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SIMULATION OF DAILY LIFE ACTIVITIES IN BIOMECHANICS

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Abstract: *Finite Element Analysis (FEA), the widely accepted virtual testing process, involves creation of a virtual geometry model that is converted into a Finite Element Model (FEM) and complemented by material and load models. The creation of geometry model is presently supported by various NURB packages while the material model can be translated from CT scan via techniques of power law or micro-mechanics. The load model remains at discretion of the researcher and his resources. This leads to a situation when the load model is limited to simulate over simplified daily life activity (DLA) such as standing position and any other activity where the forces can be computed from static equations without consideration of the muscular activity. This creates a drawback for correct simulation of the test 'in silico' that should be carried under the maximum load condition thus considering inertial forces and additional muscular load. This paper will provide brief overview of the present way to evaluate the load for required DLA and the final results obtained when modelling an ordinary walk. The results (Sant et al., 2012), based on validated model (Arjmand et al., 2006), will be compared with results of AnyBody load model applied to the lumbar spine.*

Keywords: Finite elements, Lumbar spine, Inverse dynamics, Muscular activity.

1. Introduction

The analysis of DLA effect on the bone cannot be measured directly thus 'in silico' testing is widely accepted. The results of these tests provide a valuable feedback, which can offer a better view on the outcome of the planed surgery or in case of a designer it provides a response, which initiates corrections of the design if necessary. The walk is a natural movement for people of all generations therefore it was selected as a DLA. Another reason why walk should be selected is the fact of that more than 33 % of population suffers at least once in their life a low back pain associated with this activity. It is the lumbar spine that undergoes around 3 millions of cycles per year for an average person and even more for active people or sportsmen. To model the load, which the lumbar spine must support during walking is a very complex task since the lumbar spine undergoes during each step a combine motion consisting of simultaneous flexion, and rotation of the spine accompanied by shear load between two adjacent vertebrae in anterior-posterior direction. Why is it so difficult to evaluate the load that our body must support? There are various reasons that can be summarized into a short answer: The musculo-skeletal system is an open system due to muscle physiology. There is no valid computational model that describes the behaviour of the muscle in agreement with thermodynamics laws. One of the first muscle force models is based on the proportionality between the force generated at the muscle and its geometry, mainly the length and cross-section. This model provides only maximum force that can be generated at the muscle. Better understanding of muscular physiology is necessary to understand the activation of the muscle and the force generation.

1.1. Muscle behaviour

We are able to move thanks to the voluntary muscular system, which is activated by the Central Nervous System (CNS) by the process that is still not fully understood. The brain receives a 'request' to move upon which it creates an electric signal that is transferred through the nerve system to the location of the

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muscle to be activated. Once signal reaches the muscle membrane it evokes a biochemical reaction that allows the depolarization of the membrane thus the action potential can travel along the stimulated muscle fibre. Activated muscle then forms a biomechanical bridge between two types of filaments that produces the muscle contraction as shown in Fig. 1. That explains why it is easier to measure the electric potential during the muscular activity than calculate the force magnitude at the instant when the muscle starts ‘firing’ or reaches its peak force. The measurement of el. potential is possible via use of electromyography (EMG). The necessity to convert el. volts into forces remains one of disadvantages of this method. This paper will deal with present study based on the evaluation of muscular activity by means of kinematics and dynamics applied to a recorded data to compute a specific muscular forces during normal walk of a subject.

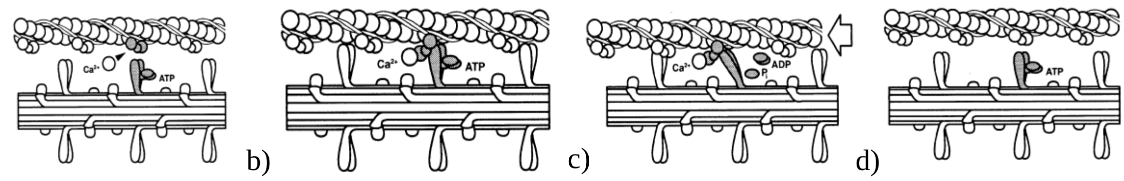
a) b) c) d)

Fig. 1: Cross-bridge connection a) activation, b) connection, c) contraction, d) detachment.

2. Methods

The geometry model of the lumbar spine as one of the most affected musculo-skeletal structures was segmented from CT scans obtained from Mater Dei hospital, Malta by means of Simpleware software. Through segmentation of geometry a FE model was created, and simplified material models, characterizing tissue property as a part of the segmentation process from CT scans as listed in Tab. 1. Then the three load models were created - the one of 2 kN compressive normal force, the second model is based on the published data (Arjmand et al., 2006) consisting of normal, and shear force together with flexion moment. The third model is a subject specific load model based on the processed data measured in the newly established Motion laboratory at University of Malta. The motion is captured by means of VICON system consisting of an eight high speed infra red cameras of 4 MP with frequency of 1000 Hz that are able to capture the light reflected from the markers attached to a moving object; in our case to the bare skin of the subject. The force plate installed within the system measures the subject’s response in static and dynamical activity during the walk. In the first step the gait cycle was recorded while in the second step the recorded data was processed by an open software AnyBody to obtain the muscular forces acting on the spine mainly on the selected vertebra.

Tab. 1: List of material properties as assigned to components of the FE segment.

Component	Young’s Modulus E[MPa]	Poisson’s Ratio ν [–]
Cortical Bone	12000	0.3
Cancellous Bone	150	0.3
Endplate	500	0.3
IVD	550	0.3
Cartilage	24	0.4

2.1. Gait cycle

When we walk, we provide support and propulsion to our body. The term gait refers to the manner of walking, rather than the actual walking process. The analysis of a gait cycle provides important information either in case of pain reported by the patient or an evaluation of the surgery outcome to compare the subject's gait with a ‘normal’ gait. For this purpose the time between two consecutive identical events of walk is considered as a one gait cycle. It is a common practice to start the cycle with first contact of the foot with the ground called ‘heel strike’. The subject was asked to move on the platform with a normal walking speed for minimum of three successful trials. The recorded positions of all reflective markers were saved, visually inspected mainly the correct record of two specific events in the gait cycle called 'heel strike' and 'toe off', which must happened within the capture area of the force plate, and for any 'drop off' markers causing then discontinued trajectories. The data was cleaned and released in C3D format for further investigation by means of use AnyBody open software.

2.2. Evaluation of muscular force using AnyBody

The AnyBody is an open simulation software with aim to analyze the musculoskeletal system, mainly humans, as a rigid-body system. In engineering tasks the application of kinematics and dynamics has a direct approach to compute a position, acceleration, and velocity based on the known applied loads. This is not the case for analysis of musculoskeletal systems where goal is to compute the muscular forces. The fundamental problem lays in the amount of muscles that is higher than it's necessary to drive the system with finite degrees of freedom. This means that there are infinitely many muscle recruitment patterns that are acceptable from a dynamical point of view (Damsgaard et al., 2006), often referred to as the redundancy problem of the muscle recruitment. There are two types of dynamical tasks – forward approach is very common in engineering to compute the motion that is driven by a defined force, and inverse dynamics where the computation of forces is derived from the known trajectory. This is a case of musculoskeletal systems. Thus solving statics and dynamics Eq. (1) can provide computation of forces and moments developed at the joints.

$$\mathbf{C}\,\mathbf{f} = \mathbf{d} \tag{1}$$

Where **C** represents the matrix of coefficients within the static equations that are dependent on the position, **f** is a matrix of muscular and reaction forces, and **d** is a matrix of external and inertia forces. Use of inverse dynamics analysis for a system of equations that include the effect of muscles resolves indeterminacy but the system of muscle recruitment has to be optimized by introduction of an objective function $G(f^{(M)})$ to minimize the muscular forces, subject to a system of static Eq. (1) with the prescribed condition

$$0 \leq f_i^{(M)} \leq N_i, \quad i \in \{1, \dots, n^{(M)} \tag{2}$$

Defining $f_i^{(M)}$ as the force of n-th muscle in the system while N_i represents the normalizing factor defining typical muscle strength. There are presently available three objective functions – polynomial criterion (3)

$$G(f^{(M)}) = \sum_{i=1}^{n} \left(\frac{f_i^{(M)}}{N_i}\right)^p \tag{3}$$

soft saturation criterion, and the Min/Max criterion (4), which represents the solution of a polynomial criterion while the polynomial degree $p \rightarrow \infty$ is considered

$$G(f^{(M)}) = max\left(\frac{f_i^{(M)}}{N_i}\right) \tag{4}$$

which leads to a distribution of muscular forces that keeps maximum relative load to minimum.

2.3. Results simulation of AnyBody and FEA

AnyBody software provides the user of AnyScrip module with possibility to control any variables and parameters associated with the model. The user has possibility to create his own model or select suitable model from AnyBody Managed Model Repository (AMMR). We adopted a generic model AnyBody with markers, which was optimized to fit the segment lengths and all other parameters recorded in our C3D database and the model can follow the marker trajectories, which was done by calling different subroutines and their adjustment. Once the model was optimized the subject specific features had to be taken into account via morphing the shape of the generic L4 vertebra of AnyBody model into the shape of a L4 model created from CT scans as shown in Fig. 2. The morphed L4 vertebra was imported back to AnyBody model via specific subroutine performing 'custom scaling'. Then the inverse dynamics was performed on the system to obtain the muscular forces that were exported in XML format to be readable by AnyFE subroutine that converts the XML data into a format suitable for FEA in ANSYS. The spinal segment as described earlier was loaded into ANSYS 15 and the coordinate system

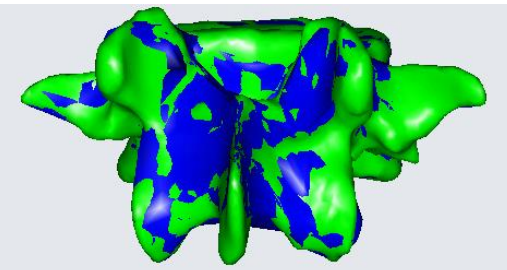

Fig. 2: The result of morphing -Green target L4 CT based with the blue transformed generic STL L4 vertebra.

was aligned to correspond the one in AnyBody prior to running a code of instruction that translates the XML data through ANSYS Parametric Design Language. Within this code a local coordinate system for each muscle insertion is created, the computed force is applied to this point, and connection between the point and the bone surface is set. Then boundary conditions were set to constrain the inferior endplate of L5 vertebra while the nodes between IVD and adjacent vertebrae were coupled by constrain equations.

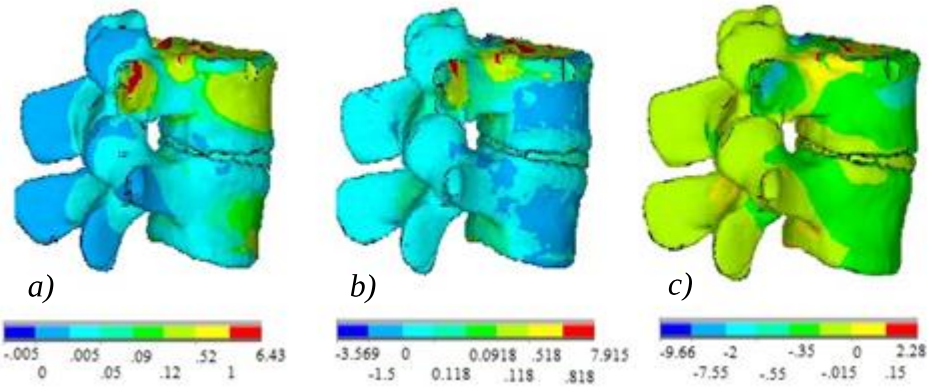

Fig. 3: Stress contour plot [MPa] at 0 % gait cycle (heel strike) – a) Von-Mises stress, b) σ_1 principal stress, c) σ_3 principal stress.

3. Conclusion

The results shown in Fig. 3 demonstrate a significant difference compared to response to two forces and moment as shown in Fig. 4 (Sant et al., 2012). The major disparity originates from the load distribution; where 64 points at 'physiologically correct positions' originated from the scaled generic AnyBody model opposed to a single point of load application at the centre of IVD. The AnyBody load forces vary in directions and magnitudes between 0 – 800 N maximum while the other model provides normal force of 688 N, shear force of 90 N and sagittal moment of 3.1 Nm. This comparison shows the necessity of better load models to simulate more specific loads corresponding closer to reality. As the von Mises, σ_1, and σ_3 stress shows in Fig. 3 in case of AnyBody load model generated maximum of 6.43 MPa, 7.92 MPa, and -9.66 MPa respectively, while the load by normal, and shear force accompanied by moment presented in Fig. 4 the stresses reached 18.47 MPa, 116.16 MPa, and -21.37 MPa respectively. The drawback of the improved loads models lays in the computational power and need for long CPU time.

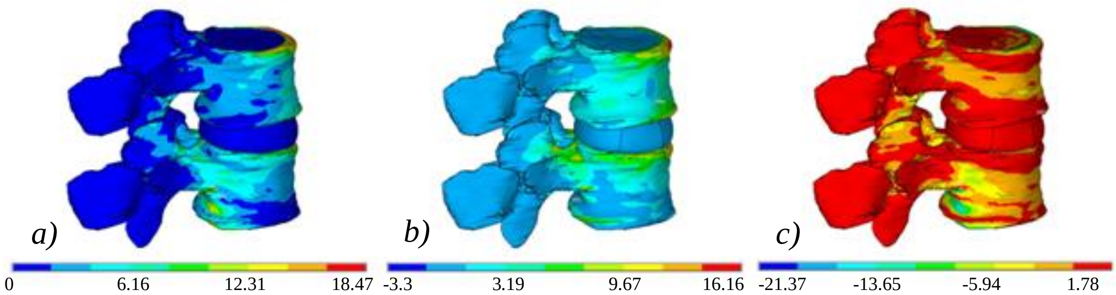

Fig. 4: Stress contour plot [MPa] – a) Von-Mises stress; b) σ_1 principal stress; c) σ_3 principal stress.

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