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Biomechanical factors influencing the beginning and development of osteoarthritis in the hip joint

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Summary. Osteoarthritis (OA) can be used as a common name for a group of overlapping pathological conditions when the balance between the processes of degradation and synthesis, in individual parts of the cartilage, is disturbed and leads to gradual cartilage destruction. A preventive approach toward OA helps with a timely diagnosis and subsequent treatment of this disease. One of the significant risk factors affecting development of hip joint OA is the mechanism and magnitude of mechanical loading on the joint. The main motivation for this work was to verify the hypothesis involving a pathologic cycle (overloading – change of locomotion – overloading) as contributory to the development of OA and whether it can be stopped, or at least partly decelerated, by a suitable change of movement stereotypes. Providing that there is a natural balance of muscular action, from the beginning of OA, the development of OA can be significantly decelerated. The return to a natural force balance can be achieved using suitable exercise and strengthening of muscular structures. In order to verify the hypothesis, we undertook experimental measurements of gait kinematics and a computational analysis of the hip joint using the Finite Element Method.

Key words: Osteoarthritis, hip joint, kinematics, FE Method, mechanical loading

Introduction

Osteoarthritis (OA) can be used as a common name for a group of overlapping pathologic conditions when the balance between the processes of degradation and synthesis, in individual parts of the cartilage and subchondral bone, is disturbed and leads to gradual destruction of these structures [7]. OA is a common, painful, degenerative joint disease. It affects 12% of the population, and after heart diseases, it is the second most common cause of disability in males over 50 years of age and in females after menopause. OA represents not only a health issue but also a significant social and economic problem. A preventive approach toward OA diseases can therefore help with a timely diagnosis and subsequent treatment of the disease.

One of the significant risk factors affecting the development of hip joint OA is the mechanism and magnitude of loading on the joint [4]. Mechanical loading of the joint significantly affects the condition of the cartilage [9]. Loading on the joint cartilage is primarily repetitive in nature. This kind of loading is advantageous not only from a mechanical point of view, but also from a nutritional point of view. Healthy joint cartilage is a porous tissue containing a large amount of synovial fluid [7]. During loading of the joint, this fluid is squeezed out and the cartilage, which decreases its thickness. When the loading is relieved, the synovial fluid reenters the cartilage and the cartilage returns to its original dimensions. The role of the fluid in the cartilage is not only mechanical but also metabolic – it perfuses the cartilage with nutrients. This happens while synovial fluid, acting as an intermediary, moves between the cartilage, which has no blood circulation, and the synovial membrane, which is nourished by blood vessels. The afore-mentioned fluid movement is necessary for appropriate functioning of the cartilage, and disturbing this physiological condition by loading changes, e.g. by permanent pressure on the cartilage,

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can lead to local mechanical overloading of the cartilage and its primary degeneration [5]. In the area under constant pressure, there is no physiological nourishment of the cartilage tissue, which results in irreversible damage to the tissue. In the primary degenerative area (PDA), there is an increased concentration of stress, due to mechanical loading, and the cartilage is misshaped, and therefore overloaded again, which leads to secondary degeneration and the damaged area gets larger. The human body attempts to rectify this situation by changing locomotive patterns, causing additional mechanical loading on even smaller areas of cartilage tissue, which is then further overloaded. This continually developing circulus vitiosus, overloading – change of locomotion – overloading (OCO), is one of the main factors affecting the development of hip joint OA.

The objective of this work was to verify the hypothesis that the pathologic cycle, which contributes to the development of OA, i.e. OCO, can be decelerated with a suitable change in mechanical loading on the joint. Pain during movement causes contraction of muscle groups that participate in the particular movement. Gradually these contractions become permanent and there are permanent changes in the size and mode of loading on the affected joint. However, provided that there is a natural balance of muscular action, from the beginning of the process, the development of OA can be significantly decelerated. If the development of OA is decelerated in the early stages, the necessity for a total implant is delayed, and this is very important especially with regard to the lifespan of total hip replacements and the necessity for a second implant. Return to a natural force balance can be achieved by targeted exercise and strengthening of specific muscular structures.

Patient and methods

Experimental measurements

In the first part of this project, experimental measurements of a patient's gait, with the objective of determining the orientation angles of individual segments of the lower limbs and pelvis during the gait, and to record reaction forces and moments while the patient bares weight during the step process, were performed. This measurement was performed with a patient having clinically indicated right hip joint OA – grade 1. The measurements were performed before and after 8 months of physiotherapy. The results of these measurements were used as input data for computational

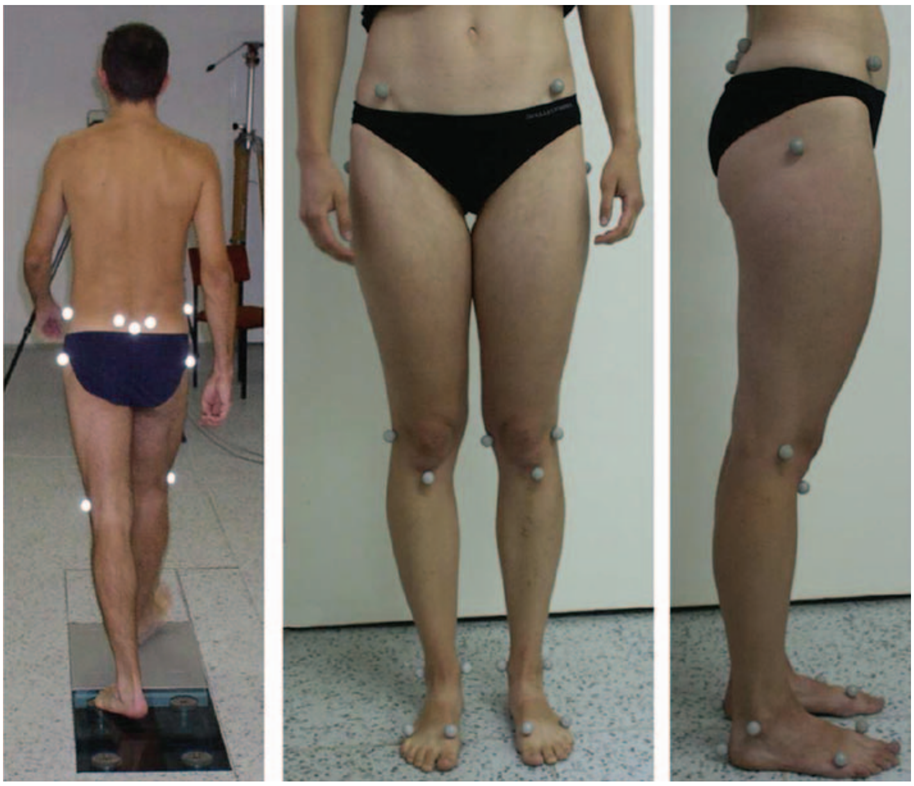

Fig. 1: Demonstration of experimental measurement of gait kinematics while stepping on a sensor (left) and detailed placement the markers on the patient's body

analyses of hip joint loading – using the Finite Element Method (FEM) and as a method for objective evaluation of the efficiency of applied treatment methods.

The measurement of gait kinematics itself was carried out using the Qualisys® motion capture system (Qualisys AB, Sweden) consisting of 6 infrared cameras recording the spatial motion of markers – fixed to the patient's body at points of anatomic importance (frequency of 200 Hz) (see Fig. 1). The trajectories of the motion of individual markers were exported as continuous data that determined the spatial location of the markers as they varied with time. Using Qualisys®, individual segments of the lower limb and pelvis were designed, and centers of rotation were determined, into which the data from marker movements in space and time were imported. The magnitude of reaction forces and their components, during the step process, was measured using two measuring plates (Kistler® 9285, Kistler, Switzerland), which recorded data with a frequency of 500 Hz. Recording of motion and reaction forces were synchronous.

A slow gait was chosen as the motion reference because it is a complex motion and it is possible to monitor even slight deviations from normal (Fig. 1). The results of experimental measurement of gait kinematics were processed using a C-Motion® (C-Motion Inc., USA) system and evaluated using MATLAB (MathWorks Inc., USA) software. In order to preserve equal conditions, only data for a speed-of-gait of 1.2 ms^{-1} were processed, and the record covered complete data for 100% of the stages of a step. Using MATLAB, all the collected data were fit using a third-degree polynomial curve, where the error was never greater than 2%, and it was subsequently standardized to 100% of the phase of a step. The mean value was determined for each resulting average curve.

Hip joint FEM analysis

The second part of this paper covers complete stress-strain analysis of the hip joint using FEM with loading equal to a slow gait. The whole model was considered as a contact task, where the contact was realized between the joint cartilage of femur head and fossa acetabuli. A real geometric 3-D model was based on a reconstruction of a femur and os innominatum geometry from CT images (Computer Tomography) and reconstruction of joint cartilage and ligament geometry from MRI images (Magnetic Resonance Imaging) (see Fig. 2).

The models of individual parts of the hip joint were subsequently exported into the preprocessor of ABAQUS software, where the FEM grid of individual parts was drawn. For modeling of the joint cartilage, we used 3D pore pressure elements, where each knot was connected to a spring element. The ligaments were modeled using special connector elements, which transfer only axis loading. The topic of interest in this paper was the strain of joint cartilage during gait; therefore, it was possible to use a simplified model of bone tissue. For this reason, both bones, femur and os innominatum, were modeled as absolutely rigid bodies.

Fig. 2: Complete geometric model of the hip (left) and computational FEM model (right)

The joint cartilage in joint fossa acetabuli and on the femur head was modeled as a biphasic poroelastic material fortified with fibers. This model is based on biphasic theory [8], which represents point cartilage as a complex consisting of two non-combinable phases [1, 2, 6, 11]: solid phase (collagen – proteoglycan matrix) and liquid phase (interstitial liquid). Both these phases are considered non-compressible. The solid phase is considered as homogenous, isotropic, porous, and linearly elastic, while the liquid phase is considered to be non-viscose. The ligaments were modeled as non-linear, homogenous material; the material properties of which were taken from the literature [3, 10].

The hip joint computational model was loaded using force and kinematic conditions, which were implanted into the center of rotation of the femur head. The size of rotation angles was collected from experimental measurements, and the magnitude of the reaction forces applied to the hip joint corresponded with the patient's reaction while stepping on the floor. There were two numerical simulations analyzing loading on the hip joint, one before and one after treatment, when the patient's gait kinematics and the magnitude of forces were changed due to physiotherapeutic treatment.

Findings

To evaluate the findings of measurement of the patient's gait, we used only a comparison of the relative magnitude and range of motion of the femur with regard to the pelvis. For this evaluation, the pelvis was considered as immovable and the movement (rotation in three planes) was performed only by the lower limb. The following parameters were used for better transparency of the findings: maximum angle of rotation of femur toward pelvis φ_{max}, mean value of the rotation angle s_φ, representing the mean value of rotation angle for 100% of the step phase and the difference of mean value of rotation angle Δs_φ, which represents the difference in s_φ before and after the treatment.

The graphs in Fig. 3 show that the mutual relationships, relative to the shapes of curves representing abduction/adduction of both the patient's legs, before and after treatment, are nearly identical. A significant difference is clear only when comparing the curves before and after treatment: before treatment, there was a larger deviation relative to adduction in the first half of the step phase (both legs equally). When comparing results obtain for both legs, the difference of magnitudes i.e. s_φ is small – the difference of values s_φ before treatment is $\Delta s_\varphi = 1.2°$ and after treatment is $\Delta s_\varphi = 0.1°$. The small value of Δs_φ and similar shape of curves indicate that after treatment the loadings on both legs were nearly equal. After treatment, the maximum rotation angle (φ_{max}) for both legs decreased during adduction (4.6° for the left leg and 2.4° for the

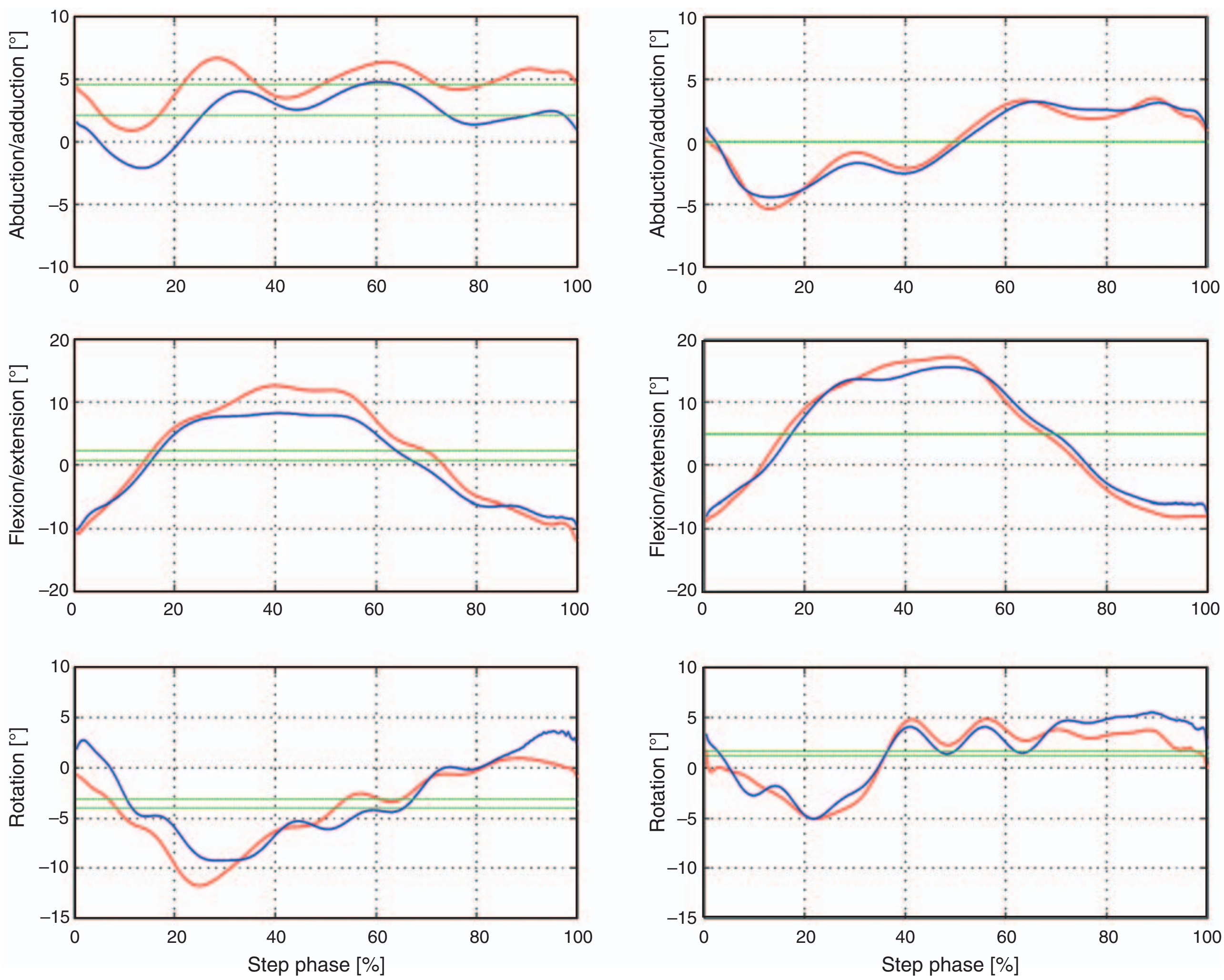

Fig. 3: Demonstration of patient's motion kinematics before (left column) and after (right column) treatment. The diagrams represent the shapes of rotation angles for each motion of left (red line) and right (blue line) leg and their mutual comparison. The mean value s_φ [°] is represented by the green line

right leg); on the other hand, φ_{max} increased during abduction (6.2° for the left leg and 2.7° for the right leg). This change made the whole process of femur rotation, for abduction/adduction, balanced and symmetrical. The patient's motion kinematics when flexing/extending both legs was nearly identical before and after treatment; the only difference was a smaller range of movement for the right leg before treatment. When comparing both legs, there was a difference in the average value of the rotation angle (s_φ) – the difference of s_φ, before treatment was $\Delta s_\varphi = 2.1°$ and after treatment it was $\Delta s_\varphi = 0.1°$. Just like in the case of abduction/adduction after treatment, there was balanced loading on both extremities, even relative to flexion and extension. It is also clear that for both limbs, there was a decrease in the maximum rotation angle (φ_{max}) during flexing (4.3° for the left leg and 8.3° for the right leg). The φ_{max} for extension stayed nearly equal for both extremities before and after treatment. The last monitored motion of the patient's gait kinematics was rotation of femur around the vertical axis.

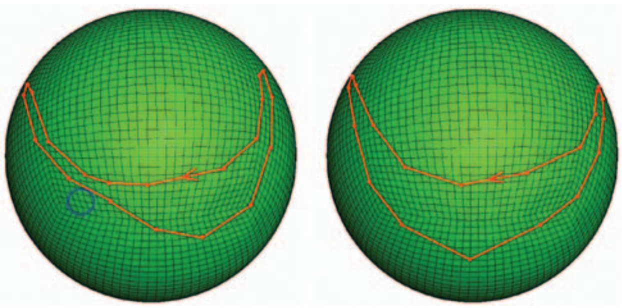

Fig. 4: Trajectory of maximum values for pressure in pores P_{por} on the femur joint cartilage surface. Before treatment (left) and after treatment (right). The highlighted area in the left picture represents the approximate location on the surface that is continually loaded during 100% of the step phase

Curve shape diagrams (see Fig. 4) representing the rotation of both legs have very similar character, but their shapes differ significantly before and after treatment. When comparing both legs, before and after treatment, there is a difference in the average value of the rotation angle (s_φ) – the difference of values s_φ before treatment is $\Delta s_\varphi = 3.4°$ and after treatment is $\Delta s_\varphi = 0.0°$. However, what is significant is not the zero value of Δs_φ after treatment, but the symmetrical distribution s_φ along the horizontal axis. There was also a decrease in values for the maximum rotation angle (φ_{ma}) for medial rotation (7.1° for the left leg and 5.2° for the right leg). The φ_{max} for lateral rotation increased for the left leg by 3.2° and for the right leg by 1.3°.

The results of the FEM computational analyses indicate that during loading equal to 100% of the step phase, there were no limit values indicating tissue damage caused by tension deformation or compression in any part of the hip joint model. For this study, it was more important to know the physiological of joint cartilage loading rather than finding specific values for tension or deformation. The main indicator supporting our hypothesis was monitoring the progress of cartilage loading during the step. For this reason, the maximum values of pressure in pores P_{por} for the point cartilage model were monitored together with the maximum magnitude of contact pressure C_{pres} in each phase of the step. These values were then used to determine the position of maximum values and to draw a trajectory of these maximum values on the femur joint cartilage surface during 100% of the step phase.

The results of numerical analyses presented in Figs. 4 and 5 clearly indicate that the locations of

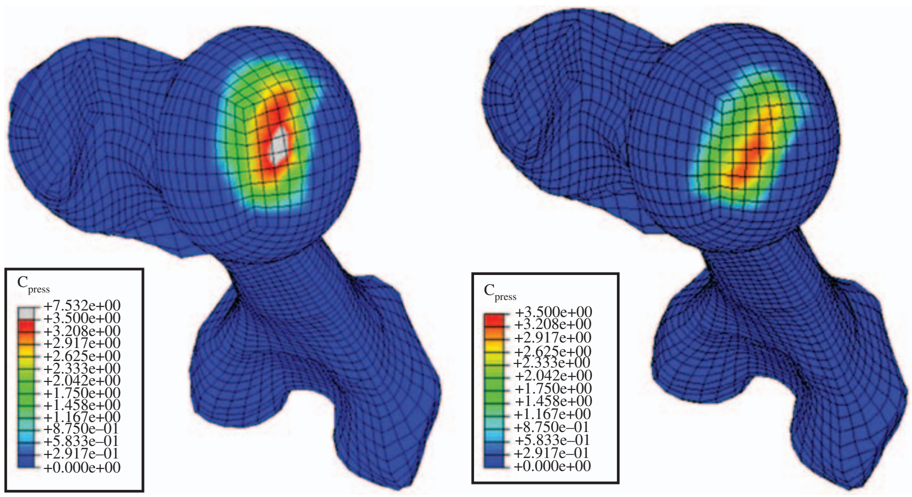

Fig. 5: Comparison of contact pressures C_{press} [MPa] on the femur joint cartilage surface: situation before treatment (left) and after treatment (right)

maximum P_{por} and C_{pres} are nearly identical. Furthermore, while simulating hip joint loading before treatment, the area of permanent joint cartilage loading was identified (Fig. 4 – left – blue circle), indicating that synovial liquid was not being reabsorbed in this area.

Discussion

The objective of the experimental measurements and numerical simulations was to verify the hypothesis that a pathologic cycle contributes to the development of OA, overloading – change of locomotion – overloading, and that the cycle can be decelerated using a suitable change of mechanical loading on the joint. Provided that there is a natural balance of muscular action from the beginning of the process, joint loading can be significantly reduced and thus the development of OA decelerated.

The collected results from experimental measurements demonstrate the influence of applied treatment on the change in range and course of the patient's gait motion. Generally speaking, the therapy increased the range of motion, especially for the affected limb; while an increase in temporary motion deviations and loading on both limbs occurred after therapy. Uniform loading on both limbs was evaluated not only according to the shapes of curves for each motion, which were very similar, but also according to the resulting difference of the average values of rotation (Δs_{φ}) before and after treatment. A verifiable improvement in the patient's gait kinematics, before and after therapy, significantly supports our hypothesis, however, in order to credibly verify this hypothesis it will be necessary to perform these measurements using a statistically meaningful sample size.

The objective of the computational FEM analyses of the hip joint model was to evaluate the influence of change in mechanical loading on tension, and especially, the reaction of joint cartilage to this load. For this purpose, we completed computational analyses before and after targeted physiotherapy. The results of these analyses do not show any significant difference in the monitored values for loading on the joint before and after treatment. There are slight differences, however, it is impossible to definitely and objectively evaluate whether the differences are caused by changes in loading (i.e. treatment) or by errors in FEM analyses. However, to support our hypothesis, the important feature was to know the "nature" of cartilage loading rather than finding specific numerical values. The main indicator supporting our hypothesis involves the progress of cartilage loading during the step. The course of maximum pressure in the pores P_{por} on the femur joint cartilage surface shows a significant change in the trajectory of these maximum values before and after therapy. Not only the course of P_{por} after therapy was slightly increased, which corresponds with an increased range of motion after treatment, but also its course compares well with the condition before treatment. Moreover, we identified areas on the joint cartilage surface (for simulation of the condition before therapy), where the pressure in pores was permanent, even when the leg was not exposed to load through body weight, i.e. in the phase of step when the leg was moving forward. In our opinion, reabsorption of synovial fluid back into the cartilage was completely or partially disrupted and thus nourishment and regeneration of the cartilage was reduced.

The main motivation for this work was to verify the hypothesis that the pathologic cycle, which contributes to the development of OA, overloading – change of locomotion – overloading, can be stopped, or at least decelerated, by a suitable change of mechanical loading on the joint. If the development of OA is decelerated in the early stages, the necessity for a total implant is delayed, and this is significant with regard to the lifespan of a total hip joint replacement and the necessity for a second implant later in life. A return to a natural force balance can be achieved using targeted exercise and strengthening of specific muscular structures.

The main motivation for this study, relative to clinical application, was to verify of the hypothesis that a pathology cycle, which contributes to the development of OA (overloading – change of locomotion – overloading), can be stopped, or, at least, slowed using an appropriate change in mechanical stress on the joint. If an appropriate treatment is used during the early phases to slow the development of OA, the necessity for total replacement can be postponed, which is essential with regards to the lifespan of a total replacement and the necessity for a second implant as the patient advances in age.

Kinesitherapy, physiotherapy and chiropractics indicated for monitored patient

After examination of the patient, the following facts were identified: failure of statics and dynamics of the thoracic and lumbar spine, bad stereotype for right hip abduction and bad walking stereotype. Dynamics and muscle imbalance of the right hip joint were limited. While standing on the right leg, there was a lateral shift of the pelvis and a slight shift in the pelvis on the left side (positive Trendelenburg – Duchenne test). The patient had developed a reasonably strong muscle apparatus.

A natural force balance can be recovered using physiotherapy and strengthening of muscle groups. The following treatments were applied to the monitored patient:

1. Removing a right sacroiliac joint blockade by Lewit [12]
2. Release of fasciae in the lumbar spine using soft techniques by Lewit
3. Post-isometric relaxation (PIR) of the m. piriformis and m. iliopsoas in the right hip by Lewit
4. Strengthening of weakened abductors of the right hip (repeated abduction in lying position under loading)
5. Post-isometric traction in the lumbar spine and right hip area
6. Stabilization of spine and hip joint in the Proprioceptive Neuromuscular Facilitation concept (PNF concept) and sensomotorics. Training on unstable surfaces
7. Training of correct movement stereotypes, gait, and sitting
8. Practice of autotherapy, i.e. muscle strengthening, auto mobilization of sacroiliac joint, and PIR (Post-Isometric Relaxation) m. piriformis.

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Conflict of interest

The authors declare that there is no conflict of interest.

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