Acetabular Loading in Active Abduction

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***Abstract*—Operative fixation of fragments in acetabular fracture treatment is not strong enough to allow weight bearing before the bone is healed. In some patients, even passive or active nonweight-bearing exercises could lead to dislocation of fragments and posttraumatic osteoarthritis. Therefore, early rehabilitation should avoid loading the acetabulum in the regions of fracture lines. The aim of the paper is to estimate acetabular loading in nonweight-bearing upright, supine, and side-lying leg abduction. Three-dimensional mathematical models of the hip joint reaction force and the contact hip stress were used to simulate active exercises in different body positions. The absolute values of the hip joint reaction force and the peak contact hip stress are the highest in unsupported supine abduction (1.3 MPa) and in side-lying abduction (1.2 MPa), lower in upright abduction (0.5 MPa), and the lowest in supported supine abduction (0.2 MPa). All body positions the hip joint reaction force and the peak contact hip stress are the highest in the posterior–superior quadrant of acetabulum, followed by anterior–superior quadrant, posterior–inferior quadrant, and finally anterior–inferior quadrant. Spatial distribution of the average acetabular loading shows that early rehabilitation should be planned according to location of the fracture lines.**

***Index Terms*—Acetabular fracture, biomechanics, hip contact stress, rehabilitation.**

I. Introduction

ACETABULAR fractures are produced by high energy injuries that often cause dislocation of the fragments with gaps and steps [1]. The goal of operative treatment of such fractures is to restore acetabular anatomy with perfect fragment reduction and stable fixation in order to enable early joint movement [2], [3]. The fixation of the fragments is not strong enough to allow weight bearing before the bone is healed [4], [5] and in some patients even physical therapy with initial passive motion and continued active exercises without weight bearing could lead to dislocation of fragments and early posttraumatic osteoarthritis [2]. Early physical therapy of patients with acetabular fractures, therefore, requires careful selection of exercises in order to prevent excessive loading of the injured acetabular region. Current guidelines for nonoperative management of acetabular fractures and postoperative management of surgical procedures in the acetabular region recommend initial bed rest followed by passive motion in the hip joint. Initial active nonweight-bearing exercises commence a few days after surgery and include active flexion, extension, and abduction in the hip in the upright position. The same set of exercises in supine or side-lying abduction is usually postponed until five to fourteen days postoperatively. Partial weight-bearing with stepwise progression usually starts six weeks postoperatively and full weight bearing is eventually allowed at ten weeks [6].

Recently, interesting information was obtained by direct measurements of acetabular contact pressures during rehabilitation exercises in subject with pressure-instrumented partial endoprostheses where it was found that acetabular pressures may not follow the predicted rank order corresponding to the commonly prescribed temporal order of rehabilitation activities [7], [8]. It has been found that hip stress magnitudes in some nonweight bearing exercises can exceed hip stress in weight bearing exercises or even gait. However, the abduction exercises in these studies were performed with angular velocities exceeding 30°/s which is considerably faster than early rehabilitation exercises in acetabular fractures which are the scope of this work. According to our knowledge, none of the studies has focused on the average loading of different acetabular regions in slow active nonweight-bearing abduction.

Due to technical complexity and invasiveness of direct contact stress distribution measurement, various mathematical models for calculation of the contact stress distribution in the hip joint have been proposed [9]–[15]. Recently, a mathematical model has been developed that enables computation of the contact stress distribution at any given position of acetabulum and also allows simulation of different body positions and variations in pelvic morphology [10]–[12]. However, such estimation of the contact stress distribution necessarily includes determination of the hip joint reaction force magnitude/direction. Noninvasive determination of the localization of dynamic acetabular loading during gait has so far been performed by inverse Newtonian computations based on kinematic measurements of individual body segments and a muscle model [16]. For slow rehabilitation exercises, the loading in the hip joint at a given leg and/or body position approximates to static biomechanical equilibrium and the hip joint reaction force can then be numerically calculated by using a muscle model and a suitable optimization function without kinematic measurements [15], [18].

The aim of the paper is to compare acetabular loading in nonweight-bearing upright, supine and side-lying leg abduction by using a muscle model for computation of the hip joint reaction force and a previously developed mathematical model of contact hip stress distribution. With this knowledge, the range-of-motion and body position during active exercises can be suggested

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TABLE I
MUSCLES INCLUDED IN THE MUSCULOSKELETAL MODEL OF THE HIP JOINT

No.	Muscle	No.	Muscle
1	adductor brevis	15	gluteus minimus 3
2	adductor longus	16	iliacus
3	adductor magnus 1	17	pectineus
4	adductor magnus 2	18	piriformis
5	adductor magnus 3	19	psoas
6	gemelli inf. et sup.	20	quadratus femoris
7	gluteus maximus 1	21	biceps femoris long
8	gluteus maximus 2	22	gracilis
9	gluteus maximus 3	23	sartorius
10	gluteus medius 1	24	semimebranosus
11	gluteus medius 2	25	semitendinosus
12	gluteus medius 3	26	tensor fascie latae
13	gluteus minimus 1	27	rectus femoris
14	gluteus minimus 2		

that would prevent excessive loading of particular acetabular regions and displacement of fracture fragments.

II. METHODS

Biomechanical estimation of the hip joint loading was based on a mathematical model for computation of the hip joint reaction force and a previously developed model for computation of the contact stress distribution in the hip articular surface. The model for force assumes that the abduction exercise is performed slowly, i.e., the dynamic effects related to motion can be neglected and, therefore, the static calculation for given position of the leg is considered.

In computation of the hip joint reaction force $(\mathbf{R})$, the equilibrium equations of forces and torques acting on the lower leg are solved. The body weight is taken to be 800 N and the weight of the leg is taken to be 0.161 of the body weight [10]. The musculoskeletal geometry defining positions of proximal and distal muscle attachment points in neutral position and cross-sectional areas of the muscles is based on the work of Delp *et al.* [17]. Muscles attached over a large area are divided into separate units. Hence, the model includes 27 effectively active muscles of the hip (Table I, Fig. 1). The muscle attachments and the center of gravity correspond to the position of the lower extremity with extended knee and ankle in neutral position. We used a straight-line muscle model without taking into account the properties of the muscle-tendon unit. The force due to passive response of the muscle was neglected. Muscle activity required to maintain equilibrium in a given position of body is computed using the method of inverse dynamic optimization [18] proposed by Crowninshield *et al.* [19], [20]. Within this method, it is assumed that muscles are activated in a way that optimizes activity of the musculoskeletal system [18]. It has been suggested that the optimal muscle activation was found by minimization of the sum of muscle stresses cubed [18], [20] while the force of each muscle is constrained not to exceed the maximum isometrical force [17]. This optimization criterion is based on the experimentally determined nonlinear dependence between the muscle force and the endurance time of muscle contraction and on the idea that muscles are activated in a way that maximizes their endurance time [20]. The optimization criterion was justified by comparison of the resultant hip force calculation with the experimental measurements using an implanted instrumented endoprosthesis [16], [21]. After computation of the muscle forces, the components of the hip joint reaction force are determined from the force equilibrium equations for the leg.

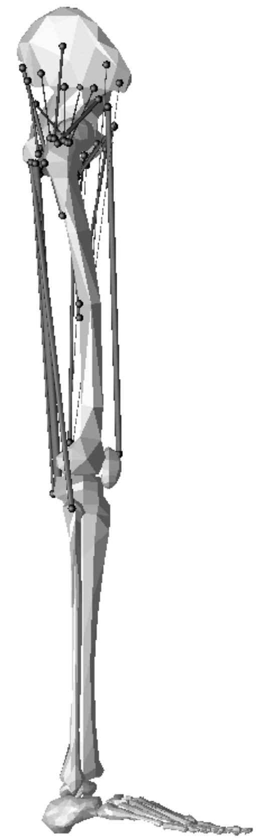

Fig. 1. Lateral view of the model of bone and muscles adapted from [17]. 27 muscle units crossing the hip were taken into account.

Each specific type of abduction exercise was modeled by rotation of the leg in the frontal plane of the body around the center of the femoral head (Fig. 1) while the pelvis was taken to be fixed in a laboratory coordinate system. The position of the leg during abduction exercise was defined by the abduction angle [Fig. 2(a)]. For a given abduction angle, the muscle geometry was adapted considering change in the attachment points of muscles on the bones of the leg. Simulation of various positions of the body (upright, supine, and side-lying) was performed by varying direction of the force of the leg weight. Supine abduction of unsupported straight leg without touching the ground and supine abduction of straight leg with 80 support were analyzed separately. The supporting force of the ground was considered to act in the center of the gravity of the leg.

Hip joint reaction force and the geometry of the acetabulum are the input parameters of the second model that determines hip contact stress [11]. Within this model it is assumed that the spherical femoral head and the hemispherical acetabulum are separated by a layer of cartilage. Upon loading, the femoral head is moved towards the acetabulum and the cartilage is squeezed.

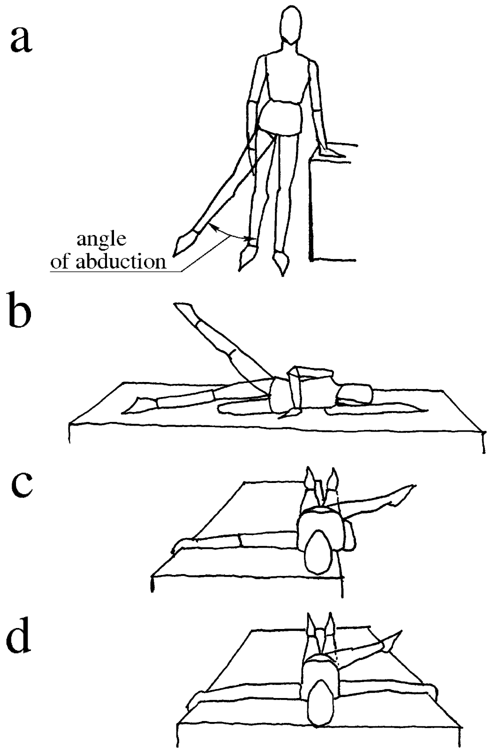

Fig. 2. Body position during (a) standing abduction, (b) sidelying abduction, (c) supported supine abduction, and (d) unsupported supine abduction.

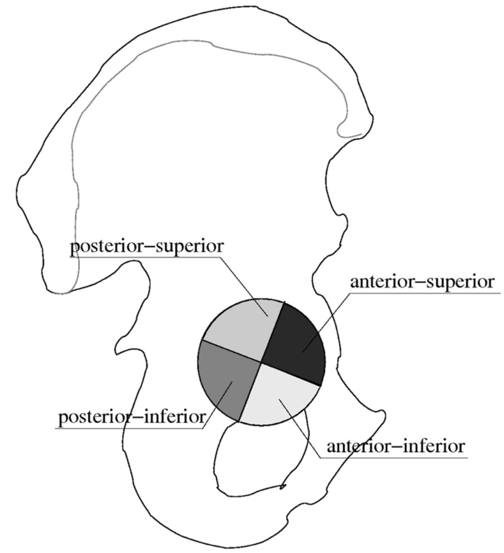

Fig. 3. Schematic view of pelvis with denoted quadrants of acetabulum.

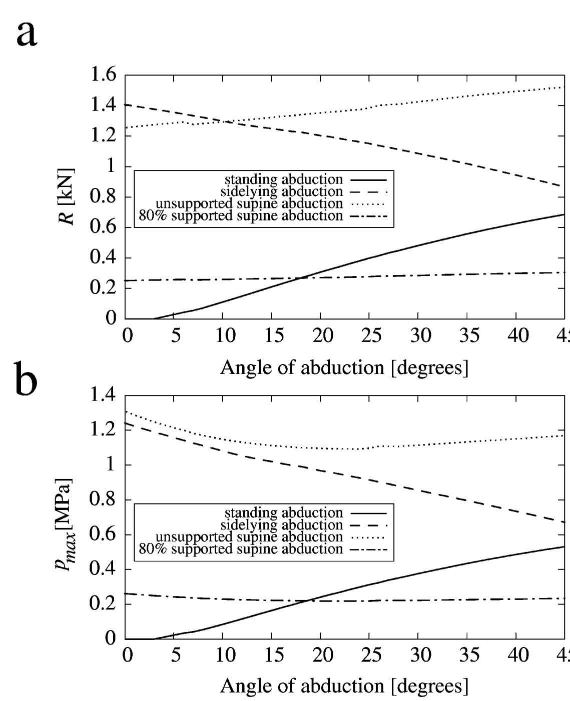

Fig. 4. (a) Magnitude of hip joint reaction force $\mathbf{R}$ and (b) the peak contact stress $p_{\max}$ during abduction exercises.

The calculation of the hip stress distribution is based on the assumption that the hip cartilage stress is proportional to the cartilage strain induced by hip loading [10]. The contact stress integrated over the articular surface is equal to the hip joint reaction force which is obtained by optimization method as described above. A system of three equations (one for each component of the force) is expressed by a single nonlinear algebraic equation which is solved numerically. The distribution of the hip contact stress was computed using the computer program HIPSTRESS [10]–[12].

Radius of acetabular surface was taken to be 25 mm, the lateral inclination and anteversion of acetabulum was taken to be 30° and 15°, respectively. In order to determine average loading of various acetabular regions, the acetabulum was divided into four quadrants with equal surface areas according to classification of Wasielewski *et al.* [22] (Fig. 3). The average resultant force loading the segment during whole range of exercise was determined by numerical integration of contact stresses acting on a given acetabular segment during whole range of exercise.

III. Results

The magnitudes of the hip joint reaction force $\mathbf{R}$ and the peak contact stress $p_{\max}$ during abduction exercises in different body positions are shown in Fig. 4(a) and (b), respectively. The loading of the acetabulum is the lowest in supported supine abduction and the highest in unsupported supine abduction. The force $\mathbf{R}$, as well as the peak contact stress $p_{\max}$ increase with the angle of abduction during standing and decrease during sidelying. When the supine abduction is performed, the hip joint reaction force $\mathbf{R}$ and the peak contact hip stress $p_{\max}$ vary only a little.

Comparative graph of the average hip joint reaction force loading in the four acetabular quadrants during abduction exercises in different body positions is presented in Fig. 5. In all four types of abduction exercise, the point of the peak contact stress is located in the posterior–superior quadrant of the acetabulum throughout the abduction arch and this quadrant also endures the highest average loading.

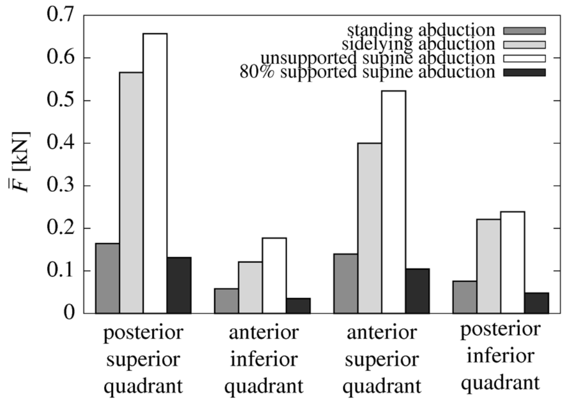

Fig. 5. Loading of the acetabular quadrants during abduction exercise. Value of $\bar{F}$ corresponds to the average force acting on a given acetabular quadrant during whole abduction exercise defined in Fig. 3.

IV. Discussion

We have found that in the neutral leg position the hip joint reaction force is high for side-lying or unsupported supine body position and low for upright standing. This can be explained by considering the equilibrium of the moments of the gravitational and muscular forces with respect to the center of rotation of the hip joint in different body positions. In standing and side-lying abduction, the equilibrium is maintained by the activity of abductors. In side-lying abduction higher abductor force is required to compensate the weight of the lower leg than in the standing abduction because of larger lever arm of the weight of the lower leg in former case. After increasing the angle of abduction in upright standing, the center of the gravity of the lower leg moves laterally, which further increases the gravitational moment. Hence, the counteracting muscle activity as well as the hip joint load must be increased. On the other hand, abduction of the lower leg in the side-lying exercise decreases the gravitational moment of the lower leg with respect to the hip and the hip load is decreased.

However, in the unsupported supine abduction, the leg has a tendency to extend and hence the activity of flexors is required. In the supine abduction flexors that are required to maintain this posture have smaller moment arms and thus demand high flexor forces. Therefore, the hip joint reaction force magnitude in unsupported supine position is considerably higher when compared to other body positions, however, ground support of the leg can proportionally reduce its magnitude.

The course of $p_{\max}$ follows the course of the hip joint reaction force for upright and side-lying abduction [Fig. 4(b)]. In contrast, abduction of the leg does not considerable change the peak contact hip stress in supine abduction and $p_{\max}$ remains almost constant throughout the abduction arch both with unsupported and supported leg. The average loading of the hip joint is the lowest in 80 supported supine abduction and the highest in unsupported supine abduction. In all four types of abduction exercise the point of the peak contact stress is located in the posterior–superior quadrant of the acetabulum, which is also reflected by the highest average hip joint reaction force loading in the posterior–superior quadrant of the acetabulum (Fig. 5) and considerably lower loading of the anterior–inferior and the posterior–inferior quadrants. Upright abduction produces considerably lower average loading magnitudes in all four quadrants while side-lying abduction mainly differs from the unsupported supine abduction in lower average loading of the anterior–inferior acetabular quadrant.

Computed values of hip joint reaction force and peak contact hip stress reported in our paper are of the same order of magnitude as the ones performed in nonweight-bearing exercises measured *in vivo* [7], [8]. Peak stress in direct measurements was also located in the posterior–superior acetabular quadrant, which is the case also in our study. Direct measurements of peak contact stress in supine abduction were found to be 2.8 MPa and 3.8 MPa *in vivo* [7], [8] versus 1.3 MPa in our study. The reports do not specifically mention the amount of vertical leg support in supine adduction, but considering the fast velocities it could be inferred that the abduction was unsupported. Contrary to our findings and to clinical guidelines, the only *in vivo* study that compared abduction in different body positions has found quite a different rank order of the peak contact hip stress values with 8.9 MPa in standing hip abduction, 5.6 MPa in side-lying hip abduction and 2.8 in supine hip abduction [7]. It should be noted, however, that these *in vivo* measurements were performed with angular velocities above 30°/s and, therefore, also include the dynamic component of loading. Furthermore, a change from a side-lying body position to an upright position considerably reduces the moment arm of the leg weight but it does not substantially influence the moment arms of individual muscles. In static conditions, a reduced moment arm of the leg weight in the upright position reduces the calculated muscle forces and consecutively lowers the hip load, as shown in Fig. 4. However, in dynamic motion, a smaller moment arm of the leg weight in the upright position would facilitate an initial acceleration of the leg that later requires higher muscle strength to stop the movement at maximal abduction. Comparison between dynamic measurements and static computations therefore indicates that at very slow motion the upright abduction causes lower contact hip stresses than side-lying abduction, but this may be reversed in maximum abduction at high angular velocity. One of the reasons for performing only high-speed measurements may have been the measurement error of approximately 0.2 MPa that was not accurate enough for slow nonweight bearing measurements with magnitudes below 1 MPa. When direct measurements of contact hip stress were compared with simultaneous hip stress estimations through kinematic measurements, it was found that direct measurements of the same activities yield considerably higher contact stress than inverse Newtonian analyses [23]. This effect has been attributed to cocontraction of muscles that is especially apparent in relatively slow, controlled movements [23] and this may to some extent explain the discrepancy between our results and results obtained by direct dynamic measurements.

The limitations of the direct stress measurement method [7], [8] include the facts that the sensors measure the cartilage-on-metal surface and not the cartilage-on-cartilage surface, that the metal prosthesis in contact with natural acetabulum may differ from physiologic morphology of the hip, that sensors were located on the femoral head surface while the values of stress on the acetabular joint surface were estimated from the kinematic data. Further, hip pressure measurements of abduction exercises were performed in one patient only.

On the other hand, our method is limited by the model assumptions. In the calculation of the acetabular loading, it was assumed that the pelvis is fixed. Small rotations of the pelvis are not likely to influence the hip joint reaction force. However, if the rotations of the pelvis were large, this would represent a significant limitation of the study. The muscles were considered as "active springs" which are able to generate force in order to maintain body position [15], [18]. The passive forces generated by the muscle-tendon unit were not taken into account. To better describe the properties of the muscle, a forward dynamics optimization including activation dynamics and musculo-tendon contraction dynamics could be used [24]. The static optimization method used in this study to compute muscle forces cannot predict muscle cocontraction [18]. To improve the description of the muscle forces during exercises, a dynamic optimization approach that takes into account dynamic properties of neuromusculoskeletal system could be used [24]. Also, within the static approaches, there are differences according to the choice of the optimization criterion [18]. However, comparison of measurements and calculations of the hip joint reaction force showed that the type of the optimization criterion employed does not significantly influence the calculated hip joint reaction force [25].

We conclude that absolute values of the hip joint reaction force and the peak contact hip stress are highest in unsupported supine abduction, slightly lower in side-lying abduction and lowest in upright abduction. In all body positions, the hip joint reactive force and the peak contact hip stress are the highest in posterior–superior quadrant of acetabulum (acetabular dome, weight bearing area), followed by anterior–superior quadrant (anterior wall and column), posterior–inferior quadrant (posterior wall and column), and finally anterior–inferior quadrant. Our results are in agreement with the clinical guidelines as they indicate that upright abduction should be commenced first [6]. Supine abduction in initial rehabilitation phases should be recommended with ground support (on the bed) without excessive vertical leg lifting. In maximal upright abduction the magnitude of forces and stresses approaches the values of side-lying abduction and therefore patients should be instructed against excessive abduction amplitudes in the initial upright rehabilitation.

The spatial distribution of the acetabular loading indicates that early physical therapy should be planned according to the preoperative location of the fracture line in the acetabulum. If the fracture fragments lie in the anterior–inferior quadrant of the acetabulum, the resultant force acting on this region of the acetabulum is low (Fig. 5) and no restriction of active abduction is needed while planning early physical therapy. On the other hand, when the acetabular fracture line occurs in the posterior–superior quadrant of the acetabulum, high resultant force on this part of the acetabulum in active abduction (Fig. 5) may cause dislocation of the acetabular fragments which were previously fixed by the surgeon. Early rehabilitation phase in this case should therefore be restricted to active abduction only (where the acetabular load is not too high), i.e., to supine abduction throughout the whole range of motion and/or upright exercise with maximum 20° of abduction (Fig. 4). Based on the presented results, we suggest that detailed calculations of spatial distribution of the hip joint contact stress are required before starting rehabilitation procedure as to individually design the rehabilitation procedure for a given patient. In the planning, the spatial position of the fracture lines and dislocations of the acetabular fragments should be taken into account.

Our results complement the results of direct measurements of stress during exercises and the experience-based exercise protocols in elucidating the mechanical impacts on the rehabilitation.

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