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ADJUSTMENT OF MASS MOMENT OF INERTIA OF LOWER LIMB ACCORDING TO GAIT PHASE

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Abstract: *The paper presents time dependent adjustment of total moment of inertia of the lower limb during gait movement. The kinematic data consisting in: Cartesian coordinates, relative and absolute angles of joints and segments, angular velocities and accelerations were determined from movement analysis of a healthy subject during gait. The mass characteristics were computed according to the literature and relies on the body weight and segmental lengths of the same subject. The total moment of inertia was then computed about the transversal axis of rotation that passes through the hip joint, on a three-segments model of the lower limb. The results show the process of total moment of inertia adjustment in relation to the gait phase, all in the purpose to facilitate the acceleration of the foot and the progression of the total body center of mass.*

Key words: *moment of inertia, angular velocity, center of mass, gait phase, lower limb model, motion analysis.*

1. INTRODUCTION

Assessing gait kinematics can be done by recording a subject walking either on the ground, or on a treadmill. The motion analysis instruments are capable of recording and storing data in both conditions, providing kinematical parameters (coordinates, joint angles, linear and angular velocities and accelerations) in high accuracy [1-4].

The standardization of the gait phases was done in the early '80s as a necessity of all researchers and clinicians to refer the same concept [5, 6]. In gait, the lower limbs are successively alternate the phases for producing the progression of the body center of gravity.

From engineering perspective stance and swing phases of the gait are significantly different, since in stance the limb has a contact to the ground while in swing the limb rotates its segments in the air.

The recording of gait can be done can be done using treadmill. In this way, the possibility of acquiring multiple consecutive strides without the necessity of having a large area of investigation arise. In addition, by controlling the velocity of the treadmill the velocity of walking can be imposed, leading to a more harmonic gait [7, 8].

Mathematical modeling of human lower or upper limb as three-mass models is proposed in many researches [9-11] an all of them relay on inertia matrix among the elasticity end damping coefficients. Other studies use rigid elements and rotation couples to write conventions from robotics in order to determine the law of motion of the final effector, or to establish various possible configurations of the system [12-14].

Calculation of inertial parameters: center of mass and moment of inertia in random postures of the human body can be approached by means of transformation matrix. The results obtained by Yifang et.al are in accordance to other methods of determination, like balance plate or trilinear pendulum [15].

Some authors work on estimating the whole mass moment of inertia of a human body during forward bending movement. Developing a model for computing moment of inertia of the whole body can be used in designing a robotic self-transfer facility for elderly and disabled people [16]

Taking into account that all main active movements in human body are rotations, estimating the mass moment of inertia of each segment can be done in respect to its own center of mass and in respect to the proximal or distal ends. This inertial characteristic of human body is a key factor in approaching movement analysis.

The purpose of this study is to present the adjustments of the total mass moment of inertia (mMoI) of the lower limb in accordance to the linear acceleration of the foot that characterize the swing phase and angular velocity of the hip that characterize the stance phase.

2. MATERIAL AND METHODS

2.1 Kinematic data acquisition

The kinematic data consist in spatial coordinates of the following anatomical landmarks: hip joint center, external and internal points that define the knee axis of flexion, external and internal points of malleolus that define the flexion of the ankle joint, the heel and the toe points. All landmarks were defined for both legs of a healthy feminine subject at the age of 30. Starting from the Cartesian coordinates of the landmarks, the relative and absolute angles between and of the segments were computed. Also, by numerical differentiation, the linear and angular parameters of velocity and acceleration were computed.

The measurements were conducted in the Motion Analysis Laboratory, within the Politehnica University of Timisoara on Zebris ultrasound system, using a sampling rate of 25 Hz.

2.2 Three segments rigid body model

In order to compute the mass moment of inertia and represent it according to linear and angular kinematic parameters that change during gait cycle, a simplified 3 segments model of the lower limb was made. The model is presented in the Figure 1. The considered segments of the model are: the thigh symbolized by index 1, the shank symbolized by index 2 and the foot, symbolized by index 3. Also, six axes of rotation are depicted in the model: three axes that transversally pass through the center of mass (COM) of each segment, and another three that transversally run through the joint centers.

In addition to the kinematic data, the mass, the COM position and the radius of gyration of each segment was estimated using the whole-body mass of the subject and the ratios estimated and presented by Dempster, Erdmann, Williams and Hall [17-20].

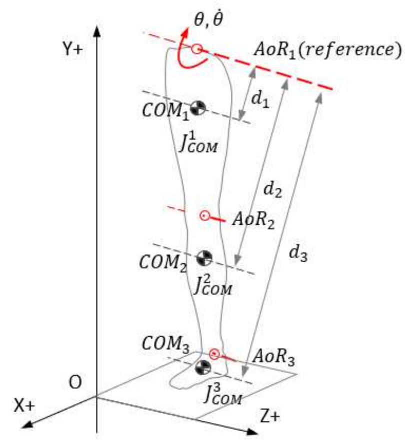

Fig. 1. Simplified 3-segments model of the lower limb

Starting from the 3D coordinated of the anatomical landmarks, as function of time during gate, the COM, length and radius of gyration were computed, considering also the mentioned ratios. The derivative parameters: angular velocity of the hip joint and the linear acceleration of the foot COM were calculated using the equations (1) and (2). The numerical differentiation was done for a constant Δ_t= 33 ms.

$$\dot{\theta}_{AoR1} = \frac{\theta_t - \theta_{t-1}}{\Delta t} \quad (1)$$

$$\ddot{x}_t^{COM} = \frac{x_{t-2}^{COM} - 2 \cdot x_{t-1}^{COM} + x_t^{COM}}{\Delta t^2} \quad (2)$$

The individual moment of inertia for each segment was computed using the parallel axis theorem (3), while the total moment of inertia of the lower limb using equation (4). Here m_i represents the own mas of the segment, ρ_i represents the radius of gyration defined by the proximal end of the segment, d_{COMi} the distance from the reference axis of rotation to each segmental COM. The total moment of inertia in respect to the reference axis of rotation was obtained by summing the individual segmental moments of inertia, in respect to the same axis.

$$J_{l_i}^{ref.} = m_i \cdot {\rho_i}^2 + m_i \cdot d_{COM_i}^2 \tag{3}$$

$$J_{total}^{ref.} = \sum_{i=1}^{3} J_{l_i}^{ref} \tag{4}$$

3. RESULTS AND DISCUSSIONS

The results of mMoI are presented for each segment in accordance to the horizontal acceleration (a_x) of the foot COM (Figures 2, 3 and 4) and to the vertical acceleration (a_y) of the foot COM COM (Figures 5, 6 and 7). All mMoI were computed in respect to AoR_1 axis. Both accelerations of the foot occur in the swing phase, between pre-swing subphase and terminal swing. The vertical acceleration manifests when limb folds and reduce its mMoI while the horizontal acceleration increases during the extension of the limb from the folded position.

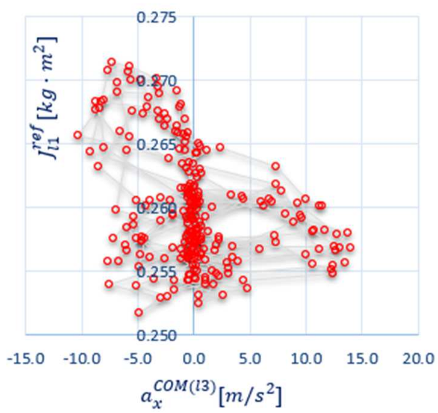

Fig. 2. Adjustment of mMoI of the segment 1, according to horizontal acceleration of the foot COM

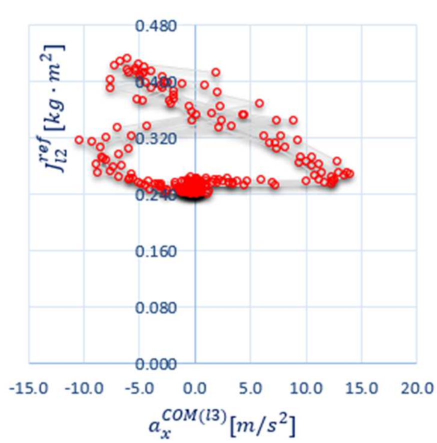

Fig. 3. Adjustment of mMoI of the segment 2, according to horizontal acceleration of the foot COM)

The maximum absolute values of horizontal accelerations are recorded for low values of segmental mMoI, the body accelerating more efficient a structure with lower mMoI.

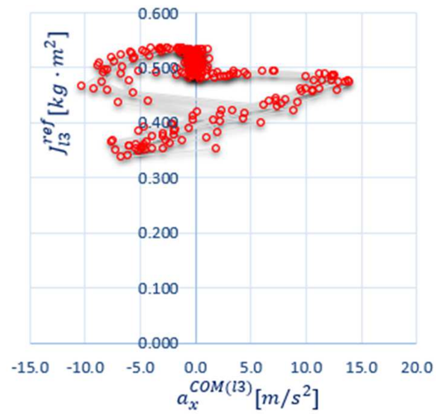

Fig. 4. Adjustment of mMoI of the segment 3, according to horizontal acceleration of the foot COM

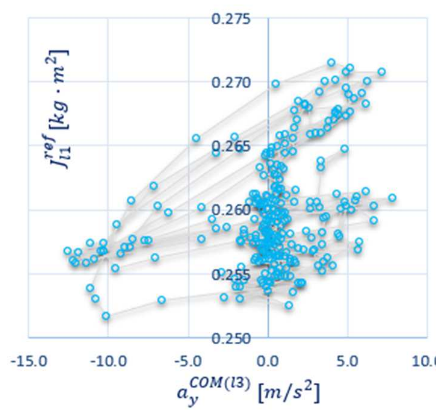

Fig. 5. Adjustment of mMoI of the segment 1, according to vertical acceleration of the foot COM

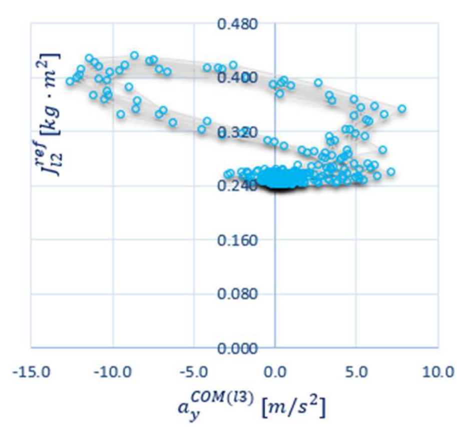

Fig. 6. Adjustment of mMoI of the segment 2, according to vertical acceleration of the foot COM

The figures 8 and 9 present the adjustment of total mMoI in respect to the angular parameters around AoR_1.

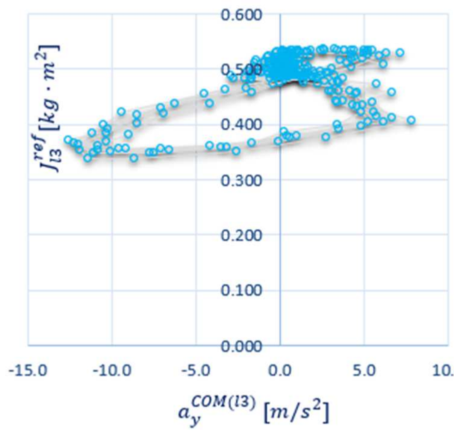

Fig. 7. Adjustment of mMoI of the segment 3, according to vertical acceleration of the foot COM

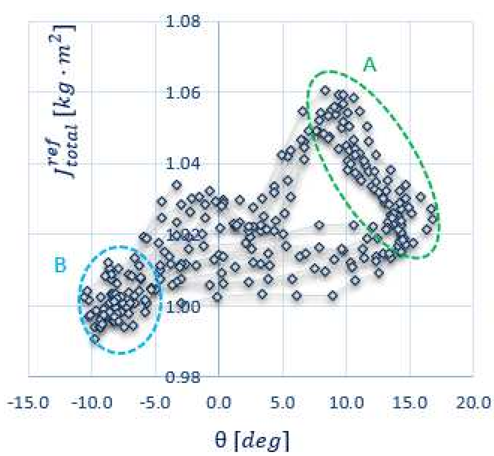

Fig. 8. Adjustment of total mMoI , according to the angular displacement around AoR_1

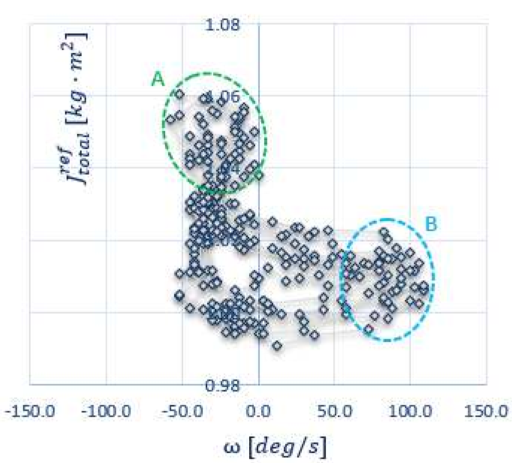

Fig. 9. Adjustment of total mMoI , according to the angular velocity around AoR_1

The detail A in the figure 8 and 9 underline the phase of full extension of the leg, just before the heel strike event, when the total moment of inertia is at its highest level. The high angles recorded in in hip joint are measured in respect to the vertical axis. The full folding of the leg is depicted by the detail B of the figures 8 and 9. Here, the mMoI is at its lowest value and represents the pre-swing and swing subphases of gait cycle.

The angular velocity around AoR_1 occur when the limb is fully folded and therefore the total mMoI minimum, this proving the minimal energy cost principle of the human locomotion [21-23].

4. CONCLUSION

The paper presents the adjustments of the segmental and total mass moment of inertia of the lower limb in accordance to the linear acceleration of the foot that characterize the swing phase and angular displacement and velocity of the hip that characterize the stance phase. The estimation of the mass moment of inertia was based on a 3-segments model of the lower limb, and the variation was explained according to gait phases. Some specific conclusions can be underlined:

- The vertical linear acceleration manifests when the limb folds and reduce its mMoI;
- The horizontal linear acceleration increases during the extension of the limb from the folded position;
- The maximum absolute values of horizontal accelerations are recorded for lower values of segmental mMoI.
- High total mMoI are associated with low values of angular velocity and vice versa.
- The correlation between mMoI adjustment and the kinematical parameters reveal a minimal energy cost strategy of the human walking.

5. REFERENCES

[1] Sun, J., Wang, Y., Li, J., Wan,W., Cheng, D., Zhang, H., *View-invariant gait recognition*

based on kinect skeleton feature, Multimed Tools Appl, 77(295) (2018).

[2] Luca, R., Bejinariu, S.I. *Classification Method for Human Locomotion.* Proceedings of the 10th International Conference on Electronics, Computers and Artificial Intelligence, ECAI 2018, Iasi, Romania, (2019)

[3] A.J.J. Smith, E.D. Lemaire, *Temporal-spatial gait parameter models of very slow walking*, Gait&Posture, 61, 125-129 (2018).

[4] Stoia D.I., Vigaru C., Nicoara A., Herisanu N. (2021) *Angular Momentum About the Total Body Center of Mass Computed at Different Speeds.* In: Herisanu N., Marinca V. (eds) Acoustics and Vibration of Mechanical Structures—AVMS 2019. Springer Proceedings in Physics, vol 251. Springer.

[5] Soudan, K. (1982). *Standardization of Gait Kinematic Data Using a Gait Symmetry Index and Fourier Analysis*. In: Huiskes, R., van Campen, D.H., de Wijn, J.R. (eds) Biomechanics: Principles and Applications. Developments in Biomechanics, vol 1. Springer, Dordrecht.

[6] Jacquelin Perry, Judith Burnfield. *Gait Analysis: Normal and Pathological Function*, Slack Incorporated; Second edition, 2010

[7] Tarnita, D., Georgescu, M., Tarnita, D.N., *Applications of Nonlinear Dynamics to Gait Analysis on Plane& Inclined Treadmill,* New Trends in Medical and Service Robots, Springer Publishing House, 39, 59-73, (2016).

[8] Van Caekenberghe I, Segers V, Aerts P, Willems P, De Clercq D. 2013 *Joint kinematics and kinetics of overground accelerated running versus running on an accelerated treadmill.* J R Soc Interface 10:20130222

[9] Simona Gabriela Jurco, Eugen Jurco, Gheorghe Tomoaia, Mariana Arghir. *Study of human body mobility through mechanical modeling,* ACTA TECHNICA NAPOCENSIS Series: Applied Mathematics, Mechanics, and Engineering Vol. 60, Issue I, 2017

[10] Aurora Felicia CRISTEA, *Theoretical interpretations concerning the stability of the human hand-arm system at low frequencies, system comparative with a mechanical system.* ACTA TECHNICA NAPOCENSIS Series: Applied Mathematics, Mechanics, and Engineering Vol. 62, Issue III, 2019

[11] Andreea Stoia, Dan Ioan Stoia, Nicolae Herisanu. *A biodynamic multibody system. OHAM solution.* AIP Conference Proceedings 2425, 310007, 2022

[12] Dragulescu D, Perdereau V, Drouin M, Ungureanu L, Menyhardt K., *3D active workspace of human hand anatomical model,* Biomed Eng. Online, 6:15, (2007)

[13] Marius Georgescu, Daniela Tarnita, Ilie Dumitru, Alin Petcu, Razvan Vaduva, Dan Marghitu. *Analysis of human hip movement using nonlinear timeseries analysis methods*, ACTA TECHNICA NAPOCENSIS , Series: Applied Mathematics, Mechanics, and Engineering Vol. 64, Issue Special II, 2021

[14] Tarnita, D., Catana, M., Tarnita, D.N.; *Nonlinear analysis of normal human gait for different activities with application to bipedal locomotion,* Rom. J. Techn. Sci. – Appl. Mechanics., 58(1-2), 177–192 (2013).

[15] Yifang Fan, Zhiyu Li, Changsheng, Calculation of Center of Mass and Inertia *Moment of Human Body in Motion by Means of Transformation Matrix,* Seventh International Symposium on Instrumentation and Control Technology: Measurement Theory and Systems and Aeronautical Equipment, Proceedings of SPIE Vol. 7128, 71280I, (2008)

[16] Ramachandran, Hari & Vasudevan, Devanandh & Brahma, Aditya & Pugazhenthi, S. (2016). *Estimation of mass moment of inertia of human body, when bending forward, for the design of a self-transfer robotic facility*. Journal of Engineering Science and Technology. 11. 166-176.

[17] Dempster, W. T. (Wilfrid Taylor), *Space requirements of the seated operator: geometrical, kinematic, and mechanical aspects of the body, with special reference to the limbs,* (1955)

[18] Erdmann, W.S. (1999). *Geometry and inertia of the human body-review of research.* Acta of Bioengineering and Biomechanics, 1(1), 23-35.

[19] Williams, R.L. (2014). *Engineering biomechanics of human motion.* Supplement notes, Mechanical/Biomedical Engineering, Ohio University.

[20] Hall, S. J. (2012). *Basic biomechanics*. (6th ed.), McGraw-Hill, Boston, MA.

[21] Holt KG, Hamill J, Andres RO. *Predicting the minimal energy costs of human walking*. Med Sci Sports Exerc. 1991 23(4):491-8.

[22] Faraji, S., Wu, A.R. & Ijspeert, A.J. *A simple model of mechanical effects to estimate metabolic cost of human walking.* Sci Rep 8, 10998 (2018).

[23] Fariba Hossein Abadia, Tajul Ariffin Muhamada, Norlena Salamuddin. *Energy Expenditure through Walking: Meta Analysis on Gender and Age*. Procedia Social and Behavioral Sciences 7(C) (2010), 512–521.

MODIFICAREA MOMENTULUI MASIC DE INERȚIE AL MEMBRULUI INFERIOR UMAN ÎN CONFORMITATE CU FAZA CICLULUI DE MERS

Rezumat: Lucrarea prezintă modificarea în timpul mersului a momentului de inerție total al membrului inferior uman. Abordarea pornește de la determinarea parametrilor cinematici: coordonate carteziene, unghiuri relative și absolute ale articulațiilor și segmentelor anatomice, vitezele și accelerațiile unghiulare, toți ca funcții de timp determinați pentru un subiect sănătos, aflat în mers. Caracteristicile masice au fost calculate pe baza literaturii de specialitate și iau în considerare masa totală și lungimile segmentelor aceluiași subiect. Valoarea totală a momentului masic de inerție a membrului inferior a fost calculată față de axa transversală perpendiculară pe planul sagital și care trece prin centrul articulației coxofemurale, utilizând un model din 3 segmente rigide. Rezultatele prezintă modul în care se modifică momentele de inerție total și segmental în timpul mersului, în scopul eficientizării accelerației labei piciorului și al realizării progresiei centrului de greutate al corpului.

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