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Original scientific paper

SIMULATION OF WEAR IN A SPHERICAL JOINT WITH A POLYMERIC COMPONENT OF THE TOTAL HIP REPLACEMENT CONSIDERING ACTIVITIES OF DAILY LIVING

UDC 617.582

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Abstract. *The present study assesses the impact of the main typical activities of patients' daily living (ADL) after total hip arthroplasty (THA) on the wear parameters of sliding couple's materials by simulating linear and volumetric wear according to the Archard's law in a spherical joint with a polymeric element of the total hip replacement (THR). The mathematical wear model, built on the basis of algorithms and custom codes of the finite element analysis in ANSYS and MATLAB software systems, has been studied numerically. The activities used in the model are: level walking, stair ascending-stair descending, chair sitting-chair rising, and deep squatting. They were described by typical waveforms of the angular displacements of the THR's femoral component and the waveforms of the applied force. The results of the simulation show that for the same duration the overall wear value with ADL is significantly higher than in the case of level walking according to the requirements of ISO 14242-1. Therefore, the evaluation of the wear value for ADL is more informative for predicting the functional life time of the THR. Analysis of the simulation results shows that the amount of wear calculated for all activities separately is practically the same as the overall wear value obtained at summary action of ADL. This effect of the independence of contributions to the total amount of wear of each activity makes it possible to significantly simplify the solution of the problem of wear estimation for typical activities, including stochastic ones.*

Key Words: *Activities of Daily Living, Total Hip Replacement, Wear, Finite Element Simulation, Spherical Joint*

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1. INTRODUCTION

Activities of daily living are attracting increasing attention in the studies related to their impact on the life behavior of an ordinary patient in the post-operative period after THA since they represent more natural conditions of his life. In addition to level walking, these activities can include the most frequent ones, such as stair climbing, chair sitting and squatting [1].

In some studies, the maximum range of motion (ROM) in different planes in the hip joint after THA was determined with different activities [2-4]. Reference [5] estimates the effect of ADL on the fatigue longevity of the implant's stem and its microscopic displacement using a test device. But the greatest number of studies is related to kinematics and kinetics of large joints of the lower limb, such as the hip, knee and ankle with various activities. For example, only kinematics during stairs up, stairs down was studied in [6-8], and kinematics and kinetics of these same activities were studied in [9-11]. In Refs. [12, 13], the kinematics of the chair up activity (during standing up from the chair or, in other words, sit-to-stand) was studied, and the kinematics of the chair down, chair up (or stand-to-sit-to-stand) were studied in [14-16]. In [17, 18], studies are carried out of both kinematics and kinetics for the chair up of large joints of the lower limb. In Refs. [19, 20], kinematics of the same joints was studied during squatting activity. In addition, for the completeness of the ADL evaluation, the frequency and duration of each of the activities evaluated in [21] is of particular interest. It should be noted that in many of the above references, the results are presented in terms of the maximum values of the corresponding parameters, for example, angular displacements, loads, moments, and very rarely in the form of wave dependencies with the cycle time of the specified parameter. Moreover, in spite of the fact that each of the joints has a minimum of three rotational degrees of freedom, and the vector of the applied force, as a rule, has non-zero projections in three-dimensional space, most of the data presented are given in a simplified form. For example, the angular displacement is indicated only in one plane, only the resultant vector of the applied force is indicated, the study of this motor activity is not performed during the whole time step cycle, but only on the stance phase, etc.

Of special interest are studies in which estimates of the ADL impact on the wear parameters of artificial joint elements in terms of, for example, linear, volume or gravimetric (mass loss) wear are performed. In this connection, very few references have been found in the available literature. In particular, in ref. [22], the results of studies of wear performances on a device for wear testing a total knee replacement for various ADL, which were previously taken into account by us in the synthesis of a bio like artificial knee joint [23], are presented. And ref. [1] describes similar studies of a THR's spherical joint with a polymeric element, but on a specialized device for wear testing spherical sliding couples. It is clear that for successful implementation of such tests it is necessary to know the real kinematic motion dependencies in the joint, including angular displacements in three planes for the full step cycle for all the activities used, and also their kinetics, i.e. the coordinate's change of applied force vector during a similar step.

An attempt of a mathematical simulation of wear according to the Archard's law [24, 25] in a spherical joint with a polymeric element of the THR for various ADL using approximating expressions depending on the load parameters, head dimensions and surface roughness was undertaken in [26]. But despite the potential possibilities of the FEM used in the work, the authors made a number of serious simplifications, as a result of which it was possible only to judge the comparative evaluation of the ADL impact on the volume wear of the polymeric element.

The objective of this study is to assess the impact of the main typical activities of patients' daily living after THA on wear parameters and their contribution to the overall wear value by simulating linear and volumetric wear according to the Archard's law in a spherical joint with a polymeric element of the THR. The study was carried out on the basis of finite element analysis methods using previously developed approaches, detailed in [27, 28]. In the computational algorithms implemented in ANSYS and MATLAB, the kinematic waveforms of the angular movements of the femoral component and the waveforms of the coordinate's change of applied force vector, characteristic for the main typical activities, were used.

A review of the available literature data suggest that the impact of the patient's daily activities in a post-operative period after THA on the wear of a friction couple has been fulfilled for the first time and can serve as additional clarifying information in selecting materials and system synthesis of innovative designs of the THR with a polymeric element [29]. The algorithms of numerical modeling of wear at the synthesis stage developed in this paper, in contrast to simulation on simulators, make it possible to significantly accelerate and reduce the cost of analyzing the set of variants of new friction couples and of choosing the best among them in accordance with the specified quality criteria.

2. Materials and Methods

The model of the THR's sliding couple, described in detail earlier in Refs. [27, 28], contains a solid femoral ball head of cobalt-chromium alloy or ceramics (alumina or zirconia) with widely used standard diameters of 32 mm employed against a soft (UHMWPE) acetabular cup. The radial clearance between them is of 0.15 mm. Such a so-called ball-in-socket mated couple can be considered as rigid-to-soft one, where only the soft cup is subjected to wear. The elasticity modulus and Poisson's ratio were chosen as 1.4 GPa and 0.46, respectively, for the cup (conventional UHMWPE) and 210 GPa and 0.3, respectively, for the head. The right hip joint is defined in anatomical fixed coordinates $x' y' z'$ and shown in Fig. 1. In the simulation of wear, a simplified coordinate system *XYZ* fixed to the cup and placed in its center is used (Fig. 2). The movable coordinate system used for the Euler angles coinciding with the center of the cup, is placed in the center of the head and fixed to the head. The head has three rotational degrees of freedom, known as FE (flexion-extension), AA (abduction-adduction), and IOR (inward-outward rotation).

The wear simulation for a soft cup was based on the classical Archard's law according to it, for the ideal uniformly loaded isotropic surfaces with a nominal contact pressure in the linear elastic condition, an accumulative local linear wear depth at the contact surface $\Delta H(\theta,\varphi)$ in a spherical coordinate system in a discrete kind can be described as following [30]

$$\Delta H(\theta,\varphi) = \sum_{i=1}^{n} k_w \sigma(\theta,\varphi,t_i)\Delta S(\theta,\varphi,t_i) \quad (1)$$

where $\sigma(\theta,\varphi,t_i)$ is a normal contact pressure between the counter-face surfaces at the same point of time instant t_i of the gait cycle; $\Delta S(\theta,\varphi,t_i)$ is an increment of the arc sliding distance between the adjacent measuring points under the same conditions; k_w is a wear factor which depends on the material, nature of the surface and, as was found, the nominal contact pressure.

A modified formula for determining the depth of wear is presented in [28], considering the change in the wear factor as a function of the contact pressure.

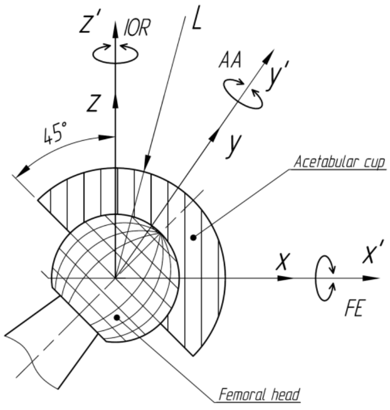

Fig. 1 Front view of the right hip joint with the specified directions of rotation (L is a resultant load vector) [27, 28]

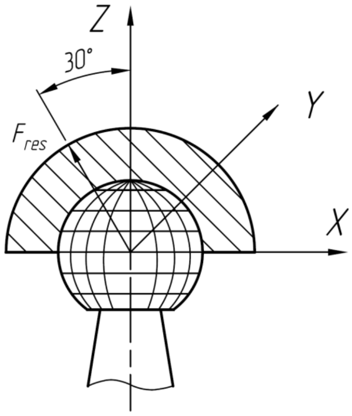

Fig. 2 A simplified coordinate system *XYZ* [27, 28]

The finite element analysis of the wear depth model created by considering dependence (1) is performed in ANSYS and MATLAB software systems. Kinematic waveforms of the angular displacements of the femoral joint component are used to determine the increment of arc sliding distance $\Delta S(\theta,\varphi,t_i)$, and in determining normal contact pressure $\sigma(\theta,\varphi,t_i)$ by solving the contact problem, the change in magnitude of the applied force vector during the step cycle is taken into account. The calculation algorithm is presented in detail in [27, 28], where the parameters of wear are studied during the level walking in two versions: according to the demands of ISO 14242-1, as well as for profiles of angular femoral positions, measured by Jonhston and Smidt, and for patterns of the applied force, measured by Paul [31]. In fact, the patient, in addition to level walking, daily performs other motor actions, the consideration of which allows determining the wear parameters closer to the real ones.

Ref. [1] shows the results of the THR's wear tests on the simulator taking into account the next main typical normal activities of the patient's daily living: level walking, stair ascending and stair descending, chair sitting and rising, and deep squatting. In the same study, based on the frequency of the cycle of each activity, the relative duration of each type of activity, reduced to the same cycle duration for 1 s, is determined, and is distributed as follows: 44% - level walking, 24% - stair ascending-stair descending, 12% - chair sitting, 12% - chair rising, and 8% - deep squatting. The tests during level walking presented in [1] are carried out in accordance with the ISO 14242-1 demands for angular movements of the femoral joint component and the applied force. For other activities, only the resulting vector of the applied force is shown for the worst case load when the patient's weight force is of 1000 N, but no kinematic profiles of the angular displacements are presented.

In the current study, numerical wear simulation of the THR's spherical polymeric component is conducted, based on the activities indicated in [1] and their duration. At the same time, the wear during the level walking activity is evaluated using parameters regulated by ISO 14242-1. In the design scheme of the model, the outer surface of the cup is stationary relative to the base coordinate system, so when determining the applied force, the inversion method is used, according to which the force vector is applied to the center of the head and directed toward the cup. The profiles of components of the two-dimensional resulting vector of applied force F_{res}, given in accordance with ISO 14242-1 in the coordinate system shown in Fig. 1, are depicted in Fig. 3, and the profiles of components of the three-dimensional vector of the applied force corresponding to the load worst case with the patient's weight force in 1000 N for such motor activities as stair ascending, stair descending, chair sitting and chair rising, borrowed from ref. [32], are shown, respectively, in Figs. 4-8. When converting the coordinates of the applied force vector from the coordinate system indicated in ref. [32] to the simplified coordinate system (Fig. 2), the average anteversion angle equal to 12.5º is taken into account [32], and the angle between the axis of the stem's neck of the THR and the vertical component of the force vector when the position of the axis of the stem is parallel to the axis of the femur, is equal to 45°. Therefore, the directional coefficients of the force vector coordinates transformation are taken as follows

$$k_X = \cos 12.5^\circ \cos 45^\circ = 0.69\,;\ k_Y = \cos 12.5^\circ = 0.976\,;\ k_Z = \cos 45^\circ = 0.707\,.$$

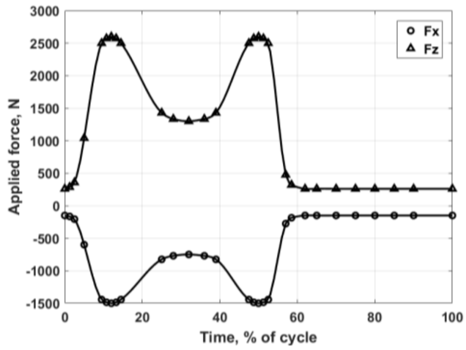

Fig. 3 Waveforms of two-dimensional force vector according to ISO 14242-1

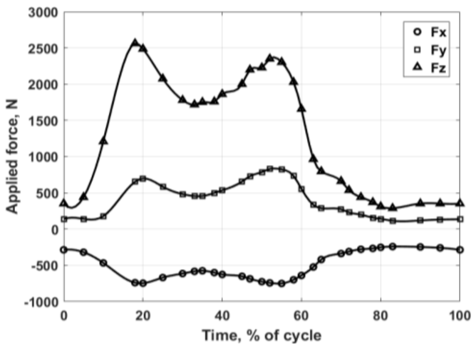

Fig. 4 Waveforms of three-dimensional force vector for stairs up

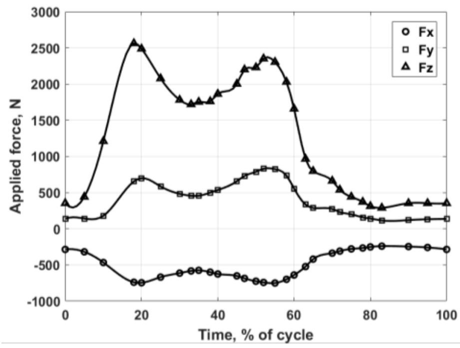

Fig. 5 Waveforms of three-dimensional force vector for stairs down

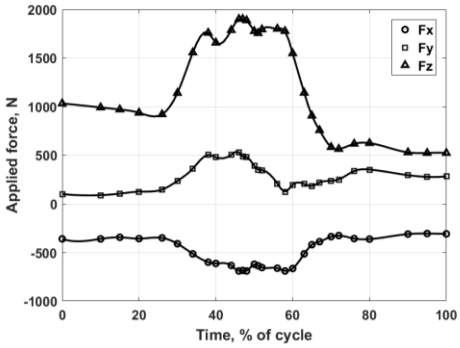

Fig. 6 Waveforms of three-dimensional force vector for sitting down to chair

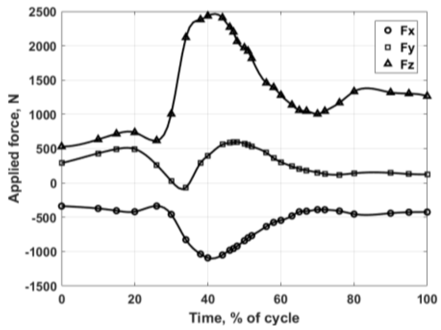

Fig. 7 Waveforms of three-dimensional force vector for standing up from chair

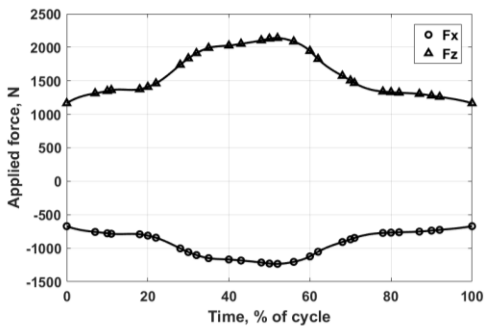

Fig. 8 Waveforms of two-dimensional force vector for deep squatting

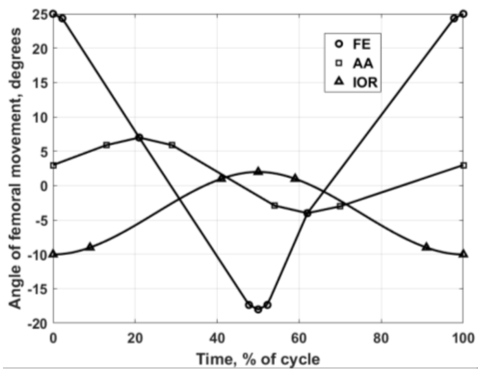

Fig. 9 Waveforms of angular movements of the femoral head according to ISO 14242-1

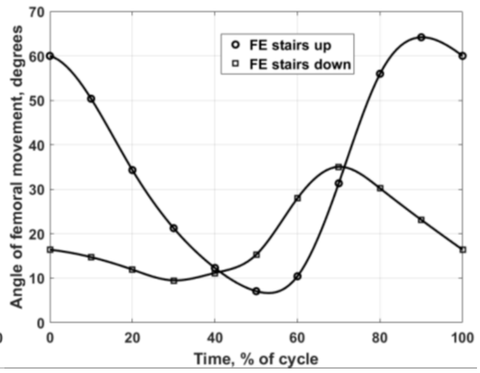

Fig. 10 Waveforms of angular movements for stairs down-stairs up [10]

Reliable profiles of the change in the coordinates of the applied force vector during deep squatting are not found in any available literature data. Therefore, to estimate the wear parameters for this motor activity, the profile of the resultant force vector, given in [1], is used, which is also transformed at the transition to a simplified coordinate system in analogy with the level walking and is shown in Fig. 8.

The kinematic waveforms of the angular displacements of the femoral joint component required for modeling which are usually used by researchers for above ADL and are available in a number of literature sources, are shown in Figs. 10-13 with the corresponding references from which they are taken.

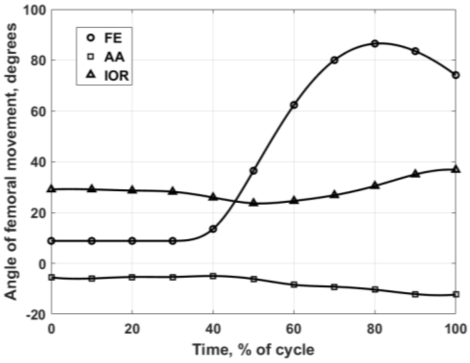

Fig. 11 Waveforms of angular movements for sitting down to chair [14]

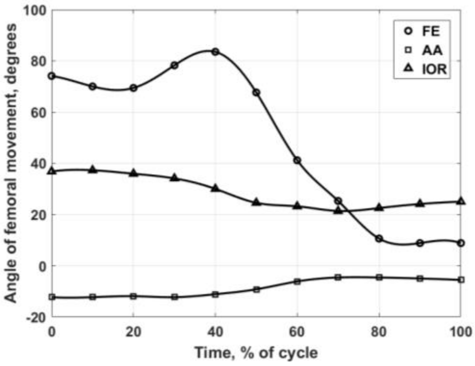

Fig. 12 Waveforms of angular movements for standing up from chair [14]

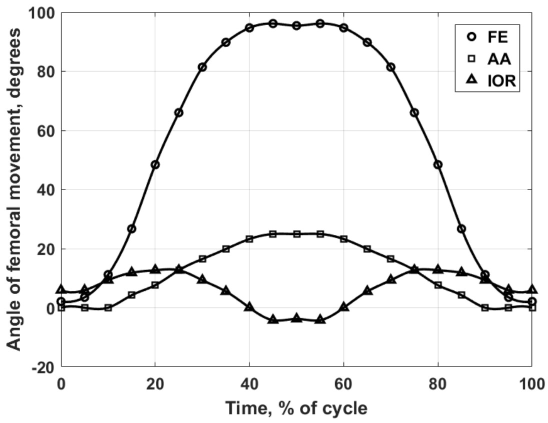

Fig. 13 Waveforms of angular movements for deep squatting [19]

Numerical simulation of wear is carried out in two ways. The first one is that all ADL listed above are lined up in a predetermined order with a preset duration of motor task for the one simulation cycle: 44% - level walking, 24% - stair ascending-stair descending, 12% - chair sitting, 12% - chair rising, and 8% - deep squatting. In order to follow the approximate number of gait steps at which the correction of the supporting surface of the cup geometry is performed by moving the nodes to the value of the obtained linear wear [28], their minimal number of 500,000 is taken.

In this case, the number of replicates of each activity, which is necessary to fulfill their duration, is taken to be a multiple of 20,000 wherein they correspond to the number of steps: 220,000, 120,000, 60,000, 60,000, and 40,000, respectively. The total number of steps in the model is assumed to be 5 million, which allows 10 cycles of the geometry correction of the cup's supporting surface. Given that each step is divided into 25 time slots, the solution of contact problem in ANSYS is carried out inside each one, and it is easy to calculate the number of such solutions that for six separate activities it makes up 150 for one cycle, while for 10 cycles it is of 1500, which significantly increases computer time. Taking into account that in this work the task of obtaining absolutely accurate values of wear parameters is not being posed, and the assessment is mainly being made of the qualitative impact of each ADL on wear parameters and their contribution to the overall wear value, the chosen simulation duration allows us to do this sufficiently earnestly.

The second way of simulating is by evaluating the contribution of each of these ADL separately to the total amount of wear. For this, only one of these activities is modeled in each cycle with its duration in the range of up to 10 cycles of the geometry correction of the cup's supporting surface, i.e. up to 5 million steps.

3. RESULTS

Figure 14 shows the linear wear patterns for simulation of the ADL influence for the first and second ways for each of the activities, and in Fig. 15 is for the same ones, but for the cumulative volume wear.

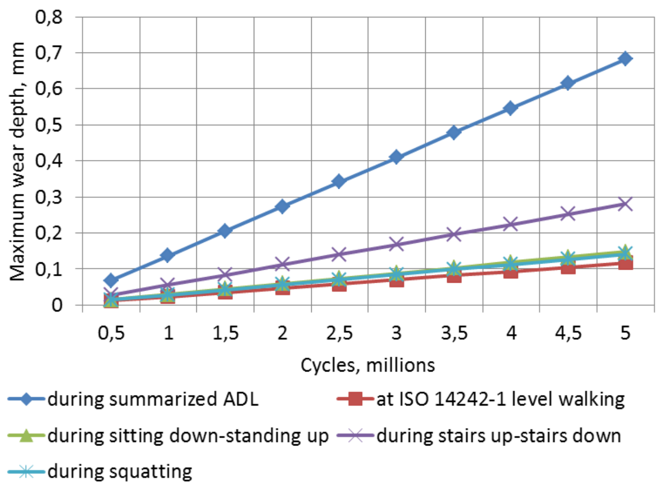

Fig. 14 Graphs of linear wear during summarized ADL and for each the ADL contributed

In Figs. 16-17 profiles are depicted of linear and cumulative volume wear during ADL simulation by the first way and in the case when the patient performs only level walking according to ISO 14242-1 demands for the same cycle duration as for the total ADL. Fig. 18 shows the contribution of each of the above ADL to the overall wear value.

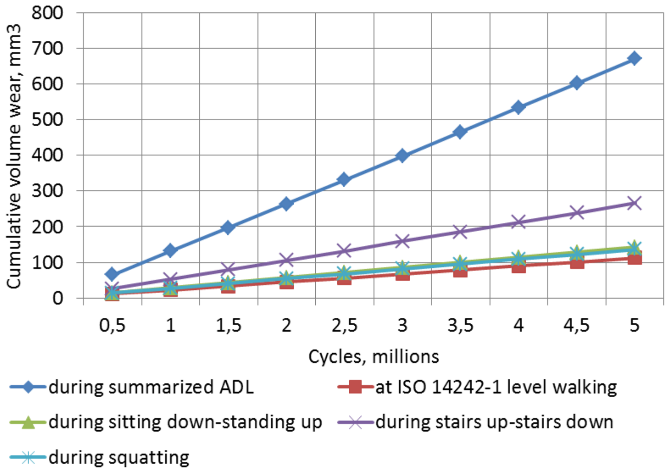

Fig. 15 Graphs of cumulative volume wear during summarized ADL and for each of the ADL contributed

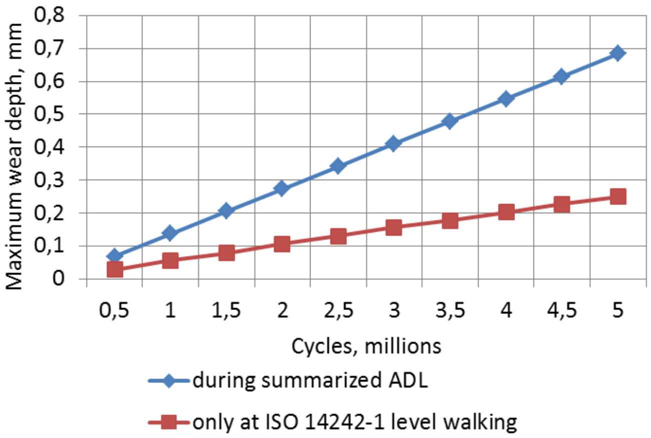

Fig. 16 Graphs of linear wear during summarized ADL and for level walking according to ISO 14242-1 at the same cycle duration as during ADL

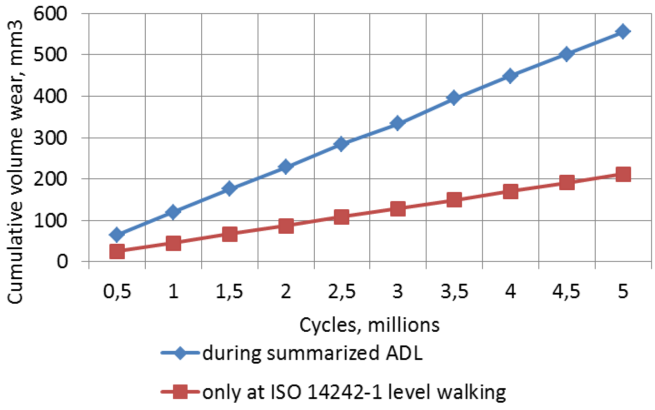

Fig. 17 Graphs of cumulative volume wear during summarized ADL and for level walking according to ISO 14242-1 at the same cycle duration as during ADL

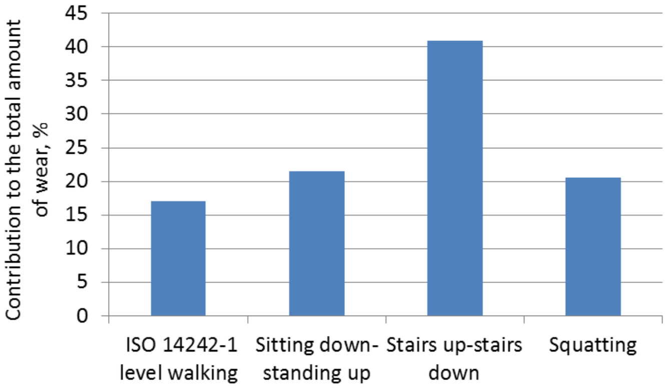

Fig. 18 Bar chat on the contribution of each the ADL into the overall amount of wear

4. Discussion

The profiles depicted in Figs. 14-15 show a pronounced linear relationship observed for both linear and cumulative volume wear from the number of gait steps for all indicated ADL and their total impact. Their elementary analysis also makes it possible to note that the total wear calculated due to effect of each separate ADL practically coincides with the overall wear value obtained at summary action of ADL. This effect of the independence of contributions to the total amount of wear of each activity makes it possible to significantly simplify the solution of the problem of estimating wear for atypical activities, including stochastic ones.

From the profiles shown in Figs. 16-17 it follows that the total wear value for ADL is significantly higher than in the case of only level walking, which is recommended by ISO 14242-1 for the THR wear tests. In this case, the largest contribution to the total wear amount (about 41%) gives the stairs up-stairs down moving (see Fig. 18), despite the fact that its duration is only 24% of the total cycle time. About 21% of the total wear amount occurs during sitting down-standing up and squatting, although in duration they differ 3 times. And, finally, the least contribution to total wear (about 17%) pay in level walking, despite the longest duration of the action (44% of the total cycle time). Therefore, the estimation of the wear value with considering of ADL is more informative for predicting the functional longevity of the THR compared to the recommended one during level walking at ISO 14242-1.

The linear form of the dependencies of the wear characteristics obtained in Fig. 14-17 can be probably explained simply by the geometric relationships between the parameters of the figures formed from the sphere. The shape of the contact in the spherical joint on the surface of the cup as the head is deepened into it during its wear is an approximately spherical surface of the ball segment where, as the head becomes deeper, this surface and the volume of the segment increase linearly depending on the wear depth according to known mathematical relationships. This conclusion is also confirmed by the results obtained in [32] and in a number of references cited in [27, 28].

The values of wear parameters for level walking, presented in Figs. 16-17, are consistent with their values in [27, 28, 30, 31], which indicates the reliability of the study results. But they are in some ways different from simulation results on the simulator specified in [1], although the used head diameter is the same and is of 32 mm. This may be due to the fact that the kinematic profiles of the angular movements of the femoral component used in [1] likely differs from those used in this study. These profiles influence the length of the slip path of points on the surface of the head over the surface of the cup and, by that, affect the depth of wear according to formula (1). Therefore, the presence of such a factor could lead to serious differences in the results obtained.

The results obtained in this study show that in order to extend the resource of the THR, patients after THA can be given the following recommendations: significantly restrict the activities of stair ascending-stair descending by using the elevator and deep squatting by its possible replacement for chair sitting-chair rising.

5. Conclusion and Outlook

In this study, the effect of the main typical activities of patients' daily living after THA on wear parameters and their contribution to the overall wear value is estimated for the first time by simulating linear and volumetric wear according to the Archard's law in a spherical joint with a polymeric element of the THR. The study was carried out on the basis of finite element analysis methods using previously developed approaches, detailed in [27, 28]. In the computational algorithms implemented in ANSYS and MATLAB, the kinematic waveforms of the angular movements of the femoral component and the waveforms of the coordinate's change of applied force vector, characteristic for the main typical activities such as level walking, stair ascending-stair descending, chair sitting, chair rising, and deep squatting, were used. To verify the adequacy of the mathematical model used in the study and to confirm the reliability of the results obtained, in the near future an experimental evaluation of wear will be performed on the device developed for wear testing of the THR. Its design and operation principle of which is described in [33, 34] and which allows to replicate with high accuracy all kinematic and kinetic parameters of movements in the joint, characteristic for ADL. Such confirmation is necessary for the development of well-founded methods for predicting the functional life time of the THR's different design for arbitrary regimes of the patient's motor activity after THA.

Simulations in this paper have been carried out under assumption of validity of the Archard's law in its simplest form. However, it is known that it is only a very rough approximation. Under certain conditions, the wear in a given tribological pair can increase dramatically or vanish almost completely [35]. It would be interesting to find out the conditions for transitions between "normal" wear, severe wear and almost wear-less conditions. Further, in the present approach we have not explicitly considered the transport of the wear particles in the frictional zone, which, however, may significantly influence the wear process [36]. Finally, it would be interesting to consider a possibility of similar simulations using the Boundary Element Method which has shown its specific efficiency for simulation of contact problems [37].

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