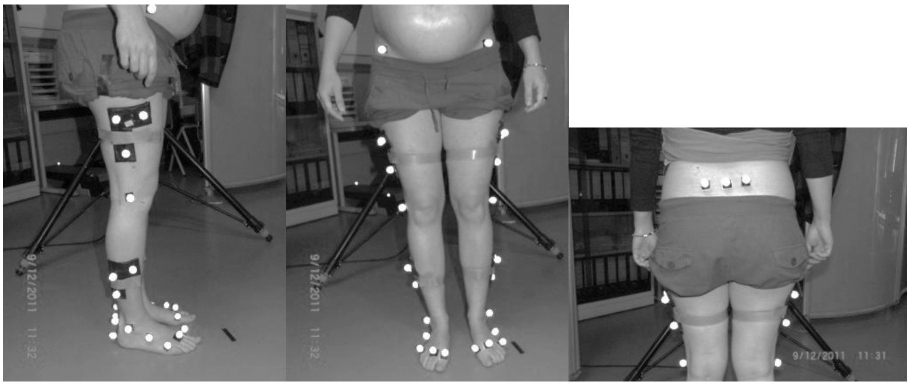

Figure 16 - Marker setup used for motion captures and lower limb and pelvis constructions: five markers for pelvis, five markers for thigh, seven markers for shank an eight markers for foot a) Posterior setup. b) Anterior setup. c) Reconstructed biomechanical model in Visual 3-D.

Motion capture was performed with an optoelectronic system of twelve cameras Qualisys (Oqus-300) operating at a frame rate of 200 Hz, synchronized with two Kistler force platforms (Kistler AG, Winterthur, Switzerland) and one AMTI force platform (Advanced Mechanical Technology, Inc, Watertown, MA), which collected ground reaction force (GRF) data. The participants performed three non-consecutive minutes walking at a comfortable speed, with a time break of thirty seconds between each minute.

The gait cycles were selected based on both feet GRF data and the heel markers pattern recognition. A complete recorded cycle started with the right heel strike (RHS) and ended at the next RHS.

Five cycles per subject were analyzed. The average joint angles were calculated for each participant and the mean curve for each joint in each group is reported. Angular displacement data were normalized to the duration of the cycle. To compare the results between the three models, a repeated measures ANOVA test was used. The level of statistical significance was set at $p \leq 0.05$ and all the tests were performed using the software PAWS Statistics 19.0.

4.4.3 Global Optimization Method

Considering the model segments as rigid bodies and with six degrees of freedom, the position and orientation of the local SCS (segment coordinate system) was described with respect to the laboratory coordinate system (LCS). The location of one point in the

LCS was given by equation 23, where *T* was the rotation matrix from the SCS to the LCS and *O* is the translation between coordinate systems.

$$P = TA + O \qquad \text{(Eq. 23)}$$

Thus, *A* is a point determined from measurement of *P* at this position (equation 24). The matrix T and translation vector *O*, were computed in all instants, provided for at least three noncolinear points and by minimizing the sum of squares error (equation 25), under the orthonormal condition (equation 26), where *m* is equal to the number of targets on the segment. Lagrange multipliers were used, to find the maxima and minima of the function subject to constraints, given by equation 27 (Spoor & Veldpaus, 1980).

$$A = T^{-1}(P - O) \qquad \text{(Eq. 24)}$$

$$\sum_{i=1}^{m}((P_i - TA_i) - O)^2 \qquad \text{(Eq. 25)}$$

$$T^{-1}T = T^{T}T = I \qquad \text{(Eq. 26)}$$

$$g(T) = T^{T}T - I = 0 \qquad \text{(Eq. 27)}$$

All the markers coordinate data were interpolated using a third degree polynomial function to fulfill gap displacements. To reduce the noise the motion data was filtered, using a low pass Butterworth filter, with a cutoff frequency of 15 Hz (Winter, 1991).

Due to the STA, the markers move relatively to the bone, modifying the initial configuration and therefore reproducing noisy motion data. Moreover, the segments are considered linked rigid bodies and joints have 6DOF, making them independent from each other and allowing more movement. The GOM was used (Lu & O'Connor, 1999) to find an optimal position for the set of segments, which constitute the model on each frame. To estimate the general pose for each frame, the differences between the measured coordinates in the static position and the measured coordinates during motion were minimized by the method of least squares. The use of GOM undertakes that all the participants present healthy joint kinematics not allowing the effect of excessive joint movements.

The STA was estimated using the calculation of the segments residuals. The difference between each segments pose with respect to the static trial was calculated using a least squares fit of the tracking markers locations. The amount of STA is given by that difference.

The degrees of freedom of each joint were manipulated and three constraints' sets were chosen. In the first model (URS), it was used a universal joint in the ankle, allowing rotations in sagittal and frontal planes, a revolute joint in the knee only with the sagittal rotation permitted and a spherical joint in the hip allowing the rotations in the three planes of motion. This is a common way of modeling the joints sharing points and vectors among the adjacent segments (Czaplicki, Silva & Ambrósio, 2004; Charlton, Tate, Smyth and Roren, 2004). For the second model (SRS), it was used two spherical joints in the ankle and hip, allowing rotations in the three planes of motion and one revolute joint in the knee for the sagittal rotation. Finally, in the third model (SSS) we used three spherical joints for the ankle, knee and hip, allowing the rotations in the three planes of motion. In the three models, all the translations were restricted.

Given a set of measured marker coordinates *P* on a data frame, the global optimization at the system level was to find a set of generalized coordinates ξ so that the following error function (equation 28) is minimized, where *W* is a positive-definite weighting matrix, *P'* (ξ) is the corresponding set of marker coordinates calculated by the transformation from the segment frame to laboratory frame (equation 29).

$$f(\xi) = [P - P^{'}(\xi)]^T W [P - P^{'}(\xi)] \qquad \text{(Eq. 28)}$$

$$P^{'}(\xi) = T(\xi)P^{*} \qquad \text{(Eq. 29)}$$

W is a ($3m_i$ x $3m_i$) weighting matrix assigned to the each segment to reflect the error distribution among the m_i markers. A different weighting factor was given to each segment, reflecting its average degree of soft tissue artifacts. In this study it was used 1 for all segments weights. For each segment, the equation 30 was solved to find the minimum value, under the orthogonal constrain $R^T R = I$, where x_i and y_i are position vectors of marker *i* at the reference and global positions, respectively. *R* is the rotation matrix, *v* is the translation vector and *m* is the number of markers.

$$f = \sum_{i=1}^{m} (Rx_i + v - y_i)^T (Rx_i + v - y_i) \qquad \text{(Eq. 30)}$$

The minimum value of f is the segmental residual error (e), a measure of the markers distortion which is mainly due to STA. The weighting matrix W_i is defined as equation 31, and it was 1 for all the segments.

$$W_i = \frac{1}{e_i} I \qquad \text{(Eq. 31)}$$

These calculations are suggested for the kinematics of a rigid body between two sequential positions, where the first one works as the reference (Cappello et al., 1996; Challis, 1995; Spoor & Veldpaus, 1980; Veldpaus et al., 1988).

The GOM application modifies the SCS location and consequently the marker coordinates. To add more information about this issue, the new coordinates were calculated and also the differences between these and the 6DOF approach, since this is the reference method most commonly used. The differences of three markers coordinates from each segment, between 6DOF and the three GOM setup of joints (URS, SRS, SSS) (figure 17), were calculated.

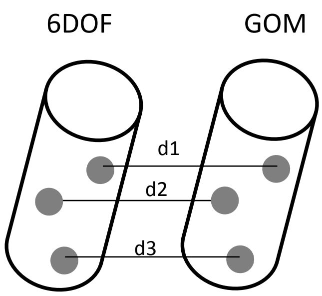

Figure 17 - In the same frame of the same cycle d1, d2 and d3 are the differences between three markers associated to a segment, calculated with 6 degrees of freedom (6DOF) and global optimization model (GOM) approaches.

It was used one cycle for each subject, and three markers for each segment. For each marker the average distance (d_j) between coordinates using the two methods is given by equation 32.

$$d_j = \frac{\sqrt{\sum_{i=0}^{n}(a_x(i)-b_x(i))^2+\sum_{i=0}^{n}\left(a_y(i)-b_y(i)\right)^2+\sum_{i=0}^{n}(a_z(i)-b_z(i))^2}}{n} \qquad \text{(Eq. 32)}$$

where a and b are the 3-D correspondent coordinate markers between 6DOF and GOM. The number of markers is given by j and the frame number by i, where the total number of frames in the cycle is n. For each segment it was calculated the norm given by equation 33 that represents the general difference between coordinate measurements. The segments coordinates distance were calculated with Matlab (Mathworks, USA).

$$\|d_{segment}\| = \sqrt{(d_1^2 + d_2^2 + d_3^2)} \qquad \text{(Eq. 33)}$$

Joint angles were calculated in order to a SCS, using a Cardan sequence of rotations x-y-z, for the ankle, knee and hip (Cole, Nigg, Ronsky and Yeadon, 1993). However, as Baker (2001) suggested, pelvis angles were computed using z-y-x rotation to describe the movement relatively to the laboratory coordinate system, that is, pelvis axial rotation, obliquity and tilt. To calculate and anatomically describe the joint angles between two adjacent coordinate systems a transformation matrix $[T]$ was used (equation 34),

$$[T] = \begin{bmatrix} cos\,\beta cos\gamma & cos\gamma sin\beta sin\alpha + sin\gamma cos\alpha & sin\alpha sin\gamma - cos\alpha sin\beta cos\gamma \\ -sin\gamma cos\beta & cos\alpha cos\gamma - sin\alpha sin\beta sin\gamma & sin\gamma sin\beta cos\alpha + cos\gamma sin\alpha \\ sin\beta & -cos\beta sin\alpha & cos\alpha cos\beta \end{bmatrix} \qquad \text{(Eq. 34)}$$

Where α represents the rotation about the mediolateral axes, β the rotation about the anteroposterior axes and γ the longitudinal rotation. The transformation matrix is given with numerical results with the following generic configuration (equation 35).

$$[T] = \begin{bmatrix} r11 & r12 & r13 \\ r21 & r22 & r23 \\ r31 & r32 & r33 \end{bmatrix} \quad \text{(Eq. 35)}$$

The angles can be obtained using the correspondence in equations 36 to 38.

$$\beta = sin^{-1}(r31) \quad \text{(Eq. 36)}$$

$$\alpha = -sin^{-1}\left(\frac{r32}{cos\beta}\right) \quad \text{(Eq. 37)}$$

$$\gamma = -sin^{-1}\left(\frac{r21}{cos\beta}\right) \quad \text{(Eq. 38)}$$

4.4.4 Statistical Analysis

To compare the results between the three models, a repeated measures ANOVA test was used. The level of statistical significance was set at $p \leq 0.05$ and all the tests were performed using the software PAWS Statistics 19.0.

4.5 Results

Table 6 presents the differences between groups, concerning segments' residuals, measured during the five gait cycles. No significant differences were found on pelvis or thigh residuals. On the other hand, the PG group had a lower shank residual and a higher foot residual, when compared with the NPG groups.

Table 6 - Segments' residuals (m) in the non-pregnant (NPG) and pregnant (PG) groups regarding the five gait cycles.

Segment	NPG	PG	p value
Pelvis	0.0031±0.0014 (Me=0.0028)	0.0033±0.0021 (Me=0.0027)	p=0.975
Right shank	0.0013±0.0005 (Me=0.0013)	0.0007±0.0006 (Me=0.0006)	p<0.001[b]
Right thigh	0.0019±0.0004 (Me=0.0021)	0.0019±0.0009 (Me=0.0020)	p=0.770
Right foot	0.0028±0.0006 (Me=0.0027)	0.0049±0.0031 (Me=0.0037)	p<0.001[b]

Me (median) for variables that were not normally distributed.

[b] A significant difference (level of significance $p<0.001$).

Regarding the segments' coordinates differences between the three models of joint constraints, for the NPG group (table 7), we can observe that the foot didn't revealed significant differences ($p=0.066$) between URS and SRS models, but when compared with the SSS model the differences were significant. The shank, thigh and pelvis coordinate differences, were higher in URS method, when compared with the other two as the SRS when compared with SSS.

Table 7 - Segment coordinates' differences (m) between the 6 degrees of freedom (6DOF) and the three constraint models in the non-pregnant group (NPG).

NPG	URS	SRS	SSS	F	p	Eta squared
Foot	0.0168±0.0150[3]	0.0075±0.0036[3]	0.0047±0.0022[1,2]	8.669	0.001[a]	0.338
Shank	0.0184±0.0073[2,3]	0.0115±0.0039[1,3]	0.0072±0.0027[1,2]	34.323	<0.001[b]	0.669
Thigh	0.0224±0.0080[2,3]	0.0126±0.0040[1,3]	0.0076±0.0017[1,2]	46.819	<0.001[b]	0.734
Pelvis	0.0146±0.0044[2,3]	0.0102±0.0024[1,3]	0.0080±0.0024[1,2]	33.334	<0.001[b]	0.662

URS – Universal-Revolute-Spherical; SRS – Spherical-Revolute-Spherical; SSS - Spherical-Spherical-Spherical; [a] A significant difference (level of significance p<0.05); [b] A significant difference (level of significance p<0.001).

In the PG (table 8), the URS method was the one that presented higher differences. The differences decreased significantly, with SRS and SSS methods for all the segments' coordinates.

Table 8 - Segment coordinates' differences (m) between the 6 degrees of freedom (6DOF) and the three constraint models in the pregnant group (PG).

PG	URS	SRS	SSS	F	p	Eta squared
Foot	0.0138±0.003[2,3]	0.0067±0.0020[1,3]	0.0035±0.0014[1,2]	193.667	<0.001[b]	0.919
Shank	0.0170±0.022[2,3]	0.0123±0.0030[1,3]	0.0063±0.0019[1,2]	153.601	<0.001[b]	0.900
Thigh	0.0212±0.0048[2,3]	0.0125±0.0025[1,3]	0.0080±0.0018[1,2]	90.703	<0.001[b]	0.842
Pelvis	0.0151±0.0040[2,3]	0.0087±0.0026[1,3]	0.0070±0.0023[1,2]	103.095	<0.001[b]	0.858

URS – Universal-Revolute-Spherical; SRS – Spherical-Revolute-Spherical; SSS - Spherical-Spherical-Spherical; [b] A significant difference (level of significance p<0.001).

Table 9 shows how the joints kinematics changes with different combination of constraints. Taking the 6DOF as the reference method, we found significant differences when comparing with URS, SRS and SSS methods. For the NPG, URS presented the lowest number of differences, where the ankle range of motion (ROM) in the sagittal plane increased about 5 degrees (p<0.001), the knee ROM on sagittal plane and hip maximum internal rotation were lower, 5 (p=0.039) and 6 degrees (p=0.035), respectively, and maximum pelvic right obliquity and the ROM in frontal plane, presented higher values, around 3 (p=0.002) and 5 degrees (p=0.020), respectively (figure 18). In the PG, URS was the model which revealed the highest number of variables with significant differences. Thus, regarding the ankle joint, maximum dorsiflexion angle during stance (ST) and swing phases (SW) increased around 2.5 (p=0.001) and 2 degrees (p=0.009), respectively, as the sagittal plane ROM which was 3 degrees higher. Ankle maximum eversion also increased 6 degrees (p=0.002) as the eversion/inversion ROM (p<0.001), which was higher in 8 degrees. No significant differences were found in the knee joint. Hip joint ROM on the sagittal plane increased around 3 degrees (p<0.001), as the maximum hip abduction angle that was 3.6

degrees (p=0.049) higher using the URS model. For the same joint, maximum internal rotation decreased 5 degrees (p=0.003) and the maximum external rotation increased 9 degrees (p<0.001). In the pelvis, some kinematic changes were found. Maximum posterior pelvic tilt decreased 1.5 degrees (p<0.001) and the sagittal ROM increased 1 degree. Maximum pelvic right and left obliquities increased about 1.3 (p=0.005) and 1.5 degrees (p=0.003), respectively, and the frontal plane ROM also suffered an increase of almost 3 degrees (p<0.001). In the transversal plane, maximum left and right rotations were higher around 1 degree (p=0.025) as the respective ROM, which increased 1.5 degrees (p=0.004) (figure 19).

Table 9 - Significant differences, in the maximum joint angles and range of motion (ROM) of stance phase (SP) and swing phase (SW), between 6 degrees of freedom (6DOF) and global optimization methods, for the non-pregnant group (NPG).

Maximum angle and ROM (degrees)	Method 1 (6DOF)	Method 2 (URS)	Method 3 (SRS)	Method 4 (SSS)	F	p	Eta squared
ankle dorsiflexion SP	87.91±3.82[3,4]	85.25±9.70	86.62±3.35[1]	86.39±3.65[1]	0.654	0.584	0.037
ankle dorsiflexion SW	80.85±4.41[3,4]	77.85±8.33	78.55±3.65[1]	77.92±3.56[1]	1.821	0.155	0.097
ankle plantarflexion SP	65.90±4.93[3,4]	58.92±12.70	62.09±4.78[1]	62.18±5.08[1]	3.109	0.034[a]	0.155
ankle plantarflexion SW	61.41±5.47[3,4]	53.18±13.77	56.29±5.10[1]	56.87±5.38[1]	3.665	0.018[a]	0.177
plantarflexion/dorsiflexion ankle ROM	26.50±3.63[2,3,4]	32.07±5.45[1]	30.33±3.74[1,4]	29.52±3.67[1,3]	23.778	0.000	0.583
ankle eversion	17.40±10.73	19.84±11.44	19.66±4.31	20.01±3.02	0.567	0.639	0.032
ankle inversion	8.40±2.48	6.54±10.69	8.04±3.03	8.92±2.17	0.640	0.593	0.036
eversion/inversion ankle ROM	11.69±2.91	13.30±4.19	11.62±2.99	11.10±3.19	9.006	0.001[a]	0.145
ankle adduction	-15.41±4.51[3,4]		-18.61±4.18[1]	-18.09±4.11[1]	9.006	0.001[a]	0.346
ankle abduction	3.83±3.96[3,4]		-1.83±4.53[1]	-3.63±3.79[1]	35.269	<0.001 [b]	0.675
ankle adduction/abduction ROM	19.24±4.20[3,4]		16.78±3.04[1,4]	14.45±3.00[1,3]	25.746	<0.001 [b]	0.602
knee flexion	66.43±4.73	63.98±4.70	63.98±4.21	63.68±4.30	2.419	0.077[a]	0.125
knee extension	6.39±4.82	9.04±4.72[3,4]	6.66±4.76[2]	6.28±4.89[2]	4.191	0.010[a]	0.198
knee flexion/extension ROM	60.04±4.60[2]	54.94±5.11[1,3,4]	57.32±5.04[2]	57.39±5.35[2]	5.483	0.002[a]	0.244
knee adduction	8.22±5.22[4]			3.54±3.90[1]	12.835	0.002[a]	0.430
knee abduction	-3.64±5.14			-6.31±4.21	2.914	0.106	0.146
knee adduction/abduction ROM	11.85±3.57			9.85±3.08	3.475	0.080	0.170
knee internal rotation	-11.08±7.80			-8.89±7.54	1.167	0.295	0.064
knee external rotation	4.67±5.82			5.00±5.95	0.044	0.836	0.003
knee internal/external rotation ROM	15.75±4.88			13.89±5.33	1.178	0.293	0.065
hip flexion	33.97±3.82	34.53±3.87	34.45±4.02	34.58±3.87	0.343	0.794	0.020
hip extension	-6.88±4.92	-8.28±4.25	-8.84±4.66	-8.71±4.85	1.535	0.217	0.083
hip flexion/extension ROM	40.85±3.73[4]	42.81±3.79[4]	43.29±4.08[4]	43.29±4.02[1,2,3]	2.956	0.041[a]	0.148
hip adduction	10.58±2.90	10.61±3.88[3,4]	13.56±3.60[2]	13.28±3.54[2]	10.405	0.005[a]	0.380
hip abduction	-7.93±2.40	-10.11±3.28	-9.23±3.36	-8.77±3.48	3.007	0.039[a]	0.150
hip adduction/abduction ROM	18.51±3.62	20.72±5.55[3]	22.79±5.41[2,4]	22.05±5.13[3]	4.563	0.007[a]	0.212
hip internal rotation	8.63±6.47[2]	2.71±5.14[1,3,4]	8.03±5.60[2]	10.90±5.77[2]	10.777	<0.001	0.388
hip external rotation	-10.91±7.20	-17.79±6.51[3,4]	-11.78±5.72[2]	-11.74±7.47[2]	5.684	0.002[a]	0.251
hip internal/external rotation ROM	15.76±4.88	20.50±5.33	19.80±5.22	21.83±5.73	0.906	0.355	0.051

anterior pelvic tilt	10.37±2.94	10.23±2.73[3,4]	10.72±3.05[2]	10.91±2.99[2]	0.408	0.748	0.023
posterior pelvic tilt	7.60±3.08	6.67±2.56[4]	7.16±2.69[4]	8.01±2.89[2,3]	1.226	0.310	0.067
pelvic tilt ROM	2.77±1.52	3.59±1.24[4]	3.56±1.40[4]	2.90±1.45[2,3]	3.315	0.027[a]	0.163
pelvic left obliquity	5.67±2.918	8.25±4.01	8.82±4.03[4]	8.43±4.04[3]	6.460	0.021[a]	0.275
pelvic right obliquity	-5.67±2.02[2,3,4]	-8.64±2.45[1]	-8.87±2.47[1,4]	-8.58±2.44[1,3]	18.212	<0.001	0.517
pelvic obliquity ROM	11.34±3.44[2,3,4]	16.89±5.20[1]	17.69±5.45[1,4]	17.01±5.34[1,3]	12.407	<0.001	0.422
pelvic transversal left rotation	6.83±3.11	8.44±2.72	7.06±3.13	6.95±3.17	1.567	0.209	0.084
pelvic transversal right rotation	-7.72±2.54	-7.25±3.10	-8.12±2.64	-7.95±2.60	0.380	0.768	0.022
pelvic rotation ROM	14.55±4.09	15.69±4.01[4]	15.18±4.25[4]	14.90±4.17[2,3]	0.021	0.886	0.001

6DOF - 6 degrees of freedom; URS – Universal-Revolute-Spherical; SRS – Spherical-Revolute-Spherical; SSS - Spherical-Spherical-Spherical; [a] A significant difference (level of significance p<0.05); [b] A significant difference (level of significance p<0.001).

Comparing the SRS with the 6DOF model, the NPG decreased the maximum dorsiflexion ankle angle during SP and SW, in 1 (p=0.001) and 2 degrees (p=0.003), respectively, as the maximum plantarflexion angle which decreased 4 degrees (p<0.001) during ST and 5 degrees (p<0.001) during SW. Ankle ROM on sagittal plane was 4 degrees higher (p<0.001). Also ankle angle maximum adduction increased 3 degrees (p=0.003) and the maximum abduction angle decreased about 6 degrees (p<0.001) as the adduction/abduction ROM, which presented 3 degrees (p=0.015) above the 6DOF value for the same variable. No significant differences were found in the knee and hip joints. Maximum right pelvic obliquity angle increased 3.2 degrees (p=0.002) as the frontal plane ROM that was around 5.5 degrees (p=0.009) higher with SRS method application (figure 17). In the PG, only the knee wasn't affected. In the ankle joint, maximum inversion angle decreased 5 degrees (p=0.001) as the maximum abduction angle with 6 degrees (p<0.001). The maximum ankle adduction angle was around 8 degrees higher (p<0.001). In the hip, sagittal ROM increased 3.5 degrees (p<0.001), as the maximum adduction angle and frontal ROM that were around 1 (p=0.028) and 3 degrees higher (p=0.041), respectively. In the pelvis, maximum posterior pelvic tilt decreased 0.5 degrees (p=0.008) and the sagittal ROM increased around 0.6 degrees (p=0.002). Maximum right and left obliquities increased about 2 degrees (p<0.001), as the respective plane ROM that was 3 degrees higher (p=0.011). Pelvic maximum right rotation was 0.3 degrees higher (p=0.045) as the pelvic transversal ROM which increased 0.7 degrees (p=0.041) (figure 19).

Table 10 - Significant differences, in the maximum joint angles and range of motion (ROM) of stance phase (SP) and swing phase (SW), between 6 degrees of freedom (6DOF) and global optimization methods, in the pregnant group (PG).

Maximum angle and ROM (degrees)	Method 1 (6DOF)	Method 2 (URS)	Method 3 (SRS)	Method 4 (SSS)	F	p	Eta squared
ankle dorsiflexion ST	83.98±3.76[2]	86.51±3.37[1,3,4]	84.58±3.48[2]	84.82±3.70[2]	13.746	<0.001[b]	0.447
ankle dorsiflexion SW	77.45±3.33[2]	79.50±2.16[1,3,4]	77.75±2.06[2,4]	76.64±2.26[2,3]	14.679	<0.001[b]	0.463
ankle plantarflexion ST	63.41±4.83	63.53±4.67[4]	62.62±4.32	62.38±3.97[2]	1.897	0.142	0.100
ankle plantarflexion SW	56.40±6.16	55.53±5.14[4]	55.37±5.35[2]	54.77±4.91	1.073	0.369	0.059
plantarflexion/dorsiflexion ankle ROM	27.57±4.04[2,4]	30.98±4.47[1]	29.21±4.79	30.04±4.43[1]	5.893	0.002[a]	0.257
ankle eversion	17.15±3.04[2]	23.48±7.99[1,3,4]	15.79±5.82[2]	17.89±2.85[2]	16.271	<0.001[b]	0.489
ankle inversion	6.36±3.73[3]	4.40±4.50[3]	1.69±4.19[1,2,4]	5.93±2.53[3]	15.350	<0.001[b]	0.474
eversion/inversion ankle ROM	10.79±2.22[2]	19.07±5.11[1,4]	14.10±5.13[2]	11.95±2.54[2]	20.085	<0.001[b]	0.542
ankle adduction	-13.50±4.32[3,4]		-21.37±5.58[1,4]	-15.98±4.55[1,3]	35.080	<0.001[b]	0.674
ankle abduction	2.11±6.60[3,4]		-3.30±5.24[1,4]	-0.57±6.02[1,3]	22.390	<0.001[b]	0.568
ankle adduction/abduction ROM	15.61±4.04		18.07±3.48	15.41±3.69	4.980	0.013[a]	0.227
knee flexion	59.33±7.73	63.48±4.832[4]	62.10±5.90	62.49±4.67[2]	3.930	0.013[a]	0.188
knee extension	0.79±4.15	3.37±5.32[3]	0.41±3.78[2]	1.05±3.67	7.479	<0.001[b]	0.306
knee flexion/extension ROM	58.54±7.58	60.11±5.52[3]	61.69±6.11[2]	61.44±4.90	2.494	0.070	0.128
knee adduction	8.09±3.72			6.86±4.28	3.016	0.101	0.151
knee abduction	-1.33±3.36			-1.54±4.21	0.040	0.843	0.002
knee adduction/abduction ROM	9.42±3.60			8.40±3.32	1.488	0.239	0.080
knee internal rotation	-4.11±4.59			-6.17±6.74	2.711	0.118	0.138
knee external rotation	12.62±5.75			11.42±6.02	0.636	0.436	0.036
knee internal/external rotation ROM	16.73±7.15			17.60±5.52	0.145	0.708	0.008
hip flexion	38.95±5.20	40.42±5.19	40.10±5.46	40.35±5.29	2.198	0.100	0.115
hip extension	-1.79±5.79	-3.35±5.27	-4.10±5.61	-3.80±5.70	3.438	0.024[a]	0.168
hip flexion/extension ROM	40.73±2.73[2,3,4]	43.77±3.17[1]	44.20±3.44[1]	44.16±3.51[1]	36.938	<0.001[b]	0.685
hip adduction	10.43±2.64[3,4]	9.18±2.53[3,4]	11.82±2.30[1,2]	12.51±2.68[1,2]	23.558	<0.001[b]	0.581
hip abduction	-5.78±4.37[2]	-9.39±4.87[1,4]	-7.68±4.94[4]	-6.60±5.49[2,3]	6.076	0.001[b]	0.263
hip adduction/abduction ROM	16.20±2.95[3]	18.58±4.31	19.51±4.76[1]	19.11±4.82	5.860	0.002[b]	0.256
hip internal rotation	5.23±6.09[2]	-0.24±4.71[1,3]	5.90±5.49[2]	4.68±6.744	11.804	<0.001[b]	0.410
hip external rotation	-11.60±8.21[2]	-20.58±4.72[1,3,4]	-12.63±7.05[2]	-13.79±9.32[2]	21.533	<0.001[b]	0.559
hip internal/external rotation ROM	16.82±5.21	20.34±5.02	18.53±6.04	18.47±6.60	3.351	0.026[a]	0.165
anterior pelvic tilt	17.07±4.60	16.61±4.49[3,4]	17.13±4.37[2]	17.02±4.27[2]	5.479	0.002[a]	0.244
posterior pelvic tilt	14.52±4.60[2,3]	13.01±4.47[1,3,4]	14.01±4.40[1,2,4]	14.40±4.52[2,3]	30.034	<0.001[b]	0.639
pelvic tilt ROM	2.55±0.86[2,3]	3.60±1.03[1,3,4]	3.12±0.88[1,2,4]	2.62±1.00[2,3]	17.974	<0.001[b]	0.514
pelvic left obliquity	5.50±1.79[2,3,4]	6.83±2.51[1]	7.40±2.32[1]	7.11±2.33[1]	23.167	<0.001[b]	0.577
pelvic right obliquity	-4.23±2.55[2,3,4]	-5.78±3.48[1]	-6.00±3.10[1]	-5.84±3.09[1]	15.935	<0.001[b]	0.484
pelvic obliquity ROM	9.73±2.12[2,3,4]	12.61±3.62[1]	12.97±3.75[1]	12.94±3.74[1]	8.520	<0.001[b]	0.334
pelvic transversal left rotation	5.38±2.92[2]	6.07±3.05[1,3,4]	5.76±3.11[2]	5.56±2.89[2]	8.456	<0.001[b]	0.332
pelvic transversal right rotation	-6.86±2.92[2,3]	-7.63±3.43[1,3,4]	-7.20±3.15[1,2]	-7.00±3.06[2]	11.697	<0.001[b]	0.408
pelvic rotation ROM	12.23±4.19[2,3]	13.70±4.95[1,3,4]	12.97±4.76[1,2]	12.57±4.42[2]	15.153	<0.001[b]	0.471

6DOF - 6 degrees of freedom; URS – Universal-Revolute-Spherical; SRS – Spherical-Revolute-Spherical; SSS - Spherical-Spherical-Spherical; [a] A significant difference (level of significance p<0.05); [b] A significant difference (level of significance p<0.001).

Regarding the SSS model, the NPG group revealed a larger number of variables that showed significant differences. The ankle was the most affected joint. Thus, maximum dorsiflexion angle in both ST and SW phases decreased in 2 ($p<0.001$) and 3 degrees ($p<0.001$), respectively. Also the maximum plantarflexion ankle angle was lower in 4 ($p<0.001$) and 5 degrees ($p<0.001$). The sagittal ROM increased 3 degrees ($p<0.001$). The maximum adduction and abduction ankle angles revealed the same trend as the SRS model. The first variable increased 3 degrees ($p=0.004$) and the second one decreased 8 degrees ($p<0.001$). In the same plane, adduction/abduction decreased 5 degrees (figure 18). The PG presented significant differences regarding maximum adduction ($p=0.003$) and abduction ($p=0.007$) angles, where the first variable increased and the second one decreased, both, in 3 degrees.

Still in the frontal plane, maximum knee adduction angle decreased around 5 degrees ($p=0.002$). Sagittal hip joint ROM increased around 3 degrees. In the pelvis, maximum right obliquity and frontal plane ROM were 3 ($p=0.004$) and 1 degrees higher ($p=0.021$), respectively. In the PG, the comparison between 6DOF and SSS models showed no differences in the hip joint, and it was the one that revealed the lowest number of variables with significant differences. In the ankle, sagittal and frontal planes ROM increased 2 degrees ($p=0.004$). In the knee no significant differences were found. Maximum pelvic right ($p<0.001$) and left ($p<0.001$) obliquity angles increased about 2 degrees (figure 19).

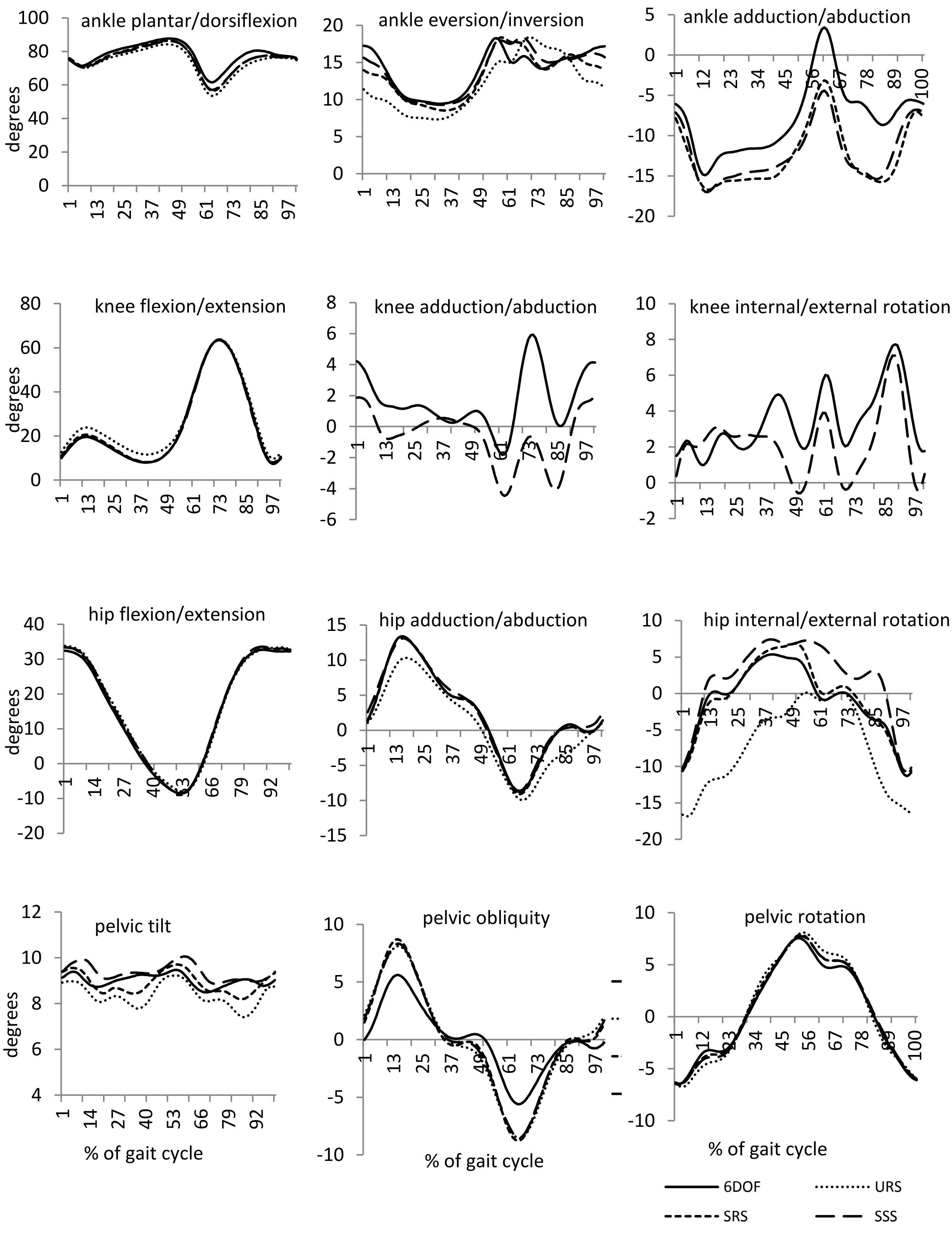

Figure 18 - Ankle, knee and hip joints and pelvis range of motion (ROM) in sagittal (first column), frontal (second column) and transversal planes (third column) for the non-pregnant group (NPG). The solid line represents the 6 degrees of freedom (6DOF) model; the round dot line the Universal-Revolute-Spherical (URS) model, the square dot the Spherical-Revolute-Spherical (SRS) model; and the dash line the Spherical- Spherical-Spherical (SSS) model.

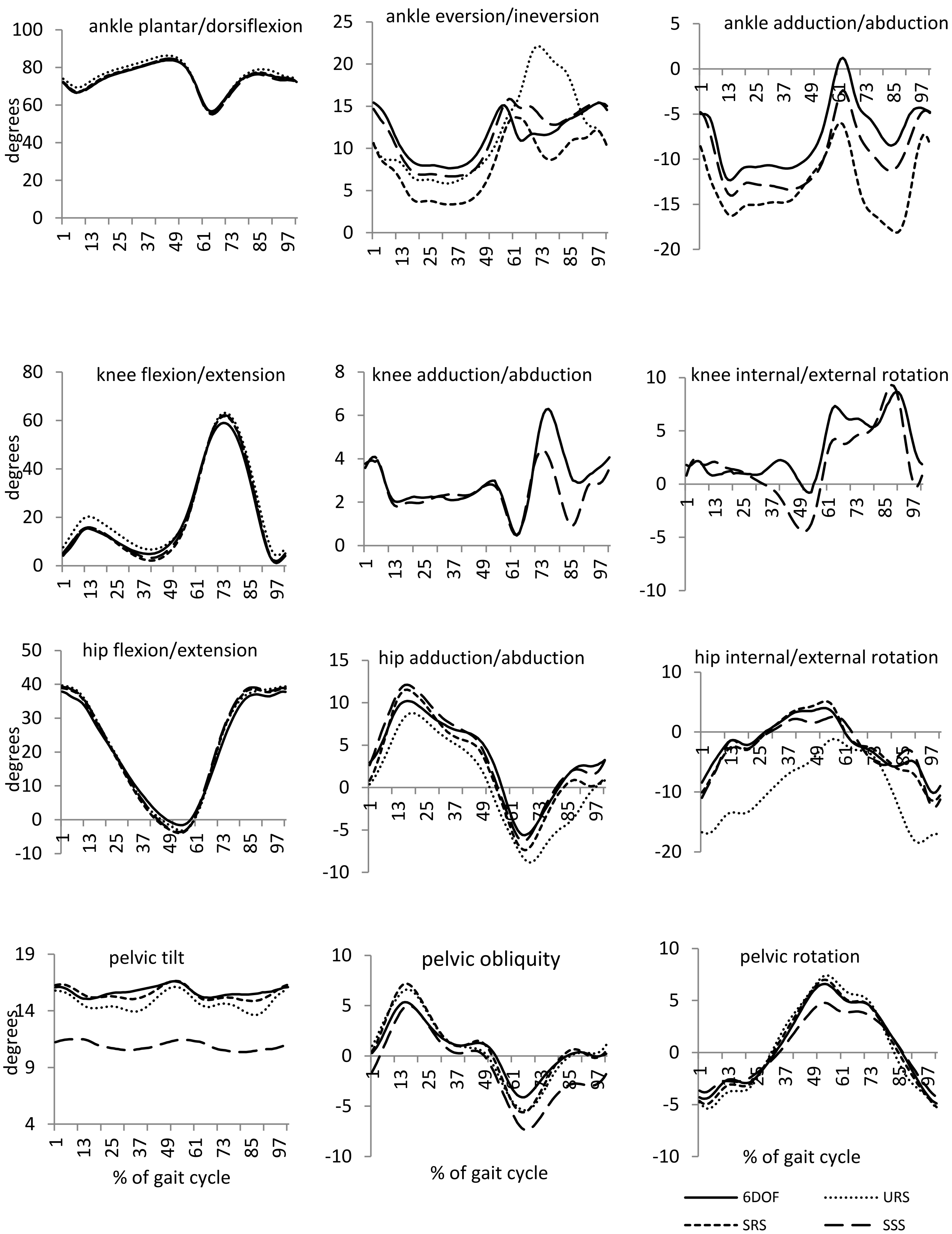

Figure 19 - Ankle, knee and hip joints and pelvis range of motion (ROM) in sagittal (first column), frontal (second column) and transversal planes (third column) for the pregnant group (PG). The solid line represents the 6 degrees of freedom (6DOF) model; the round dot line the Universal-Revolute-Spherical (URS) model, the square dot the Spherical-Revolute-Spherical (SRS) model; and the dash line the Spherical- Spherical-Spherical (SSS) model.

4.6 Discussion

The main goal of this study was to use the GOM application in the gait kinematic analysis of the pregnant women in order to obtain a more accurate kinematic model. The GOM imposes joint constraints in a multilinked musculoskeletal model and is capable of giving a more accurate segment position and orientation estimation, taking into consideration that joint kinematics are affected by the STA (Lu & O'Connor, 1999).

The major goal of GOM application was to minimize the effect of STA on the kinematics of gait. From the 6DOF model it was possible to obtain the amount of STA, given by the segment residuals. It was assumed that STA was higher in pregnant women, and the results showed that this fact was observed in the foot, maybe because of the swelling (Ponnapula & Boberg, 2010), causing an higher inertial effect, or also could have hindered the marker's locations (Capello et al., 2005), maybe due the anthropometrics characteristics associated to pregnancy. Also the overweight related to pregnancy increase the ground reaction forces and the foot impact, which in turn may increase the foot bones vibrations (Leardini et al., 2005). However STA was lower in the shank segment, suggesting that the foot may be the major source of impact absorption when walking during pregnancy. Non-pregnant women seem to distribute more the bones vibrations for the adjacent segments. In the pelvis and thigh segments, no differences were found between the two groups, probably because in these segments the anthropometric changes related to the pregnancy, such as the increased volume of the trunk, did not influenced the amount of STA.

With respect to the segment general coordinate differences, in non-pregnant women group, the foot was not affected with the URS and SRS models. The transversal foot rotation was the only constraint that differentiated these two methods. In the same group and still analyzing the foot, the SSS model presented lower differences. On the other segments, the differences decreased as the model is less constrained, this is from URS to SSS. On the pregnant women group, for all the segments, the differences are smaller as the model is less constrained. The GOM changes the SCS position relatively to the 6DOF model and the SSS combination of joints seems to be the setups that remain the segments positions most similar to the 6DOF approach.

Assuming that the 6DOF model is the one with the highest error associated to the STA, it seems that in NPG women, the SSS model is the method more appropriated since it revealed an increased number of differences regarding the kinematic variables. Although the differences between coordinates are lower in the SSS model, the use of three spherical joints, that is, a less number of joint constraints, seems to reduce the STA effect in this population performing this particular motor task such as walking. The ankle was the most affected segment when modifying the constraints, particularly the ankle plantar/dorsiflexion (Duprey, Cheze & Dumas, 2010) and adduction/abduction movements using the SRS and SSS models.

As Leardinin et al. (2005) suggested, to better understand the STA phenomenon it is necessary to characterize a large series of measurements on different subject populations. GOM was used to improve gait kinematic analysis in pregnant women. In the pregnancy population there is the possibility of higher STA, as in the foot, given the morphological circumstances that this condition provides. We could observe that the pregnancy caused more significant differences between 6DOF and URS models mainly in the pelvis ROM. In the pregnant women specific case, the URS model seems to affect more the kinematic variables when compared with the 6DOF approach. Assuming that there is a larger kinematic error associated with the pregnancy condition, the URS model is the one that affects more joint angular positions with expectation of the knee joint. Anderson et al. (2010) reported that the inclusion of a revolute joint increased the mean flexion/extension joint angle error. In the URS model, the knee was defined as a revolute joint, but only the SSS model influenced the knee flexion/extension ROM. This may point to a different knee joint configuration, regarding the URS combination. However URS model seems to be more appropriated to the second trimester pregnant women.

In both groups, maximum pelvic right obliquity changed in the three models. Maximum pelvic left obliquity changed only in the pregnant women group.

Regardless the population, the manipulation of joints degrees of freedom influences the kinematic variables, even of the adjacent joints, considering that the biomechanical model is an assembly of rigid bodies acting as a kinematic chain.

Despite the 6DOF model can provide the segment residuals as the amount of STA, currently, only invasive devices and fluoroscopy can quantify the real soft tissue movement (Gao & Zheng, 2008), but they are not appropriated to apply in pregnant women, since this population is particularly vulnerable to medical imaging radiation exposure which may cause deleterious effects (Goodman & Amurao, 2012). Regarding this, the comparison of the 6DOF approach with the GOM application (and particularly three specific joint constraints models) which is a non-invasive method, can be a starting point to provide more information about this issue in this specific population.

This research aimed to provide new information about gait and the joint kinematics of the pregnant women, but future directions for research might be needed.

4.7 References

Anderson, M., Benoit, D., Damsgaard, M., Ramsey, D. & Rasmussen, J. (2010). Do kinematic models reduce the effects of soft tissue artifacts in skin marker-based motion analysis? An in vivo study of knee kinematics. Journal of Biomechanics 43, 268-273.

Baker, R. (2001). Pelvic angles: a mathematically rigorous definition which is consistent with a conventional clinical understanding of the terms. Gait and Posture 13, 1-6.

Bell, A.L., Pederson, D.R., & Brand, R.A. (1989). Prediction of hip joint center location from external landmarks. Human Movement Science. 8, 3-16.

Bell, A.L., Pedersen, D.R. & Brand, R.A. (1990). A Comparison of the accuracy of several hip center location prediction methods. Journal of Biomechanics. 23, 617-621.

Block, R.A., Hess, L.A., Timpano, E.V. & Serlo, C., (1985). Physiological changes in the foot in pregnancy. Journal of the American Podiatric Medical Association, 75, 297–299.

Brett, M. & Baxendale, S., (2001). Motherhood and memory: a review. Psychoneur-oendocrinology 26, 339–362.

Butte, N., Ellis, K., Wong, W., Hopkinson, J. & Smith, E. (2003). Composition of gestational weight gain impacts maternal fat retention and infant birth weight. American Journal of Obstetrics and Gynecology 189(5), 1423-1432.

Capozzo, A., Cappello, A. Della Croce, U. & Pensalfini, F. (1997). Surface-Marker Cluster Design for 3-D Bone Movement Reconstruction. IEEE Transactions on Biomedical Engineering, 44(12), 1165-1174.

Cereatti, A., Della Corce, U. & Capozzo, A. (2006). Reconstruction of skeletal movement using skin markers: comparative assessment of bone pose estimators. Journal of Neuroengenieering Rehabilitation, 23, 36-7.

Calganeri, M., Bird, H.A. & Wright, V. (1982). Changes in joint laxity occurring during pregnancy. Annals of the Rheumatic Diseases, 41, 126–128.

Cappello, A., Francesco, P., Palombara, L. & Leardini, A. (1996). Optimization and smoothing techniques in movement analysis. International Journal of Biomedical Computation, 41, 137-151.

Capello, A., Stagni, R., Fantozzi, S. & Leardini, A. (2005). Soft tissue artifact compensation in knee kinematics by double anatomical landmark calibration: performance of a novel method during selected motor tasks. IEEE Transactions on Biomedical Engineering, 52(6), 992-998.

Challis, J.H. (1995). A procedure for determining rigid body transformation parameters, Journal of Biomechanics, 28, 733-737.

Charlton, I.W., Tate, P., Smyth, P. & Roren, L. (2004). Repeatability of an optimised lower body model. Gait and Posture, 20 (2), 213–221.

Cole, G.K., Nigg, B.M., Ronsky, J.L. & Yeadon, M.R. (1993). Application of the joint coordinate system to three-dimensional joint attitude and movement representation: a standardization proposal. Journal of Biomechanical Engineering, 115, 344-9.

Czaplicki, A., Silva, M. & Ambrósio, J. (2004). Biomechanical modeling for whole body motion using natural coordinates. Journal of Theoretical and Applied Mechanics, 42(4), 927-944.

de Groot, R.H., Adam, J.J. & Hornstra, G., (2003). Selective attention deficits during human pregnancy. Neuroscience Letters, 340, 21–24.

Dumas, G.A., Reid, J.G., Wolfe, L.A., Griffin, M.P. & McGrath, M.J., (1995). Exercise, posture, and back pain during pregnancy. Clinical Biomechanics, 10, 98–103.
Dunning, K., LeMasters, G., Levin, L., Bhattacharya, A., Alterman, T. & Lordo, K., (2003). Falls in workers during pregnancy: risk factors, job hazards, and high risk occupations. American Journal of Industrial Medicine, 44, 664–672.
Duprey, S.; Cheze, L. & Dumas, R., (2010). Influence of joint constraints on lower limb kinematics estimation from skin markers using global optimization. Journal of Biomechanics, 43, 2858-2862.
Foti, T.; Davids, J. & Bagley, A. (2000). A biomechanical analysis of gait during pregnancy. The Journal of Bone and Joint Surgery, 82-A(5), 625-632.
Fuller,J., Liu, L.J., Murphy, M.C. & Mann, R.W. (1997). A comparison of lower-extremity skeletal kinematics measured using skin- and pin-mounted markers. Human Movement Science 16(2-3), 219-242.
Gao,B. & Zheng, N. (2008). Investigation of soft tissue movement during level walking: translations and rotations of skin markers. Journal of Biomechanics 41, 3189-3195.
Goodman, T.R. & Amurao, M. (2012). Medical imaging radiation safety for the female patient: rationale and implementation. Radiographics. 32(6), 1829-37.
Hainline, B. (1994). Low back pain in pregnancy. Advances in Neurology. 64, 65–76.
Holden, J.P., Orsini, J.A., Siegel, K.L., Kepple, T.M., Gerber, L.H. and Stanhope, S.J. (1997). Surface movement errors in shank kinematics and knee kinetics during gait. Gait and Posture 5(3), 217-227.
Hyten, F. & Leitch, I. (1971). The Physiology of Human Pregnancy (2nd ed.). Oxford Blackwell Scientific.
Jensen, R.K., Doucet, S. & Treitz, T. (1996). Changes in segment mass and mass distribution during pregnancy. Journal of Biomechanics, 29(2), 251-256.
Leardini, A., Chiari, L., Della Croce, U. & Capozzo, A. (2005). Human movement analysis using stereophotogrammetry. Part 3. Soft tissue artifact assessment and compensation. Gait and Posture, 21(2), 212-225.
Lu, T.W. & O'Connor, J.J.(1999). Bone position estimation from skin marker coordinates using global optimization with joint constraints. Journal of Biomechanics, 32, 129-134.
McNitt-Gray, J.L., (1991). Biomechanics related to exercise in pregnancy. Exercise in pregnancy (2nd ed.). Williams & Wilkins, Baltimore, 133-140.
Pitkin, R. (1976). Nutritional support in obstetrics and gynecology. Clinical Obstetrics and Gynecology, 19(3), 489-513.
Ponnapula, P. & Boberg, J. (2010). Lower extremity changes experienced during pregnancy. The Journal of Foot & Ankle Surgery, 49, 452–458.
Shultz, R., Kedgley, A.E. & Jenkyn, T.R. (2011). Quantifying skin motion artifact of the hindfoot and forefoot marker clusters with the optical tracking of a multi-segment foot model using single-plane fluoroscopy. Gait and Posture, 34, 44-48.
Snijders, C.J., Seroo, J.M., Snijder, J.G. & Hoedt, H.T. (1976). Change in form of the spine as a consequency of pregnancy. In: Proceedings for the 11th International Conference on Medical and Biological Engineering. Ottawa, Ontario, Canada, 670-671.
Spoor, C.W. & Veldpaus, F.E. (1980). Rigid body motion calculated from spatial coordinates of markers. Journal of Biomechanics, 13(4), 391- 393.
Taggart, N., Holliday, R., Billewicz, W., Hytten, F. & Thomson, A. (1967). Changes in skinfolds during pregnancy. British Journal of Nutrition, 21(2), 439-451.
Veldpaus, F.E., Woltring, H.J. & Dortmans, L.J.M.G. (1988). A leastsquares algorithm for the equiform transformation from spatial marker co-ordinates. Journal of Biomechanics, 21, 45-54.
Winter, D.A. (1991). The biomechanics and motor control of human gait: normal, elderly and pathological. Waterloo, Ontario: University of Waterloo Press.

Chapter 5

5 Foot Contact Model to Predict Vertical Ground Reaction Forces of Gait Across Pregnancy and Post-partum[4]

[4] Submitted as: Aguiar L, Santos-Rocha R, Veloso AP (2014). Foot contact model to predict vertical ground reaction forces of gait across pregnancy and post-partum. Computer Methods in Biomechanics and Biomedical Engineering.

5.1 Abstract

The vertical ground reaction forces (GRF) are a common indicator for external biomechanical load, and a parameter to obtain the joint moments of force. When GRF are unobtainable, simulation is an useful tool to provide these data. This study aimed to predict vertical GRF generated during gait, measured in the three trimesters of pregnancy and post-partum. The used contact model allows understanding the interaction forces between the foot and the floor surface. The foot-ground contact consisted in five spring-dampers located under each phalanx of the foot and one spring-damper located under the heel. The vertical GRF were computed for five pregnant women, in four periods: first trimester, second trimester, third trimester and post-partum. In general, the model overestimated vertical GRF, especially during the loading response. The increased force values may be related to an excessively high stiffness value, which can be tuned for all the instants. One of the anthropometric adaptation in pregnant women is the foot swelling, increasing the foot mass and consequently reducing the stiffness during the contact, once that the foot bones are more compressed by soft tissue. An higher penetration can also be considered. This can also suggest that pregnant women use more cushion support during stance phase. This methodology can be used in future research about the development of a specific shoe for pregnant and post-partum women, regarding all the morphological and hormonal changes, that can be responsible for uncomfortable or painful walking.

Keywords: vertical ground reaction forces, stance phase, simulation, pregnancy.

5.2 Introduction

Human gait is a motor task intensively studied over several years in various fields. In biomechanics, populations with a specific characteristic have particular attention regarding gait analysis. The human body can suffer physical changes as consequences of a physiological process which in turn may affect locomotion. Pregnancy, as a natural and physiological process, produces in women changes involving the motor system. There is very little literature on biomechanics of gait during pregnancy and lack of consensus among scientists in body adaptation in this period (Forczek & Staszkiewicz, 2012). More research in this area remains to be done.

During pregnancy, morphological changes can modify the gait pattern. Some of this are weight gain, an anterior displacement in the location of the center of mass, increased ligamentous laxity, decreased neuromuscular control and coordination, swelling of the arms and legs, altered biomechanical parameters, decreased abdominal muscle strength, increased spinal lordosis, and changes in mechanical loading and joint kinetics (Block et al., 1985; Brett & Baxendale, 2001; Calganeri et al., 1982; deGroot et al., 2003; Dumas et al., 1995; Dunning et al., 2003; Hainline, 1994; McNitt-Gray, 1991; Snijders et al., 1976). Other studies confirmed these changes and describe that more than 50% of women reported swelling of the foot, ankle, and leg, unsteady gait, increased foot width and hip pain (Ponnapula & Boberg, 2010). An anterior shift in the location of the center of mass (Enders, Berger, Chambers, Redfern and McCrory, 2009), and changes in mechanical loading and joint kinetics have been noted in pregnant women (Lymbery & Gilleard, 2005; Nyska et al., 1997). These changes affect the posture and gait pattern of pregnant women compared with non-pregnant women (Forczek & Staszkiewicz, 2012). Lymbery and Gilleard (2005) found that, associated to pregnancy, the medio-lateral position of the center of pressure during stance shifts laterally.

Hereupon, the assessment of biomechanical loading in the musculoskeletal system of pregnant women is of particular interest and few biomechanical studies can be found in literature. Foti et al. (2000), used 3-D motion and force platform data of 15 women during the end of the third trimester, and again one year after delivery. Kinematic and kinetic parameters were compared between the two conditions. It was found that the gait pattern was remarkably changed during pregnancy.

The mass of model limbs can be changed to study the effect of the segment mass on a movement (Bach, 1995; Tsai & Mansour, 1986). As example, is to test the torso mass on ground reactions forces (Robertson, Caldwell, Hamill, Kamen & Whittlessey, 2004). This can be applied to analyze pregnant women gait pattern, regarding all the morphological adaptations associated to pregnancy condition.

To analyze human gait, ground reaction forces (GRF) and motion data are measured. Both parameters are used to calculate other variables, as joint moments which are calculated by the inverse dynamics method and add more information about the stress imposed in the joints and the necessary muscle control (Perry, 1992). Also, the GRF are a common indicator for external biomechanical load.

Anthropometric data, kinematic data and external forces (e.g. GRF) are required as inputs to use in the inverse dynamic technique (Liu, Shih, Tian, Zhong & Li, 2009; Ren, Jones & Howard, 2008). The GRF curves can also add more kinetic information about the gait pattern concerning how the foot interacts with the floor.

However, there are situations where it is not possible to obtain external forces. In some sports, or in clinical contexts it can be problematic to measure GRF, due to the inexistence of a force platform. Usually an installed force platform in a laboratory setting is used to measure GRF data and occupies a large space, such as a long way walking, thus, it can be limited to the analysis of particular human movements which are continuous or other sports motor tasks (Ren, Jones & Howard, 2008). Also, the force platform is an expensive equipment and is used on a fixed space causing some limitations on the data collection (O'Donovan, Kamnik, O'Keeffe & Lyons, 2007). It is possible to predict GRF data, using mathematical models that are applied in a simulation method of human movement, which can predict dynamic responses. When GRF are unobtainable, simulation is an useful and powerful tool to provide these data, which can add more information for the understanding of interactions of the musculoskeletal system with the physical environment, such as the foot floor contact.

In a clinical context, this methodology can be important to provide more important biomechanical information that could not be collected in another way.

The GRF is a three-dimensional vector, however, in this study, only vertical component was predicted. The vertical force curve, shows a characteristic double hump, which represents the upward acceleration force of the center of gravity during early stance, a reduction in downward force during the mid-stance, and a second peak due to deceleration as the downward motion in the late stance (Whittle, 2001). The horizontal forces called as shear forces occur in the anterior-posterior and medio-lateral planes and have influence on stability by generating the adequate friction. However, the horizontal forces magnitude is small when compared with the vertical loading (Perry, 1992). Regarding the input variables for the inverse dynamics, the horizontal component has relatively less importance than vertical direction since that is less than 10% of the vertical component (Song, Kim, Kim & Lee, 2010) and has little influence on joint moment (Audu, Kirsch & Triolo, 2007).

5.3 Objectives

This study aimed to predict vertical ground reaction forces generated during gait, measured in the three trimesters of pregnancy and post-partum. The used contact model allows understanding the interaction forces between the foot and the floor surface.

5.4 Materials and Methods

5.4.1 Subjects

The sample size consisted of five healthy pregnant women, without any clinical contraindications for physical exercise. All women were fully informed about the aims and procedures and gave informed consent to participate voluntarily in the study. The inclusion criteria of participants were to: had no history of lower limb injuries of either orthopedic or neurological conditions; have more than 18 years old and less than 40; be in first trimester of an healthy pregnancy; be available to go to the laboratory of biomechanics and functional morphology, three times during the pregnancy (first trimester, second trimester and third trimester) and once the in post-partum period (PP).

As a longitudinal study, the data was collected during the pregnancy period, in the three trimesters and in post-partum. Table 11 describes the participants' characteristics in the four moments of data collection.

Table 11 - Characterization of sample group (mean ± standard deviation): age, weight, height, body mass index (BMI) and gestational age.

	Age (years)	Weight (kg)	Height (m)	BMI (kg/m^2)	Gestational age (weeks)
T1	34.0±2.3	60.9±3.9	1.65±0.05	22.4±1.8	14.5±4.7
T2	34.0±2.3	66.8±5.5	1.65±0.05	24.6±2.8	27.6±1.2
T3	34.4±2.5	71.5±4.7	1.65±0.05	26.3±2.7	35.9±0.7
PP	34.6±2.3	62.1±4.2	1.65±0.05	22.8±2.2	18.2±3.1

T1 = first trimester; T2 = second trimester; T3 = third trimester; PP = post-partum.

5.4.2 Data Collection and Processing

Anthropometric data (weight and height) were collected by ISAK (International Society for the Advancement of Kinanthropometry) certified kinanthropometrists, to calculate de body segments masses and inertia moments. Reflective spherical markers were placed on the pelvis and lower limbs, on anatomical points according to Plug-in-Gait setup protocol (figure 20).

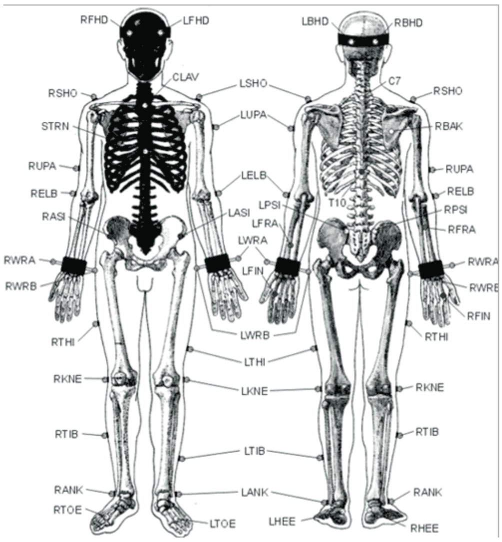

Figure 20 - Plug-in-Gait Marker Placement Protocol.

Motion capture was performed with an optoelectronic system of twelve cameras Qualisys (Oqus-300) operating at a frame rate of 200 Hz, synchronized with two force platforms (Kistler AG, Winterthur, Switzerland) and one AMTI (Advanced Mechanical Technology, Inc, Watertown, MA), which collected ground reaction force (GRF) data. The participants performed three non-consecutive minutes, walking at a comfortable speed, with a break of thirty seconds between each trial.

5.4.3 Model Construction

For each subject, a rigid multibody model was developed with the software tools of the MSC ADAMS (Mechanical Dynamics, Inc.) namely, the Lifemodeler (Lifemodeler, Inc.), allowing the model construction and the prediction of GRF data.

The multibody model was constructed with seven individual rigid segments: pelvis, two thighs, two legs and two feet. Lifemodeler provided the GeBOD database to generate human data sets, based on descriptive variables of the subject, such as age, height, weight and gender. These data allowed compute the body segment's geometric and mass properties, the joint locations and mechanical properties, using regression equations from anthropometric surveys and stereophometric data (Cheng, Obergefell & Rizer, 1994).

To connect the rigid bodies, ideal passive joints were used to add kinematic constraints to the model. The joint types used were spherical, revolute and universal, for the hip, knee and ankle joints, respectively. Stiffness and damping were the torsional viscoelastic parameters defined for passive joints response. Table 12, shows the stiffness and damping parameters defined for each rotational degree of freedom by using the Amankwah et al. (2004) equations, except of the ankle inversion/eversion and hip rotations. These last parameters were proposed by Nazer et al. (2008). Damping values were defined has 10% of the stiffness values.

These joints were used in an inverse dynamics analysis to record the joint angulations while the model was manipulated with motion agents that followed the experimental markers coordinates.

Table 12 - Joint stiffness (K) and damping (C) values.

	hip		**knee**		**ankle**	
	K	**C**	**K**	**C**	**K**	C
Sagittal	0.7	0.07	0.27	0.027	0.21	**0.021**
Transverse	0.8	0.08	-	-	-	**-**
Frontal	**1.5**	**0.15**	**-**	**-**	**10**	**1**

The foot-ground contact consisted in five spring-dampers located under each phalanx of the foot and one spring-damper located under the heel. The contact normal force calculation was based on the equation 39.

$$Fn = K * (g ** e) + \ Step\ (g, o, o, d_{max}, c_{max}) dg/dt \qquad \text{(Eq. 39)}$$

where *g* represents the penetration of one geometry into another, *dg/dt* is the penetration velocity at the contact point, *e* is a positive real value denoting the force exponent and d_{max} is a positive real value specifying the boundary penetration to apply the maximum damping coefficient c_{max} and *Step* is mathematic discontinuous function whose value is zero for negative and one for positive arguments and is used to represent a signal that can switch on at a specified period of time and stays switched on indefinitely. The stiffness (*K*), damping (*C*) and full damping depth (d_{max}) values were assumed to be 150.000N/m, 2.000Ns/m and 0.001m, respectively (Gilchrist &

Winter, 1996; Nazer et al., 2008). Transverse forces were based on a relatively simple velocity-based friction model (ADAMS, 2005).

From the experiment marker setup, the markers position over time were used for the definition of the motion agents' trajectories in ADAMS/LifeMOD. These motion agents are motion influencers and not motion governors, which accommodates the geometric differences between the body model and the actual human subject, as well as motion agent location discrepancies. The motion agents file contains the motion agents' setup, its location on the model segments, and the respective motion data. To fit the model to the initial data position, it was performed an equilibrium analysis that allowed the adjustment of the motion agents to the motion data. Inverse dynamics was computed once the model had the motion agents attached to the segments and synchronized to the motion data. They give bounded movement to the model, which is simultaneously interacting with the floor. After the inverse dynamics, the motion agents were removed from the model and the recorded joint angles were used to calculate joint torques and functions forces. With the joint angle history recorded, a proportional-derivate (PD) controller was used, to produce a torque which allowed the simulation. These PD (Kp and Kv, in figure 19) elements can influence the model, optimizing the simulation. The default values provided by the software were used to minimize the error between the desired instantaneous joint angle and the recorded model joint angle (Lifemode, 2008).

5.4.4 Inverse and Forward Dynamic Analysis

A rigid body is a particles system, for which, the distance between them is constant in time (figure 21). The rigid bogy condition must ensure that the distance between points does not change over time (equation 40).

Figure 21 - Foot segment, represented as a rigid body.

$$\Phi = \begin{cases} (x_j - x_i)^2 + (y_j - y_i)^2 - L_{ij}^2 = 0 \\ (x_k - x_i)^2 + (y_k - y_i)^2 - L_{ik}^2 = 0 \\ (x_k - x_j)^2 + (y_k - y_i)^2 - L_{jk}^2 = 0 \end{cases} \qquad \text{(Eq. 40)}$$

Regarding kinematic analysis of position, once that all the constraints are defined, was necessary to determine, for all time instants, the joint coordinate vector, that satisfy the followed equation system (equation 41), where $\Phi(q,t)$, is the kinematic constraints global vector.

$$\Phi(q,t) = 0 \qquad \text{(Eq. 41)}$$

To solve the system it was used Newton-Raphson method, and to a certain initial aproximation, is obtained equation 42.

$$\Phi(q,t) \cong \Phi(q_i) + \Phi_q(q_i)(q - q_i) = 0 \qquad \text{(Eq. 42)}$$

Φ_q is the constraints Jacobian matrix, showed in equation 43.

$$\Phi_q = \begin{bmatrix} \frac{\partial \Phi_1}{\partial q_1} & \cdots & \frac{\partial \Phi_1}{\partial q_n} \\ \vdots & \ddots & \vdots \\ \frac{\partial \Phi_m}{\partial q_1} & \cdots & \frac{\partial \Phi_m}{\partial q_n} \end{bmatrix} \qquad \text{(Eq. 43)}$$

Where *m* is the constraints equation number and *n* the joint coordinate number. Derivating the constraints equations with respect to time, is was possible to obtain the equations that allow to analyse the multibody system velocity (equation 44).

$$\frac{d}{dt}\big(\Phi(q,t)\big) = 0 \Leftrightarrow \Phi_t + \Phi_q(q,t)q = 0 \Leftrightarrow \Phi_q(q,\dot{t})\dot{q} = -\Phi_t \equiv v \qquad \text{(Eq 44)}$$

Where $\dot{q}$ is the velocity vector and v is the parcial differentiating vector, with respect to time. The acceleration vector $\ddot{q}$ was obtained by the second differentiating, with respect to time, of constraint vector (equation 45).

$$\Phi_q(q,t)\ddot{q} = \frac{dv}{dt} - \Phi_q\dot{q} \equiv \gamma \qquad \text{(Eq. 45)}$$

Thus, we have equation 46:

$$\Phi(\ddot{q},\dot{q},\ddot{q},t) = \frac{d\Phi(\dot{q},\dot{q},t)}{dt} = \Phi_q\ddot{q} + (\Phi_q\dot{q})_q\dot{q} + v_t = 0 \qquad \text{(Eq. 46)}$$

Using the Equations of Motion, inverse dynamics allows estimate forces and moments aplied to a rigid body, that produce a certain aceleration and forwad dynamics estimate new positions. The kinematic and kinetic description were obtained by derivation of movement equations, applying the methods of Lagrangian dynamics described above (equation 47). The motion equations were integrated, resulting in a model simulation (equation 48).

$$\tau = ID(model, q, \dot{q}, \ddot{q}) \qquad \text{(Eq. 47)}$$

$$\ddot{q} = FD(model, q, \dot{q}, \tau) \qquad \text{(Eq. 48)}$$

Lagrange's equations of motion (equations 49 and 50), as assembled by MSC ADAMS are:

$$M\ddot{q} + \Phi_q^T\lambda - A^T F(q, \dot{q}) = 0 \iff M\ddot{q} + \Phi_q^T\lambda = g \qquad \text{(Eq. 49)}$$

$$\phi(q, t) = 0 \qquad \text{(Eq. 50)}$$

Where $M\ddot{q}$ are the inertial forces, M is the mass matrix, $\ddot{q}$ is the vector of generalized acelerations, Φ_q is the constaints Jacobian matrix, λ is the vector of Lagrange multipliers and g is the generalized external force vector.

Joints moments of force were calculated by inverse dynamics and were used to guide de model during the forward dynamics analysis. The software provides servo controller elements, which acts as a feedback control system to optimize the simulation (figure 22).

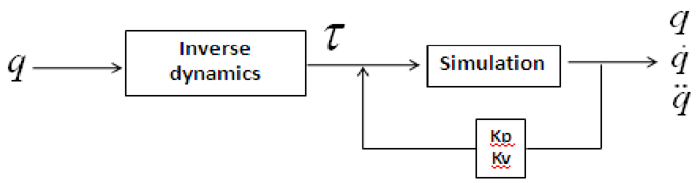

Figure 22 - Representation of the closed-loop feedback control system to optimize the simulation.

5.4.5 Statistical Analysis

The comparisons between the experimental and simulated data, were carried out with Wilcoxon test. Statistical tests were performed using the PAWS Statistics 21.0. The $p<0.05$ was used to denote statistical significance.

5.5 Results

In this section, the results of the foot contact model determined to predict vertical GRF of gait across pregnancy and post-partum period, are presented. Table 13 shows the loading response peak force (F1), mid stance response peak force (F2) and terminal stance response peak force (F3) from the experimental and simulated data, for the five subjects in the four trials (figures 23 and 24). Notice that F1 and F3 are the highest values of the vertical ground reaction force vector, during the loading and terminal stance phases, respectively and F2 is the lowest value of the same curve occurred during the mid-stance phase.

Comparing the experimental with the simulated data from 1T trial, subject 4 presented the highest differences, between peaks. Simulated F1 was 386% higher, F2 was 64% lower and F3 was 90% higher. The subject 1 presented the lowest difference in F1, with a 13% higher simulated value, 16% increased simulated value for F2 and 33% for F3. Subject 2 also presented a F1 peak 37% higher simulated force, however, F2 was 3% lower, and again F3 was 20% higher. Subject 3 showed increased simulated peak forces, with 58%, 21% and 27% for F1, F2 and F3 respectively. Peak GRF for subject 5 where 67% higher for F1, 18% and 27% lower for F2, and F3 respectively (figure 23).

Regarding 2T trial, subject 5 presented the highest F1 simulated peak GRF, which was 122% higher than the experimental value. Also F2 increased in 27% and F3 in 69%. Subject 1 showed also increased values, with 32%, 19% and 6% for F1, F2 and F3 respectively. The F1 and F2 peak forces for subject 2 where 35% and 23% higher, but F3 decreased in 2%. Subject 3 presented higher simulated forces, where F1 increased in 44%, F2 in 22% and F3 in 34%. Simulated peak forces where 33% and 61% higher for subject 4, regarding F1 and F3, respectively. F2 deceased in 7% (figure 23).

The T3 trial, revealed the highest differences between experimental and simulated GRF peak forces. Thus, F1 was the most overestimated peak, where all subjects revealed an increased value. Subject 1 showed a 348% increased value, subject 2

33%, subject 3 45%, subject 4 167% and finally subject 5 212%. Mid stance response, was higher in subjects 1, 2, 3 and 5, 28%, 10%, 7% and 17%, respectively. In subject 4, F2 decreased in 45%. F3 was also overestimated in subjects 1, 3, 4 and 5, 70%, 19%, 89% and 43%, respectively. In subject 2, this value was practically the same (figure 23).

The last trial data was obtained during PP, and F1 was higher for all the subjects, with the expectation of subject 1, which simulated value was 5% lower than the experimental. For the same subject, F2 also decreased in 4% and F3 increased in 17%. Subject 2 showed a 75% F1 higher value, 62% increased F3 value as the F3 peak force, which was 37% higher. Subject 3 presented a very high F1 peak, 214% overestimated, but F3 was almost the same and F2 17% higher. As subject 3, subject 4 also revealed an 224% higher F1 peak force, 56% lower F2 value and 93% higher F3 value. Subject 5 also had an overestimated F1 value in 139%, 52% F2 higher value and 70% increased F3 peak force value (figure 23).

By a root mean square analysis, subject 2 presented the lowest differences between the curves values. Subject 3 showed a very similar force curve pattern, in all the trials. However, there is an initial contact peak force, which was the highest value during the loading response, like it is quite visible on figure 23, on the same subject (PP).

Statistical differences were found in 1T ($p=0.043$), 2T ($p=0.043$) and 3T ($p=0.043$) trial between experimental and simulated data, where the last one presented higher values, regarding loading response peak force F1. On PP trial terminal stance response F3, also was significantly higher ($p=0.043$) regarding simulated data. No significant differences were found for F2 peak forces, and for F3, regarding 1T, 2T and 3T trials (table 13).

Table 13 - Statistical analysis between the experimental and simulated data for loading response peak force (F1), mid stance response peak force (F2) and terminal stance response peak force (F3) and for the four trials (first trimester, second trimester, third trimester and post-partum).

		1T		2T		3T		PP	
Loading response F1 (BW)	Exp Sim	1.07 ± 0.03 2.25 ± 1.58	p=0.043[a]	1.09 ± 0.02 1.67 ± 0.42	p=0.043[a]	1.05 ± 0.07 2.69 ± 1.20	p=0.043[a]	1.13 ± 0.07 2.55 ± 0.97	p=0.080
Mid stance response F2 (BW)	Exp Sim	0.79 ± 0.05 0.71 ± 0.26	p=0.686	0.80 ± 0.09 0.93 ± 0.15	p=0.080	0.77 ± 0.05 0.79 ± 0.20	p=0.500	0.78 ± 0.14 0.86 ± 0.30	p=0.686
Terminal stance F3 (BW)	Exp Sim	1.17 ± 0.04 1.50 ± 0.51	p=0.138	1.27 ± 0.38 1.11 ± 0.06	p=0.080	1.10 ± 0.08 1.58 ± 0.38	p=0.080	1.20 ± 0.04 1.73 ± 0.48	p=0.043[a]

BW = bodyweight; Exp = experimental; Sim = simulation; T1 = first trimester; T2 = second trimester; T3 = third trimester; PP = post-partum; [a] p values were considered statistical significant.

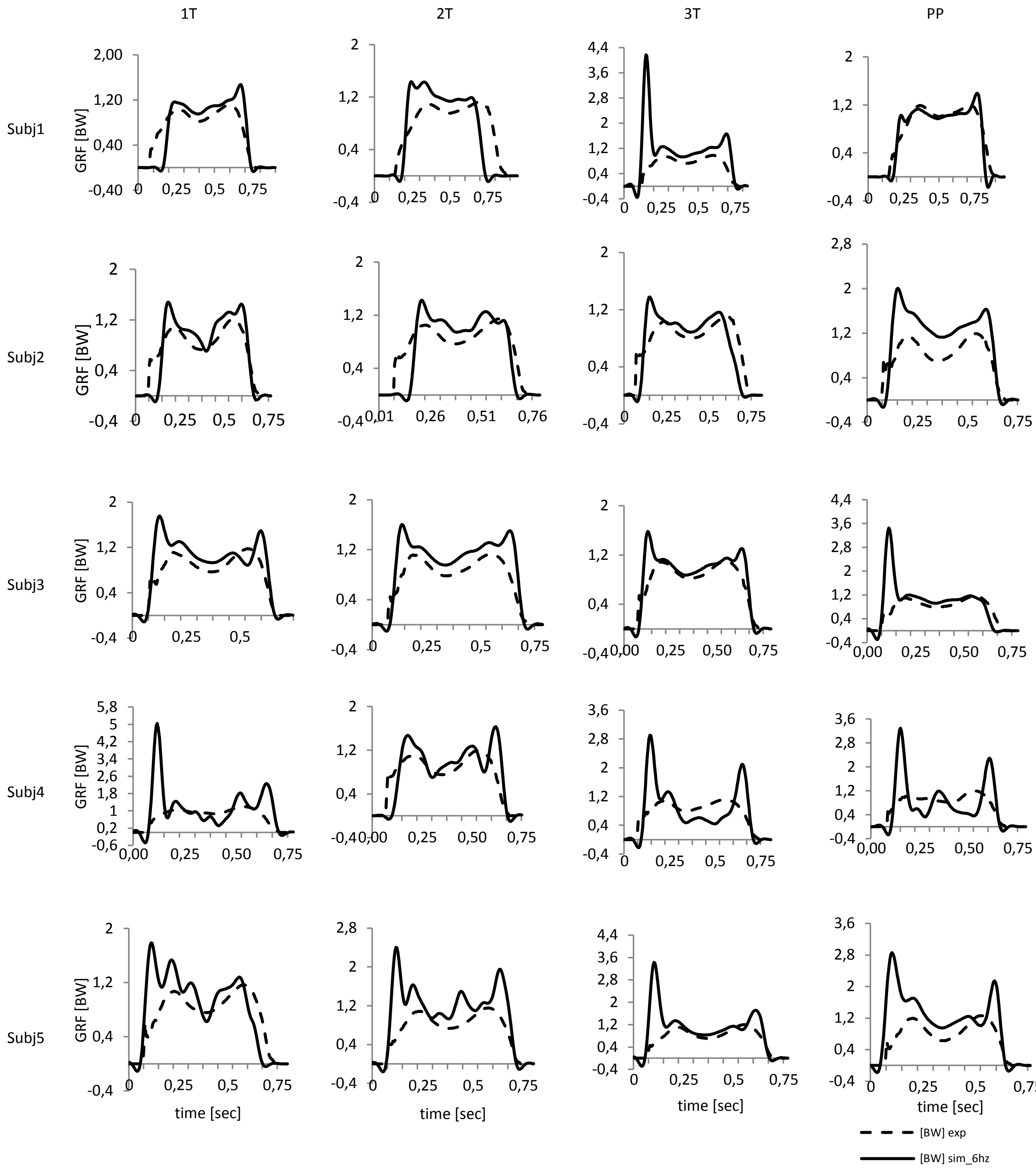

Figure 23 - Ground reaction forces curves, of the collected experimental and simulated data, for all the subjects and regarding the four moments: first trimester (1T), second trimester (2T), third trimester (3T) and post-partum (PP). The dashed line represents the experimental data. The solid line represents the simulation data.

5.6 Discussion

The purpose of this study was to use a simulation technique to predict vertical ground reaction forces generated during gait in pregnant women. It was used motion data as the input and also anthropometric data. A contact model was used to calculate the external forces, which were the output data. Several parameters were manipulated, and remain constant among the subjects and trials allowing an easier comparison of the obtained results.

The contact normal force calculation depended on stiffness, damping and penetration physical parameters. These parameters were chosen based on Gilchrist and Winter (1996) and Nazer et al. (2008) studies for barefoot condition. The results showed how the parameters can be tuned, to get a more accurate model, concerning three stages of pregnancy and post-partum.

Although the foot was represented by a rigid body, the contact model included nonlinear displacement-dependent and viscous damping terms, allowing more foot freedom movement (Gilchrist & Winter, 1996). This aspect contributed to a good progression of the foot relatively to the floor, and create a continuous vertical force curve.

Regarding curve pattern, the model generated the two peak forces, during the loading and terminal stance responses, and the valley during the mid-stance phase created by the rise of the center of gravity as the body moves forward and accentuated by the contralateral limb swinging (Perry, 1992). These happened for all the subjects. However, it can be observed that, immediately after the initial contact, the curve gets the shape of the experimental loading peak force. The peak force at the initial contact may be influenced by anthropometric foot data and by the motion data, especially in subjects 4 and 5. One other possibility is the soft tissue artifact at initial contact could be higher, contributing to a abrupt peak force, which seems to dissipate over the foot contact progression.

In general, the model overestimated vertical GRF, being very evident during the loading response. Stiffness is a mechanical propriety that represents the resistance of the foot to deformation during the contact with the floor surface. The increased force values may be related to an excessively high stiffness value, which can be tuned for all the instants. In pregnant women, one of the anthropometric adaptations is the foot swelling, increasing the foot mass and consequently reducing the stiffness during the contact, once that the foot bones are more involved by the soft tissue. An higher penetration can also be considered. This can also suggest that pregnant women use more cushion support during stance phase.

No significant differences during mid-stance loading response between experimental and simulated data, but initial contact normal force was significantly higher in the simulated results. These can be related to the anthropometric measures of the foot, which were not used in the model construction. Also related to this issue, it is known that during pregnancy the foot width increases and this can also increase the contact surface area and consequently the force distribution. Another concern was the fact that the foot segment was modeled as on rigid body, hindering the smooth progression of the contact phase.

On the terminal stance response, the model showed a more robust behavior but there is still a trend to overestimate the ground reaction forces. Refinements should be done on the model to accommodate the transition between swing phase to foot floor contact and vice versa.

Some curve oscillations were also observed, like in subjects 4 and 5 on first and second trials. This phenomenon can be related to the ankle joint stability. During pregnancy, joint laxity increases, creating more unsteady gait. The increased stride width during pregnancy in order to provide more stability (Foti et al., 2000; Bird et al., 1999) can influence the joint kinetic behavior resisting more or less to the movement. To represent the ankle it was used a universal joint, allowing dorsi and plantar flexion in the sagittal plane, and inversion and eversion in the frontal plane. The passive stiffness and the respective damping parameters can also be readjusted and in the case of subject 3 these two values should be higher, to create more stability to the ankle joint.

The used contact model allowed tuning stiffness and damping parameters, which are biomechanical properties that represent the resistance of a certain elastic body to the deformation. In this study, the main focus was the foot of pregnant women, which supports high mechanical loads either due to the increasing body weight or due to locomotion. For pregnant women, walking is the main way of locomotion and it can be difficult in different types of floor as sand or stiffer. This can be an orthopedic issue, and lead to more research about the contact problem.

To add more information, it is important to know also horizontal shear forces during the contact, which create the necessary friction between the feet and the floor surface and provide stability avoiding sliding and potential fallings. It is also important to provide the center of pressure path, once that gives the information about the subject balance and the progression patters (Perry, 1992).

The use of plantar pressure data could also be considered to improve the model, through the segmentation of the foot in specific areas, taking into account the morphological characteristics of the foot associated to pregnancy.

To develop a smooth foot contact, a multi-segmented foot model should be also used. Leardini et al. (2007) proposed a new protocol in order to track a large number of foot segments during stance phase. That could be helpful to understanding the dynamics of the pregnant foot during stance phase.

The used methodology can be used to future research about the development of a specific shoe for pregnant women, regarding all the morphological and hormonal changes that can be responsible for uncomfortable or painful walking.

5.7 References

ADAMS/View, MSC Software, 2005.

Amankwah, K., Triolo, R.J. & Kirsch, R. (2004). Effects of spinal cord injury on lower limb passive joint moments revealed through a nonlinear viscoelastic model. Journal of Rehabilitation Research and Development, 41, 15-32.

Audu, M.L., Kirsch, R.F., & Triolo, R.J. (2007). Experimental verification of a computational technique for determining ground reactions in human bipedal stance. Journal of Biomechanics, 40, 1115-1124.

Bach, T.M. (1995). Optimization mass and mass distribution in lower limb prostheses. Prosthetics and Orthotics Australia, 10, 29-34.

Bird A, Menz H & Hyde C. (1999). The effect of pregnancy on footprint parameters. A prospective investigation. Journal of the American Podiatric Medical Association, 89, 405-409.

Block, R.A., Hess, L.A., Timpano, E.V. & Serlo, C.(1985). Physiological changes in the foot in pregnancy. Journal of the American Podiatric Medical Association, 75, 297–299.

Brett, M. & Baxendale, S. (2001). Motherhood and memory: a review. Psychoneuroendocrinology, 26, 339–362.

Calganeri, M., Bird, H.A. & Wright, V. (1982). Changes in joint laxity occurring during pregnancy. Annals of the Rheumatic Diseases, 41, 126–128.

Cheng, H., Obergefell, L. & Rizer A. (1994) Generator of Body Data (GEBOD) Manual. Dayton, OH: Systems Research Labs, Inc.

de Groot, R.H., Adam, J.J. & Hornstra, G. (2003). Selective attention deficits during human pregnancy. Neuroscience Letters, 340, 21–24.

Dumas, G.A., Reid, J.G., Wolfe, L.A., Griffin, M.P. & McGrath, M.J. (1995). Exercise, posture, and back pain during pregnancy. Clinical Biomechanics, 10, 98–103.

Dunning, K., LeMasters, G., Levin, L., Bhattacharya, A., Alterman, T. & Lordo, K. (2003). Falls in workers during pregnancy: risk factors, job hazards, and high risk occupations. American Journal of Industrial Medicine, 44, 664–672.

Enders, L., Berger, K., Chambers, A., Redfern, M. & McCrory, J. (2009). Biomechanical evidence of waddling during pregnancy. In: Proceedings of the BMES 2009 annual fall scientific meeting.
Forczek, W. & Staszkiewicz, R. (2012). Changes of kinematic gait parameters due to pregnancy. Acta of Bioengineering and Biomechanics, 14, 113-119.
Foti, T. ; Davids, J. & Bagley, A. (2000). A Biomechanical Analysis of Gait During Pregnancy. The Journal of Bone and Joint Suergery, Incorporated, 82-A(5), 625-632.
Gilchrist, L.A. & Winter, D.A. (1996). A two-part, viscoelastic foot model for use in gait simulations. Journal of Biomechanics, 29, 795-798.
Hainline, B. (1994). Low back pain in pregnancy. Advances in Neurology, 64, 65–76.
Leardini, A., Benedetti, M.G., Berti, D., Bettinelli, R., Nativo, R. & Giannini, S. (2007). Rear-foot, mid-foot and fore-foot motion during the stance phase of gait. Gait and Posture, 25, 453-462.
Liu, Y., Shih, S.-M., Tian, S.-L., Zhong, Y.-J. & Li, L. (2009). Lower extremity joint torque predicted by using artificial neural network during vertical jump. Journal of Biomechanics, 42, 906-911.
Lymbery, J.K. & Gilleard, W. (2005). The stance phase of walking during late pregnancy: temporospatial and ground reaction force variables. Journal of the American Podiatric Medical Association, 95, 247–53.
LifeMOD. (2008), Biomechanics Research Group Inc.
McNitt-Gray, J.L.(1991). Biomechanics related to exercise in pregnancy, Exercise in Pregnancy (2nd ed.). Williams & Wilkins, Baltimore, 133-140.
Nazer, R.A., Rantalainen, T., Heinonen, A., Sievanen, H. & Mikkola, A. (2008). Flexible multibody simulation approach in the analysis of tibial strain during walking. Journal of Biomechanics, 41, 1036–1043.
Nyska, M., Sofer, D., Porat, A., Howard, C.B., Levi, A. & Meizner, I. (1997). Planter foot pressures in pregnant women. Israel Journal of Medical Sciences, 33, 139–46.
O'Donovan, K.J., Kamnik, R., O'Keeffe, D.T., & Lyons, G.M. (2007). An inertial and magnetic sensor based technique for joint angle measurement. Journal of Biomechanics, 40, 2604 – 2611.
Perry, J. (1992) Gait Analysis. Normal and Pathological Function. Thorofare, New Jersey: SLACK Inc.
Ponnapula, P. & Boberg, J. (2010). Lower Extremity Changes Experienced During Pregnancy. The Journal of Foot & Ankle Surgery, 49, 452–458.
Ren, L., Jones, R. K., & Howard, D. (2008). Whole body inverse dynamics over a complete gait cycle based only on measured kinematics. Journal of Biomechanics, 41, 2750-2759.
Robertson, D., Caldwell, G., Hamill, J., Kamen, G. & Whittlesey, S. (2004). Research Methods in Biomechanics. Human Kinetics, Champaign, IL, USA.
Snijders, C.J., Seroo, J.M., Snijder, J.G. & Hoedt, H.T. (1976). Change in form of the spine as a consequency of pregnancy. In: Proceedings for the 11th International Conference on Medical and Biological Engineering. 670-671. Ottawa, ONT, CAN.
Song, S.J., Kim, S. Y., Kim, Y.T., & Lee, S. D. (2010). Computation of ground reaction forces during gait using kinematic data. KSME-A, 34, 431-437.
Tsai, C.-S. & Mansour, J.M. (1986). Swing phase simulation and design of above knee prostheses. Journal of Biomechanical Engineering, 108, 65-72.
Whittle, M.W. (2001). Gait Analysis, an Introduction, Butterworth-Heinemann Ltd

Chapter 6

6 General Discussion and Conclusions

6.1 General Discussion and Conclusions

Human gait is strongly studied in biomechanics, once that is the main mean of locomotion, but can be influenced by several factors as anatomical changes, orthopedic disorders or by the different stages of growth. Regarding pregnancy, as a temporary condition, women are subject to morphological, physiological and hormonal changes, which can lead to adaptations in gait. From the 20th week onward, changes due to pregnancy are already well established and are clearly visible (Pitkni, 1976; Butte et al., 2003). To add more information to this field of study, several objectives were established (chapter 1).

Biomechanical models can help in the assessment of joint mechanical loads, kinetics and kinematics during gait and provide the knowledge about movement patterns. In chapter 2 an inverse dynamics method was used to describe joint range of motion (ROM) and joint moments of force (M_f) during second trimester of pregnancy using a 3-D biomechanical model, and from that study it was possible to conclude that both ankle and hip joint seem to be more overloaded, mainly in the sagittal and frontal plane, respectively. It was possible to observe similar joint angular displacement and joint M_f of force patterns, as well as the maximum ROM and peak M_f values, when data were compared with other studies related to pregnancy (Foti et al., 2000).

The used methodology was appropriated to describe the gait pattern of pregnant women, but some issues still need to be explored, once that literature demonstrated that there are indeed changes in gait biomechanical parameters during pregnancy (Foti et al., 2000; Branco et al., 2013). Foti et al. (2000) used three-dimensional biomechanical models to compare kinematic and kinetic parameters between the end of the third trimester, and again one year after delivery and reported an increase in the following variables: hip moment of force, hip joint power in the frontal and sagittal planes, maximum ankle plantar flexion M_f, and maximum ankle plantar flexion power absorption. An increase was also observed in the use of the abductor and extensor muscles of the thigh and in the use of the ankle plantar flexor muscles. It was also found that stride width increased.

The literature on biomechanics of gait during pregnancy is scarce and there is a lack of consensus among scientists about body adaptation in this period (Forczek & Staszkiewicz, 2012). There was a need of clarifying which gait adaptations are more either related with the increased trunk mass, or which can be more associated to physiological and hormonal changes. Thus, to add more information about this issue a comparison between the overweight condition due to pregnancy and due to an added extra weight condition to simulate body mass distribution in pregnancy was performed (chapter 3). To understand the effect of the increased mass in the trunk associated to the second trimester of pregnancy, during gait, an extra load was added in the abdomen and breasts of a non-pregnant women group, providing a simulation of this condition and taking into account only this anthropometric characteristic. Some variables showed clear relation with the extra weight: the trunk weight gain increased both stance phase duration and double limb support time in both load carrying and pregnant groups. The body's response to the external hip flexor moment is related to a higher extensor activity to support the anterior additional mass of the trunk and to the forward translation of the center or mass. The pregnant women had several months of adaptation to a gradual weight increase, and hormonal and physiological changes that adapt the body to extra load, whereas in load carrying the subject is not changed apart from the extra weight. Related to pregnancy, three variables were found: maximum knee extension during the stance phase, maximum knee extension during the swing phase and peak hip adduction M_f. The changes in variables that can be explained by the hormonal changes or by the increased trunk weight were: stride width, maximum ankle eversion, eversion/inversion ankle range of motion, flexion/extension knee range of motion, maximum anterior pelvic tilt, maximum posterior pelvic tilt, peak ankle dorsiflexion moment of force, peak ankle inversion moment of force and peak hip external rotation moment of force. Stride width is higher in pregnant women, thus creating a larger base of support and thereby providing more stability.

As stated in a previous study (Branco et al., 2013), kinematic analysis is widely used in the study of human gait, but there were some concerns about biomechanical modelling related to several sources of error that affect joint kinematics, such as instrumental inaccuracies, anatomical landmark mislocation and soft tissue artifact. Inertial effects, skin deformation and sliding, gravity and muscle contraction contribute independently for soft tissue artifact (Capello et al., 2005). Leardini et al. (2005) and Ceratti et al. (2006) found that soft tissue artifact have a frequency component in the same region as the bone motion, and is subject and task dependent. As a specific population, pregnant women present anthropometric characteristics that lead to methodological issues related to model building. Holden et al. (1997) verified that the amount of soft tissue artifact depends on the physical characteristics of the individuals and Fuller et al. (1997) demonstrated that it could also depend on the nature of the movement task. Based on the anthropometric changes associated with pregnancy the hypothesis that the soft tissue artifact is increased in pregnant women, when compared with non-pregnant women, was raised. To understand this phenomenon and how different joint constraint models affect the kinematics of gait in pregnant women, a global optimization method was used with three different sets of joint constraints (chapter 4). The ankle, knee and hip joints were modeled respectively with the following characteristics: 1) universal-revolute-spherical (URS), 2) spherical-revolute-spherical (SRS) and 3) spherical-spherical-spherical (SSS). The gait of two groups was analyzed: 1) second trimester pregnant women and 2) non pregnant women. Considering that the six degrees of freedom model is the one with the highest error associated to the soft tissue artifact, it seems that in non-pregnant women, the SSS model is the method more appropriated since it revealed an increased number of differences regarding the kinematic variables. Although the differences between coordinates, using both six degrees of freedom and joint restricted models, are lower in the SSS model, the use of three spherical joints, that is, a less number of joint constraints, seems to reduce the soft tissue artifact effect in this population performing this particular motor task. The ankle was the most affected segment when modifying the constraints, particularly the ankle plantar/dorsiflexion (Duprey et al., 2010) and adduction/abduction movements using the SRS and SSS models. Regarding pregnant women, we could observe that the pregnancy caused more significant differences between the six degrees of freedom and URS models mainly in the pelvis ROM. In the pregnant women specific case, the URS model seems to affect more the kinematic variables when compared with the six degrees of freedom approach. Assuming that there is a larger kinematic error associated with the pregnancy condition, the URS model is the one that affects more joint angular positions with expectation of the knee joint. Anderson et al. (2010) reported that the inclusion of a revolute joint increased the mean flexion/extension joint angle error. In the URS model, the knee was defined as a revolute joint, but only the SSS model influenced the knee flexion/extension ROM. This may point to a different knee joint configuration, regarding the URS combination. Regardless the population, the manipulation of joints degrees of freedom influences the kinematic variables, even of the adjacent joints, considering that the biomechanical model is an assembly of rigid bodies acting as a kinematic chain.

From the kinetic point of view, to analyze human gait, ground reaction forces and motion data are measured. Both parameters are used to calculate other variables, such as joint moments which are calculated by the inverse dynamics method and may add more information about the stress imposed in the joints and the consequent muscle control (Perry, 1992). Anthropometric and inertial parameters data, kinematic data and external forces are required as inputs to use in the inverse dynamic technique and the ground is used as the external force (Liu, Shih, Tian, Zhong & Li, 2009; Ren, Jones & Howard, 2008). The ground reaction forces curves can also add more kinetic information about the gait pattern, and in particularly concerning how the foot interaction with the floor. Finally, the last study (chapter 5) aimed to predict the vertical ground reaction forces generated during gait, measured throughout pregnancy and post-partum. The used contact model allowed understanding the interaction forces between the foot and the floor surface. Simulation techniques were used to estimate GRF, which can be quite helpful when GRF are unobtainable or difficult to assess. In a clinical context, this methodology can also be important in order to provide more important biomechanical information that could not be collected in another way. The contact normal force calculation depended on stiffness, damping and penetration physical parameters. These parameters were chosen based on Gilchrist and Winter (1996) and Nazer et al. (2008) studies for barefoot condition. The results showed how the parameters can be tuned, to get a more accurate model, concerning three stages of pregnancy and post-partum. In general, the model overestimated vertical GRF, especially during the loading response. Stiffness is a mechanical propriety that represents the resistance of the foot to deformation during the contact with the floor surface. The increased force values may be related to an excessively high stiffness value, which can be tuned for all the instants. In pregnant women, one of the anthropometric adaptations is the foot swelling, increasing the foot mass and consequently reducing the stiffness during the contact, once that the foot bones are more involved by the mass. Higher penetration can also be considered. This can also suggest that pregnant women use more cushion support during stance phase. The used contact model allowed tuning stiffness and damping parameters, which are biomechanical properties that represent the resistance of a certain elastic body to the deformation. In this study, the main focus was the foot of pregnant women, which supports high mechanical loads either due to the increasing body weight or due to locomotion. Furthermore, walking, being the main way of locomotion of pregnant women, may be more difficult in different types of ground, such as sand or in a stiffer floor. This can be an orthopedic issue, and lead to more research about the biomechanical loading.

The results of the present work will be useful to further biomechanical research with regard to gait of this special phase of life and eventually, for the prevention of orthopedic discomfort, by providing data for shoes development.

6.2 Recommendations for Future Research

The results of the present work will be useful to further the biomechanical research with regard to gait of this special phase of life: pregnancy and post-partum.

Some issues need to be explored, as the relation between anterior trunk extra weight and biomechanical gait modifications. Data from more subjects will be needed to confirm the results provided in the study presented in chapter 2 and in a previous study (Branco et al., 2013). Young adult women simulated the second trimester of pregnancy, and used additional load in breast and anterior trunk (chapter 3). The mass distribution can be optimized as the adaption to it. It is important to provide more anthropometric data, particularly of the trunk segment, from either pregnant and non-pregnant women, to improve the biomechanical models and provide more information.

Also, the inertial properties may have to be readjust to create a more representative biomechanical model of the pregnant segments, that are significantly different in pregnant women. This is a problem that needs to be studied under the anthropometric point of view.

Regarding biomechanical models, pregnant women present particular morphological characteristics, which can influence the model construction and also the joint's range of motion and moments of force calculation. Joint centers are calculated based on the marker's locations but the anthropometric changes may increase the error associated with the marker placement (chapter 3). This issue needs to have particular attention, mainly the pelvis segment, that readjust the position along the pregnancy period. It could be interesting to use medical imaging techniques to provide accurate data, however this population is particularly vulnerable to medical radiation exposure. Thus, new measurements could be done to provide mathematical models that can predict accurate joint center locations.

Pregnancy, as an overload condition, raises the importance of knowing the joint forces provided by muscle action and external forces. This can be calculated using simulation analysis, where subject specific properties are provided to get an accurate model. With motion data and anthropometric measures it can be possible to understand how lower limb joints react to a pregnancy condition.

Biomechanical models can help in evaluating joint mechanical loads, kinetics and kinematics during gait and provide information about movement patterns. Walking is a common way of physical exercise during pregnancy and a daily physical activity during almost forty weeks, that can cause injuries by overuse of certain specific muscles, joint overload, asymmetries, uncomfortable or painful walking. The foot-floor contact model can be used to future research about the interaction between the foot and the floor, providing more data about the shear forces, or about different types of surfaces. Also, the use a multi-segment foot model would be more helpful in clinical context. From this, it can be developed of a specific shoe for pregnant women for the prevention of orthopedic discomfort, regarding all the morphological and hormonal changes.

Other aspects that need further research are the behavior of biomechanical loading according to the different level of physical activity of pregnant women as well as the presence of low back, girdle and foot pain.

6.3 References

Anderson, M., Benoit, D., Damsgaard, M., Ramsey, D. & Rasmussen, J. (2010). Do kinematic models reduce the effects of soft tissue artifacts in skin marker-based motion analysis? An in vivo study of knee kinematics. Journal of Biomechanics, 43, 268-273.

Branco, M., Santos-Rocha, R.A., Aguiar, L., Vieira, M.F. & Veloso, A.P. (2013). Kinematic Analysis of Gait in the Second and Third Trimesters of Pregnancy. Journal of Pregnancy, 1 - 9.

Butte, N., Ellis, K., Wong, W., Hopkinson, J. & Smith, E. (2003). Composition of gestational weight gain impacts maternal fat retention and infant birth weight. American Journal of Obstetrics and Gynecology, 189(5), 1423-1432.

Capello, A., Stagni, R., Fantozzi, S. & Leardini, A. (2005). Soft tissue artifact compensation in knee kinematics by double anatomical landmark calibration: performance of a novel method during selected motor tasks. IEEE Transactions on Biomedical Engineering, 52(6), 992-998.

Cereatti, A., Della Corce, U. & Capozzo, A. (2006). Reconstruction of skeletal movement using skin markers: comparative assessment of bone pose estimators. Journal of Neuroengenieering Rehabilitation, 23, 36-7.

Duprey, S.; Cheze, L. & Dumas, R., (2010). Influence of joint constraints on lower limb kinematics estimation from skin markers using global optimization. Journal of Biomechanics, 43, 2858-2862.

Forczek, W. & Staszkiewicz, R. (2012). Changes of kinematic gait parameters due to pregnancy. Acta of Bioengineering and Biomechanics, 14, 113-119.

Foti, T., Davids, J. & Bagley, A. (2000). A Biomechanical Analysis of Gait During Pregnancy. Journal of Bone Joint Surgery, 82-A(5), 625-632.

Fuller,J., Liu, L.J., Murphy, M.C. & Mann, R.W. (1997). A comparison of lower-extremity skeletal kinematics measured using skin- and pin-mounted markers. Human Movement Science 16(2-3), 219-242.

Gilchrist, L.A. & Winter, D.A. (1996). A two-part, viscoelastic foot model for use in gait simulations. Journal of Biomechanics, 29, 795-798.

Holden, J.P., Orsini, J.A., Siegel, K.L., Kepple, T.M., Gerber, L.H. & Stanhope, S.J. (1997). Surface movement errors in shank kinematics and knee kinetics during gait. Gait and Posture 5(3), 217-227.

Leardini, A., Chiari, L., Della Croce, U. & Capozzo, A. (2005). Human movement analysis using stereophotogrammetry. Part 3. Soft tissue artifact assessment and compensation. Gait and Posture, 21(2), 212-225.

Liu, Y., Shih, S.-M., Tian, S.-L., Zhong, Y.-J. & Li, L. (2009). Lower extremity joint torque predicted by using artificial neural network during vertical jump. Journal of Biomechanics, 42, 906-911.

Nazer, R.A., Rantalainen, T., Heinonen, A., Sievanen, H. & Mikkola, A. (2008). Flexible multibody simulation approach in the analysis of tibial strain during walking. Journal of Biomechanics, 41, 1036–1043.

Perry, J. (1992) Gait Analysis. Normal and Pathological Function. Thorofare, New Jersey: SLACK Inc.

Pitkin, R. (1976). Nutritional support in obstetrics and gynecology. Clin Obstet Gynecol, 19(3), 489 – 513.

Ren, L., Jones, R. K., & Howard, D. (2008). Whole body inverse dynamics over a complete gait cycle based only on measured kinematics. Journal of Biomechanics, 41, 2750-2759.

Appendix

7 Appendix

Appendix 1

Autorização de Compilação

Eu, ____,
na qualidade de co-autor do artigo ***Biomechanical Model for Kinetic and Kinematic Description of Gait During Second Trimester of Pregnancy to Study the Effects of Biomechanical Load on the Musculoskeletal System***, publicado no Journal of Mechanics in Medicine and Biology, 14, 1450004, declaro que autorizo a inclusão do mesmo na dissertação de doutoramento da candidata Liliana Sofia de Aguiar Pereira da Silva, intitulada "Biomechanical models of the lower limb and pelvis, for female human gait in regular and overload conditions related to pregnancy".

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Appendix 2