A model to explain human trunk mechanics in walking

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ABSTRACT

Human trunk morphology was inherited from our philogenetical quadruped ancestors and adapted to the exigencies of bipedal locomotion. While the interaction between trunk lateral bending and extremities is the main engine by reptilians and the co-work between trunk frontal bending and legs this by quadruped mammals, the adaptations towards human bipeds have conduced to the use of the trunk torsion as the main drive mechanism for locomotion [1]. Since Braune & Fischer [2], human engineering evaluation models (anthropomorphic models) have been used to model human functional morphology. They allow computing mass, body segments' centers of gravity and moments of inertia of the different body segments, by approximating these as uniform geometric shapes. To date, kinematical studies give evidence of a systematic use of the trunk during walking [1, 3]. This is shown in the changes of trunk amplitudes, frequencies and phases as a function of gait's velocity. These changes, however, are not well understood and vary among human beings. A recently published study [4] on 106 volunteers (50 f, 56 m) showed that the variations of the 6 DOF kinematical parameters of trunk motion during walking are not simply correlated with anthropometry. The missing link between all those observations may be the individual dynamics of the trunk, determined by elastic and damping properties of muscles and soft-tissues. We propose a 14 DOF function- morphological model of the trunk based on Hanavan's

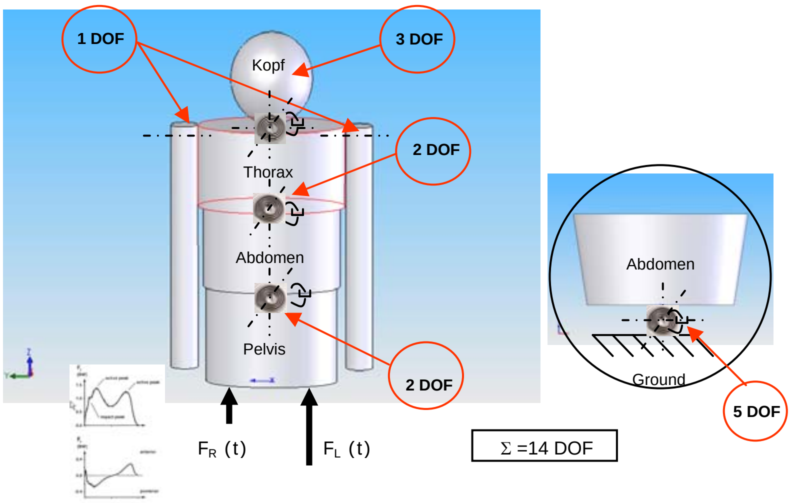

Fig.1 Anthropo-functional model of the body stem based on Hanavan's model (14 DOF).

model [5] (fig.1) to make accessible that information via experiments. The model was constructed as follows: The head was modelled as an elliptical ellipsoid, the trunk was divided in three elliptical cylinders, which represent thorax, abdomen and pelvis. The arms were modelled as simple cylinders. In all rotational joints, for each rotational degree of freedom, torsion spring and damper were mounted. Abdominal longitudinal and transverse translations are free. The legs were replaced with artificial vertical and anterior-posterior ground reaction forces applied to the hip joints. The vertical components were modelled using Alexander's equations [6]. Density values are uniform across each solid and equal to 1.075 gr/cm^3. Representation of a subject requires 15 measurements. 6 DOF kinematical data are used as matching parameters for the model. Results of the simulations present that visco-elastic parameters, which are obtained by the simulations, are a solid base to explain the inter-individual differencies in amplitudes, phases (see fig. 2) and frequencies of the different body segments in normal walking. They are also able to explain inter and intra-individual kinematical variations.

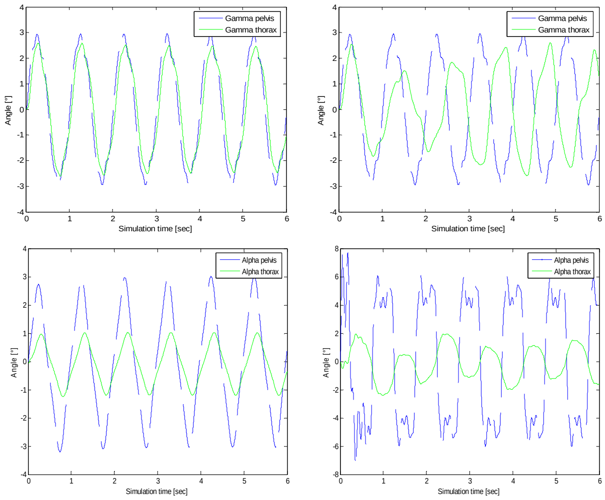

Fig.2 Change from in-phase to out-of-phase in the relative motions between pelvis and thorax. Upper: transversal plane, bottom: frontal plane. Changes were produced only by changing visco-elastic parameters in the joints and remaining ground reaction forces constant.

Spring stiffness is the main cause in chainging absolute amplitudes, while damper constants produce the variation of the phases between absolute segmental rotational motions. First results are very encouraging towards applications in fields of medicine and robotics. At present we are testing the feasibility of using motions analyses together with our model for prevention and diagnostics of trunk diseases.

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