Correlated and interconnected analyses for human walking and standing biomechanical behavior

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Abstract: In this paper I presented a series of analyses performed in complex investigation structures aiming each time at the establishment of an advanced non-invasive and objective methodology suitable for every type of locomotion or stability malfunction. At the same time we intended the assessment of these malfunctions in connection with other physiological parameters, which do not present deflections from the normal status. The objective clinical examination of the gait or standing position represents an important study of clinical semiology, being necessary in acknowledging the pathology of certain afflictions and symptoms. It must be performed on a flat ground, especially climbing up and down the stairs, in normal parameters of the recording conditions (temperature, environment, humidity, atmospheric pressure) but also at different moments of the day (morning, noon, evening). The methodology and the investigations procedures are presented in the second part of the paper and in the final part the results and the conclusions of this work are presented.

Key-Words: stability behavior, gait, human body, biomechanics.

1 Introduction

Being the result of a prolonged cycle of phylogenetic transformations and also of a complicated ontogenesis, the human body managed to become a complex organism, having most various forms of motion, and more and more developed adaptabilities and compatibilities. A multitude of morpho—functional factors that are the fundament of human locomotion should be analytically studied due to the extremely practical reason of emphasizing each and every factor. Human body, as an entity of its subsystems should be regarded as a whole and not as a manifestation of modular structures that are independently acting, so that the morpho-functional factors generating the body motions are integrated in this system by means of the analysis using reintegrating synthesis of the environment interaction feed-back reactions. A series of external factors acting upon the human organism to a greater or smaller extent, along a variable or constant period represent the source of functional elements interacting with the human body. [1]

At the same time, the human body motion itself is the one influencing its evolution, changing even its structure, making it fit to achieve more and more complicated movements. We can thus state that the human body structure consists of functional structures, created by dedicated function, with the purpose of creating functions.

In the most simplified form, the way of motion actions in the functional structure of the human body (organs and tissues) appears to be connected to the occurrence and manifestation of some couples of
forces of the type: action-reaction forces, external forces, internal forces.

2 Theoretical aspects

From the point of view of structural analyses performed upon the human body in order to achieve its static or dynamic normal or abnormal behavior assessment, the most important elements that were taken into account were: the hip, the thigh, the knee, the calf, the ankle and the foot acting during various positions or/and motions like an open or closed kinematical chain. The closed kinematical chain formed by the locomotor system components acts for maintaining and supporting the body in orthostatic position, for propulsion motions or for damping motions during a fall (on the feet). The inferior limb acts like an open kinematical chain in the different variants of adduction and abduction motions, external and internal rotation, kicking, pushing and not least in accomplishing the gait cycle, all these actions being included in a system of coordinates. (fig.2)

The stability stance as well as the integral balance around the equilibrium position are determined by the health level of the entire human body and may constitute clear informational sources for the human behavior evaluation in any situation. The small deviations of the human body posture around the vertical direction determine the occurrence of a torsion moment, which acts upon the entire structure and may unbalance the human body or may create a vibration state. However, this process of corrective torque generation is not fully understood and controversy remains regarding the organization of sensory and motor systems contributing to the postural stability of the entire human body. Balanced state of postural sway is controlled by central nervous system, and the upright stance cannot be sustained without this control. It is widely accepted that the corrective torque is generated through the action of feedback control system; the input sources include visual, proprioceptive and vestibular system [2].

Figure 3 shows the block diagram of the postural sway feedback control and also a simplified pelvic structural model during static upright stance. A, B are the masses of legs, C is the mass of pelvis and D is the mass of upper trunk. Because the lumbar-sacral always sways in inverse direction of the ankle joint with the same value of θ, the upper trunk is kept perpendicular to the horizontal (the human body symbolic represented is for a subject with leg impairment). Location of COM remains fixed as long as the body does NOT change shape.

In order to locate the center of mass it is necessary to establish some main principles:
- its precise location depending on individual's anatomical structure;
- habitual standing posture;
- current position;
- external support;
- location in human body;
- variations with body build, posture, age, and gender
- infant > child > adult (in % of body height from the floor);
- generally accepted that it is located at ~57% of standing height in males, and ~ 55% of standing height in females;

The maintenance of equilibrium in standing position is one of the most important activities for two main reasons: firstly, the center of mass must be located in the support area; secondly, for a major period of the standing action, the body is supported first by two legs and after a short time by a single limb with the center of mass inside the base of support but with the tendency of going outside it.

In elder people especially, up to 70% falls occur during standing and of course, locomotion action or stepping [3]
By the static stability margin is meant a distance of the GCOM from the edge of the support polygon, measured along a current vector of motion of the gravity center, where:

$$x_{GCOM} = \frac{\sum_{i=1}^{n} M_{xi}}{\sum_{i=1}^{n} m_i} = \frac{\sum_{i=1}^{n} m_i x_{ci}}{\sum_{i=1}^{n} m_i}$$  \hspace{1cm} (1)

$$y_{GCOM} = \frac{\sum_{i=1}^{n} M_{yi}}{\sum_{i=1}^{n} m_i} = \frac{\sum_{i=1}^{n} m_i y_{ci}}{\sum_{i=1}^{n} m_i}$$  \hspace{1cm} (2)

and $m_i$ is mass of the $i$-th body, whereas $x_{ci}$, $y_{ci}$ denotes location of the center of mass of the $i$-th body.

With respect to analyses upon the human body static stability, human gait is a motor ability by means of which the displacements are usually performed using the alternative and constant motions mechanisms of the two inferior limbs, as support and as propellant.

As a follow, the motions of the human body are performed by a series of muscles groups, which form a harmonious assembly of muscular-kinematic chains, created according to the motion particularities under the control of the cerebral cortex. The motions performed by the human body have spatial directional characteristics of the motion and of the trajectory length travelled by the body or the body segments. They may be continuous, interrupted or combined according to a certain succession. The ratio between the spatial and temporal characteristics that establish the velocity and also acceleration parameters of the motion and all these characteristics as a whole, represent the kinematic particularities of the motion: where, how much and how is the body and its segments moving, along which trajectories described by the body segments and which controlled way is the complex motion of the human body performed.

All these aspects are important to be known when we analyze the disfunctionalities of the human body locomotor system.

The motions that can be performed by the human body are translations and rotations, complex and in 3D. The locomotion motions may be also cyclic or non cyclic (when the disfunctionalities occur).

The CYCLIC motions are those locomotor motions in which each part of the body returns to the initial position and then a new cycle begins, similar to the previous one. By the concept of cyclic motion we understand the totality of the motions performed by the human body and its segments, starting from a random initial position, considered as a starting point, until it reaches the next identical position. The cyclic locomotion is the result of repeating these uniform and similar cycles, also called walking or running motion units.

The NON CYCLIC motions are those locomotor motions during which there is no successive repetition of some motions cycles, such as jumps or throws, during which the body goes from an initial position to a final one and then the motion stops. This way of manifestation could be also observed in the cyclic motion with locomotor disfunctionality, those that require rehabilitation procedures in order to become cyclic motions again.

In dynamic walking, at each foot step, the system may suffer impacts and incurs on additional accelerations that influence the forward velocity. For this reason, it was necessary to impose a set of conditions (continuity conditions) on the leg velocities so that the feet are placed on the ground carefully, without producing impact. We denote the moment of exchange of support as time $t_i$, and by $t_i^+$ and $t_i^−$ the times before and after the impact occurs, respectively. In normal and cyclic movement of gait process the mechanism for exchange-of-support, the angular velocities, before and after, must be identical, that is:

$$\dot{\theta}_i(t_1) = \dot{\theta}_i(t_2) \quad 1 \leq i \leq 3$$  \hspace{1cm} (3)

The kinematic relations have been used and the differential problem solved to obtain the Cartesian velocities immediately before and after contact. The foregoing derivation determined conditions for smooth exchange-of-support. In accordance, the equation of the tip of the swing leg along the $x$-axis is computed by summing a linear function with a sinusoidal function:

$$x_{zi}(t) = 2V_f \left[ t - \frac{1}{2\pi f} \sin(2\pi ft) \right]$$  \hspace{1cm} (4)

where $f$ is the step frequency (number of steps per unit of time).

The vertical motion, that allows the foot to be lifted off the ground, is implemented using:

$$y_{zi}(t) = \frac{F_c}{2} \left[ 1 - \cos(2\pi ft) \right]$$  \hspace{1cm} (5)
The trajectory generator is responsible for producing a motion that synchronizes and coordinates the leg behaviour. In this perspective, we make sure that the swing limb arrives at the contact point when the upper body is properly centred with respect to the two lower limbs.

The main equations describing the inferior locomotor system motion and determine the reaction forces in the ankle joint are based on the **inverse dynamics method**, in which the kinematical quantities (ground reaction forces and anthropometrical dimensions) are considered as input data for the biomechanical system approach. Besides this calculus method assumes the following working conditions:
- human body is divided in kinematical chains,
- kinematical chains are divided in segments,
- segments are considered rigid bodies,
- air friction forces and joint friction are considered zero.

We can obtain 3 equations systems and in following equations it is shown like exemple.

\[
\sum F_{si} = m_p a_s \Rightarrow R_{s0} + R_{s_i} - F_f = m_p a_s
\]
\[
\sum F_{yi} = m_p a_y \Rightarrow R_{y0} + R_{y_i} - G = m_p a_y
\]
\[
\sum M = l_p \theta_p \Rightarrow M_g + R_{d_2} + R_{d_4} - F_f d_1 - G d_3 = -ma_y a_x - ma_y a_x + l_p \theta_p
\]

After calculation of these equations using anthropometrical and kinematics data for all 4 stages of equilibrium we can obtain the numerical values of reaction forces and also the values of moments into ankle articulation. [5]

**Table 1**

<table>
<thead>
<tr>
<th>Support Periode [%]</th>
<th>M_i [kg]</th>
<th>R_{s1} [N]</th>
<th>R_{s1} [N]</th>
<th>M_g [Nm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>70</td>
<td>-111,13</td>
<td>-227,8</td>
<td>-0.6</td>
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<td>25</td>
<td></td>
<td>20,79</td>
<td>-400</td>
<td>-28,48</td>
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<tr>
<td>45</td>
<td></td>
<td>83,5</td>
<td>-529,1</td>
<td>-69,59</td>
</tr>
<tr>
<td>70</td>
<td></td>
<td>125,35</td>
<td>-675,6</td>
<td>-121.98</td>
</tr>
</tbody>
</table>

**Energy developed into human body locomotor muscles**

The amount of work done by muscles at a joint can be computed from the rate of work at the joint through time-integration. For example, the work done by the muscles at the ankle is

\[
W_{AK} = \int_{t_o}^{t_f} P(M)_{AK} \, dt = \int_{t_o}^{t_f} T_{AK} \cdot (\omega_{FT} - \omega_{SI}) \, dt \quad (7)
\]

where \(t_o\) = the initial time of the movement, \(t_f\) = the final time, \(W_{AK}\) = the amount of work done by the muscles at the ankle. Equation (7) is in fact equal to the area under the muscle power-time curve. Throughout a movement, the muscle power can show both positive and negative phases.

Computing the amount of work done in each phase is very important in the sense that it really gives the amount of efforts put by the muscles either concentrically or eccentrically. Integrating the muscle power-time curve for the entire phase is somewhat meaningless since the positive and negative phases cancel each other and one can only get the net work done.

The muscles consume energy regardless of the mode of contraction (concentric vs. eccentric) and it is only reasonable to add all the positive and negative works done.

This can be generalized as

\[
W_{\text{JOINT}}^{(+/-)} = \sum_i \left| \int_{t_o(i)}^{t_f(i)} P(M)_{\text{JOINT}} \, dt \right| \quad (8)
\]

where \(i = \) phase number, \(t_o(i) = \) the initial time of the \(i\)-th phase, \(t_f(i) = \) the final time, and \(W_{\text{JOINT}}^{(+/-)} = \) the absolute amount of work done by muscles regardless of the contraction mode.

Equation (8) is commonly used to compute the amount of internal work done by muscles and the efficiency of the human machine.

It is extremely important that in the procedures of conception and modelling of the rehabilitation system we perform the assessment of the energy consumption at the level of inferior members.

In the locomotor system semiotics, the gait analysis has a special value, the occurred changes often having patho-gnomic character and being limited by certain minimum conditions due to muscular deficiencies.

In order to counteract the deficiencies manifestation the displacement mechanisms are changing both as segments and as a whole, using at maximum efficiency the remaining muscular forces and in some situations involving even the passive stabilization mechanisms (by tensing some supporting ligaments).
3. Experimental setup for investigation

In order to start the investigation we analyzed and realized a data acquisition structure based on an assembly of measuring human physiological parameters controlled by a computer unit.

The main measuring element is the Kistler force plate, which allows the values acquisition for the forces and moments developed by the human body, along the three directions (X, Y and Z), during an established period of time according to the experiment requirements.

The analysis performed upon the subjects started by establishing an investigation protocol, which aimed at a large range of measuring the bipedal stability (big support base with different polygons, small support base trapeze shaped, open eyes and arms along the body, in three moments of the day – morning, afternoon and evening).

The corresponding soft for the values acquisition is Bioware, which allows the recording of the forces and moments values, measured along the three directions by help of some piezoelectric sensors of the force plate.

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We also aimed at the fact that the bipedal position of each subject is centered on the plate, with no high heels shoes, arms relaxed along the body, open eyes and the eyes oriented straight ahead.

In first stage of the experiments we established and kept the parameters of the laboratory environment.

Temperature into laboratory was 220°C, air humidity 80% and atmospheric pressure 755 mmHg.[4]

In the second stage we measured the physiological parameters of the human subjects (weight, height, age, pulse, blood pressure) in relaxed stance, without any general health problems and with a good metabolism (example: blood pressure 155/82 mmHg, pulse 78-88, face temperature 36,7°C, height 170-185 cm, weight from 50-95 kg).

All these parameters are necessary to establish a common modeling base to measure and to evaluate the human body stability behavior.

By this proposed investigation structure we establish that the human subjects are analyzed and compared to the corresponding virtual models of the stability measurements simulations in order to correlate all the influence factors. [4]

Thus, in fig.5 we present the data acquisition concerning the stability of a human subject, female, height 1.7m, age 53, no general health problems but with a knee impairment, weight 80kg, for which we analyzed the stability area and the force evolution along Oz in three moments of the day (morning, afternoon and evening) without any source of additional effort induced to the organism.

As we can notice from the diagrams analysis in fig.5, the evolution of the stability area in this case presents a compact and symmetrical surface for the
first time in the morning, a smaller and more concentrated area for the afternoon and a substantial change of balance – slightly shifted along Ox for the evening recording.

This manifestation can be found in all the analyses performed on the selected subjects allowing a unitary evaluation of the stability area. As far as the recorded force on Oz is concerned, these results are presented in fig.6 (up – morning, middle – afternoon, down – evening). The recording time was each time the same – 16 sec and the data set was stored in the measurements database used for evaluation.

In the case of force evolution analysis we observe the same type of manifestation for the recording performed in the evening, emphasized by an increase of the variation limits of the force along Oz, but also by a higher frequency of their occurrence, values that indicate an increased instability of the human body and a fatigue state at the inferior limbs level. In this situation and correlating with the age and the influence of the poor sensorial system we may confirm that the installation of the fatigue state as a follow of a normal daily activity takes place in the second part of the day determining a motor activity deficit and the diminishing of the orientation perception.

In the case of a male subject age 34, no health problems, not wearing glasses, weight 97kg and height 1,7m, the evolution of stability area in the three moments indicates a more compact and symmetrical shape around the theoretical equilibrium position as we can observe in fig.5. and
the forces variation diagram is changing towards the diminishing of the oscillations number for the recordings situation related to those in the morning, fig.8. This fact establishes a more equilibrated behavior for which the fatigue due to daily activity does not influence the motor capacity and also does not reduce the resistance to effort.

To correlate the analyses on biomechanics behavior of human body in fig. 9 we present a synthetic model of key factors that contribute to quasi-static and dynamic postural control. This model includes feed-forward and feedback control functions. The anticipated postural disturbances, involuntarily known by the subject arise from the forces generated during the performance of voluntary motion, or in anticipation of destabilizing external forces. This type of postural adjustment, referred to as anticipatory, involves estimation of the magnitude and direction of postural disturbance and initiation of a motor program by the central nervous system. The motor program is a set of commands that is selected based on an internal or previous model of the task to be performed. Once initiated, the motor program is executed in an open loop manner, but with the potentially processing delays inherent in feedback control. Ideally, the motor program activates the musculature that produces a set of appropriately scaled and timed pre-emptive muscle forces and joint moments that precede and deny the anticipated postural disturbance.

During feedback control, a postural disturbance, the origin of which may be either internal or external, causes a change in body posture or movements. If the change in kinematics exceeds some threshold value to which the central nervous system has assigned importance, a corrective postural response will be generated. The changes in the kinematics process stimulate visual, vestibular, and somatic-sensory system, sensors that transmit the information to the central nervous system. Information processing by the sensorial system involves comparisons for detecting the state of the body into a desired state. In an ideal configuration, if the difference between the detected and desired states, that is, an error signal, is of sufficient magnitude, a corrective postural response consisting of muscle forces and joint moments will be executed to obtain the final response. The muscle forces and joint moments affect body kinematics, generating a new set of sensor information signals and another loop of the feedback process is initiated in this model. The model of the walking human body model is presented in figure 10 and it was used to build gait simulation with LifeMOD software. Starting from a pre-defined skeleton module and considering the anthropometrical database NASA-STD-3000 we build the shape of the human inferior locomotion system with direct contact to the walking support.
For modeling human gait we considered a series of data connected to motion, trajectory, velocity or acceleration but at the same time we introduced the boundary values of the gait type (normal, malfunction of the right or left foot, jumps or steps, slips or sliding on plane surfaces etc.). The modeling stages aim at introducing data both for the normal mode and for the one used to model a certain gait type in order to simultaneously visualize these differences.

4 Conclusion

From these recordings and according to the initial conditions and the demands of the researches we can conclude: that the most important force values are the components on the direction Oz because they can establish the amplitude of the balance (moments) in other two directions Ox and Oy. Also the changes in foot position have been found to affect measurements of standing balance, force and stability surface and in normal conditions the size of the support is a primary determiner of stability. Other influences were the light stimulus on the visual system because they are the most important stimulus inducing the instability that will be bigger in the open and fixed oriented eyes position than free gaze even if the optical stimulus was the same. This situation is due of the unknown visual external stimulus reactions and concentration on the automatic activities.

Following the analysis of different modeling alternatives created by LifeMOD software, in a normal way and with imposed disabilities we discovered a series of aspects correlated with the results obtained during the investigations upon a human subject in the same conditions. These investigations and recordings, obtained by help of a force and moments acquisition system along three directions (Kistler force plate) confirm the shape of the contact force (between feet and displacement surface) variation.

Thus, for a quick response in the analysis of the gait type and forces developed in the subject locomotion system, the created model can estimate and correlate data at different recording times and respectively for different anthropometrical dimensions or mobility restrictions.

From the recordings performed using the experimental device, the most important is the evolution of the contact force between the foot and support, considering no sliding between them and which emphasize the precise moments when this contact takes place. In these simulations, muscles generated about half (57±74%) of the knee extension acceleration during the extension phase and the other half was provided by velocity-related forces that arose from the rotational motions of the limb segments. Muscles generated nearly all of the knee flexion acceleration during the braking phase. Muscles on the stance limb, particularly the hip abductors, extensors, and flexors, had a major influence on motions of the swing-limb knee in the simulations. These muscles, in combination with their induced ground reaction forces, accelerated the pelvis, simultaneously inducing reaction forces at the swing-limb hip that accelerated the thigh and knee.

The modeling and simulation structure, briefly presented for this case represents the subject of a more extended research, which allows the developing of an investigation-assessment-rehabilitation protocol for the hip implant patients or for the patients with different walking impairments. By the future researches we are going to analyze the influences of the visual and audio stimuli upon the human body equilibrium in order to obtain a correlated evaluation of the human subjects stability, subjects involved in technological activities.

5 Acknowledgment

These researches are part of the Grant PNII-IDEI 722 and 744 with CNCSIS Romania and we’ve developed the investigations with apparatus from Research Project “CAPACITATI” in Mechatronic Researches Department from University Transilvania of Brașov.

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