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ANGULAR KINEMATICS AND JOINT MOMENTS ANALYSIS IN LOWER LIMB AND PELVIS DURING GAIT IN SAGITTAL PLANE IN PREGNANT WOMEN

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This study compared sagittal plane lower limb range of motion (ROM) and joint moments of force (Mf) during gait in pregnant (second trimester) and non-pregnant women. Kinematic data were collected with an optoelectronic motion capture system (Qualysis, Ocqus 300) synchronized with two force platforms (Kistler AG, Winterthur, Switzerland) which collected ground reaction force values. The study revealed that the gait pattern in the second trimester of pregnancy is similar to the non-pregnant women pattern, in what concerns to the variables studied. Lower dorsiflexion and higher plantar flexion angles in the ankle joint in pregnant women, and higher values of hip flexion for the same group, were observed. With respect to the joint moments of force, there were higher knee flexor and hip extensor peak moments in pregnant women.

KEY WORDS: pregnancy, walk, biomechanical model, flexion and extension.

INTRODUCTION: Pregnancy causes morphological, physiological and hormonal changes that can influence performance in daily tasks and overall physical activity (Nicholls & Grieve, 1992). According to the Institute of Medicine (IOM) and National Research Council (NRC) (2009) recommendations, body mass for a woman with a normal body mass index (BMI) may increase, on average, between 11.5 and 16 kg and its distribution varies with the fetus growth, increasing the load on the trunk anterior area (Paisley & Mellion, 1988). The joint laxity increases and consequently, there is an increase on the joints amplitude (Calguneri, Bird & Wright, 1982). In the later stages (third trimester) of pregancy, the adoption of strategies such as increasing the width of base support and the reduction of obstruction caused by the other body segments, allow to minimize the increased weight and trunk girth effects. The different effects on the trunk movement can change locomotor patterns. As pregnancy progressed, the forward flexion and axial rotation motions of the trunk were restricted, during sitting and standing from a chair (Gilleard, Corsbie & Smith, 2002). Indeed, such changes can substantially modify the gait pattern, contributing to an overload on the musculoskeletal system, causing lower limbs, hip and lower back pain (Foti, Davids & Bagley, 2000). Wenhua et al. (2004) found that the pregnant gait kinematics general pattern performed at velocities between 0.17 to 1.72 m/s, is similar to non-pregnant women pattern. However, compared to higher velocity values, pregnant women revealed difficulties in the pelvis and thorax coordination. Foti et al. (2000) observed increases in the hip moments of force and power, in the frontal and sagittal planes, at the end of the second trimester of pregnancy. Huang et al. (2002) compared data from pregnant women at different gestational stages, and concluded that during pregnancy, the hip extensor moment of force increased, while the knee extensor moment and ankle plantar flexor moment decreased. The quantification or estimation of the mechanical load on internal biological structures during physical activity in sports, recreational activity, daily life and at the workplace is a prerequisite for understanding injury and overload mechanisms and for controlling that physical activity (Brüggemann, 2005). Biomechanical studies, although in small number, can contribute to improving knowledge of the anatomic changes that occur during pregnancy and influence motor coordination and musculoskeletal mechanical load, during daily or physical activities tasks like gait. This replicated study describe and compare kinematic and kinetics parameters (joint ROM and Mf), of the lower limb and pelvis, in the sagittal plane, during gait performed by pregnant and non-pregnant women.

METHODS: Subjects: The sample size consisted of five pregnant women (P) with 27 weeks (second trimester) mean gestational age, mean chronological age of 32 years and average BMI of 26.5 kg/m^2 and five non-pregnant women (NP) with chronological age average of 32.8 years and average BMI of 22.5 kg/m^2. All participants gave informed consent to participate voluntarily in the study.

Data collection and processing: Anthropometric data (weight and height) were collected from all the participants to calculate body segments masses and inertia moments. Motion capture was collected 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) which collected ground reaction force data.The subjects performed three non-consecutive minutes, walking at a comfortable speed, with a time break of thirty seconds between each trail. In each subject reflective markers were placed on anatomical points according to the defined marker setup protocol, based on Visual 3D C-Motion, Inc. Software, model construction recommendations. The biomechanical model consisted in seven body segments (pelvis, thighs, shanks and feet). The foot and leg segments were reconstructed from the medial and lateral anatomical markers, which allow defining the dimensions and the local coordinate system. The pelvis was built based on CODA model, which calculates the hip joint center, allowing the construction of the thigh segment. Body segment's dimensions were based on the relative distances between pairs of markers obtained by the motion capture system and the locations of the markers in the corresponding virtual model. In all segments were placed three tracking markers, which let the segment follow their coordinates, replicating the performed motion. All the markers coordinate data were interpolated using third degree polynomial and to reduce the noise the motion data was filtered, using a low pass Butterworth filter, with a cutoff frequency of 15 Hz. Each marker location was recognized in the static collection, working as a reference position and must be set in each frame. The segments are considered linked rigid bodies and joints have six degrees of freedom, making them independent form each other and allowing more movement. Joint degrees of freedom were manipulated through the Inverse Kinematics tool in order to achieve the correct 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, so that 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, all the joint rotations were allowed and restricted all the translations. Frame by frame, the recorded motion reproduces the model joint angles set in a configuration that best represents the experimental kinematics. The segments velocities and accelerations were obtained by derivation of the new position equations. Inverse dynamics method 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. To determine the joints net moments and forces, the equations of motion was iteratively solved, considering the equilibrium dynamic and boundary conditions. The joint torques calculation was performed using distal-proximal direction and during the left 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. To calculate ankle, knee and hip joint angles and moments of force, 3D biomechanical models were constructed, using Visual 3D C-Motion, Inc. Software.

Statistical analysis: The results are based on four cycles per subject. The average joint angles and moments curves in the sagittal plane were obtained for each participant as the mean curve for both groups, using Microsoft Office Excel 2007. Angular displacement was normalized to the gait cycle and joint moments of force data were normalized to the mass. The comparison of means between two groups was performed using the Student's t test. Mann-Whitney U test was used for variables that were not normally distributed. All the statistical tests were performed using the program PAWS Statistics 18.0.

RESULTS AND DISCUSSION: For the space-time parameters, there were no significant differences between both groups. Similar to the results found in literature, the stance phase was approximately 60% of the total gait cycle (Perry, 1992) for both groups. Pregnant women, registered lower values for cycle time (1.05±0.10 s) and the cycle length recorded also lower values (1.25±0.20 m) compared to non-pregnant group. Gait velocity was slightly higher in pregnant women (1.19±0.15 m/s).
Figure 1 shows the average curves normalized to the gait cycle of angles and moments of force of the three joints in sagittal plane, and differentiates the stance of the swing phase.

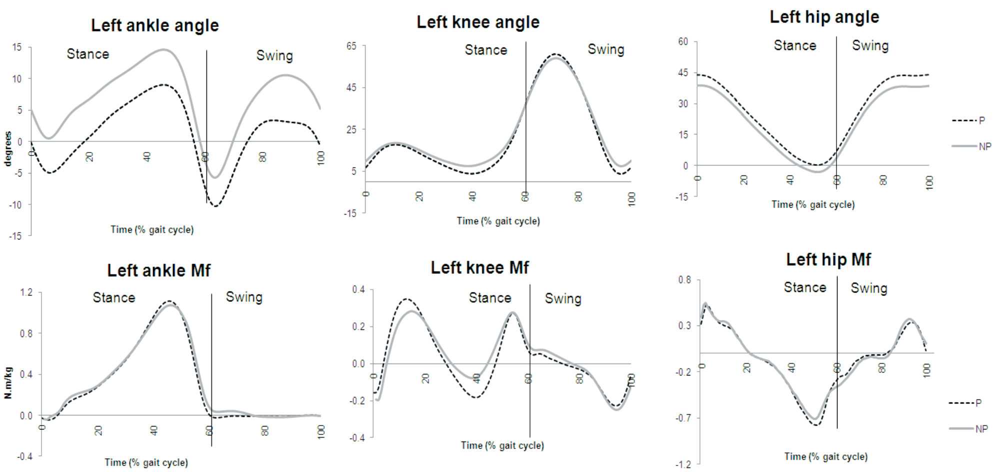

Figure 1: Average curves of left lower limb and pelvis joint angles and moments, in sagittal plane, normalized to gait cycle.

Thus, is possible to observe similar angles and joint moments curve patterns between the two groups (Wenhua et al., 2004) However, no significant differences were found for these variables. Relatively to the ankle joint, the plantar flexion peak moment occurred during the early stance phase, which was similar between groups. Then the dorsiflexor moment increased as the body moves in the forward direction (Perry, 1992). The dorsiflexion peak moment occurred at the dorsiflexion maximum angle in both groups, and it was approximately 1 Nm/kg. The maximum dorsiflexion angle was lower in pregnant women. Although, it was detected a higher plantar flexion angle, occurring already in the swing phase, immediately after the toe-off. Thus, the ankle joint range of motion was lower in pregnant women. Hip and knee joints revealed higher amplitudes in pregnant women. 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). In pregnant women, the maximum knee extension angle was lower and the maximum flexion angle and knee moments of force were higher than non-pregnant. The hip flexion peak moment of force was similar in both groups, however, slightly lower in pregnant women. The peak extensor moment of the same joint, was recorded on the final stance phase and the results indicate higher values for pregnant women (Foti et al., 2000).

CONCLUSION: The study shows a similar gait pattern between second trimester pregnant and non-pregnant women, in what concerns to the variables studied, despite sample size. Lower dorsiflexion and higher plantar flexion angles in the ankle joint in pregnant women, and higher values of hip flexion for the same group, were observed. With respect to the joint moments of force, there were higher knee flexor and hip extensor peak moments in pregnant women. Existing studies show clear gait differences in the third trimester; however these differences seem likely to emerge during the second trimester of pregnancy. It seems that second trimester pregnant women do not call for special precaution during walk. A limitation of this study may be related to the subject heterogeneity and sample size, which will be increased over the course of the project.

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