Acetabular forces and contact stresses in active abduction rehabilitation

H. Debevec[1], A. Kristan[2], B. Mavcic[1], M. Cimerman[2], M. Tonin[2], V. Kralj-Iglic[3], and M. Daniel[1,4]

[1]Laboratory of Physics, Faculty of Electrical Engineering, University of Ljubljana, Slovenia
[2]Department of Traumatology, University Medical Center Ljubljana, Slovenia
[3]Laboratory of Clinical Biophysics, Faculty of Medicine, University of Ljubljana, Slovenia
[4]Laboratory of Biomechanics, Faculty of Mechanical Engineering, Czech Technical University in Prague, Czech Republic

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 non-weight-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 non-weight-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). The results are in agreement with the clinical guidelines as they indicate that upright abduction should be commenced first.**

Keywords— **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 non-weight-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 5-14 days postoperatively. Partial weight-bearing with stepwise progression usually starts 6 weeks postoperatively and full weight bearing is eventually allowed at 10 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].

Due to technical complexity and invasiveness of direct contact stress distribution measurement, various mathematical models for calculation of the hip joint loading force and contact stress distribution in the hip joint have been proposed [9]-[16]. 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].

The aim of the paper is to compare acetabular loading in non-weight-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 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.

T. Jarm, P. Kramar, A. Županič (Eds.): Medicon 2007, IFMBE Proceedings 16, pp. 915–918, 2007

Table 1 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		

In computation of the hip joint reaction force (***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 (Tab. 1). 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].

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. 2) 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. 2a).. Supine abduction of unsupported straight leg without touching the ground and supine abduction of straight leg with 80% of the weight of the leg support were analyzed separately. The supporting force of the ground was considered to act in the center of the gravity of the leg.

The distribution of the hip contact stress for given position of the leg was computed using the computer program HIP-STRESS [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 degrees, respectively.

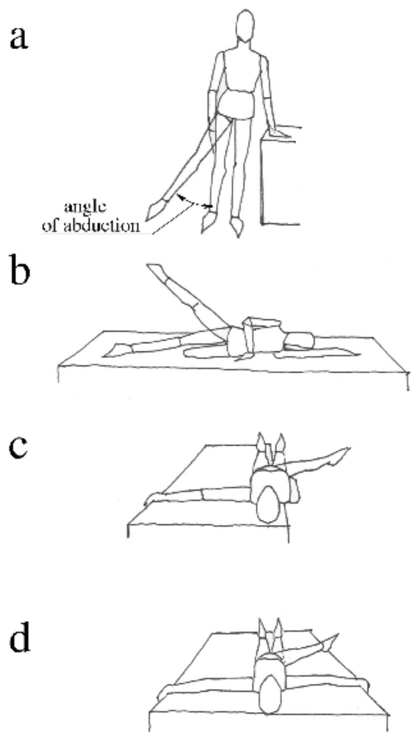

Fig. 1 Body position during standing abduction (a), sidelying abduction (b), supported supine abduction (c) and unsupported supine abduction (d).

III. Results

The magnitudes of the hip joint reaction force ***R*** and the peak contact stress p_{max} during abduction exercises in different body positions are shown in Fig. 2a and Fig. 2b respectively. The loading of the acetabulum is the lowest in supported supine abduction and the highest in unsupported supine abduction. The force ***R*** as well as the peak contact stress p_{max} increase with the angle of abduction during standing and decrease during side-lying. When the supine abduction is performed, the hip joint reaction force ***R*** and the peak contact hip stress p_{max} vary only a little.

Fig. 2 Magnitude of hip joint reaction force ***R*** (a) and the peak contact stress p_{max} (b) during abduction exercises

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. 2b). 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.

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 non-weight-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. 2. 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 non-weight bearing measurements with magnitudes below 1 MPa. When direct measurements of contact hip stress were compared with simultaneous hip stress estimations through kinematics measurements, it was found that direct measurements of the same activities yield considerably higher contact stress than inverse Newtonian analyses [20]. This effect has been attributed to co-contraction of muscles that is especially apparent in relatively slow, controlled movements [20] and this may to some extent explain the discrepancy between our results and results obtained by direct dynamic measurements.

V. Conclusion

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. 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. 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.

Acknowlegment

The research is supported by the Czech Ministry of Education project No. MSM 6840770012 and by the Slovenian ARRS Projects No. P2-232J3-619 an BI-CZ/07-08-006 and BI-S/05-07-002.

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Author: Dr. Matej DANIEL

Institute: Laboratory of Biomechanics, Department of Mechanics, Biomechanics and Mechatronics, Faculty of Mechanical Engineering, Czech Technical University in PRague
Street: Technicka 4
City: Prague
Country: Czech Republic
Email: matej.daniel@fs.cvut.cz