Chapter 10

CONTACT HIP STRESS MEASUREMENTS IN ORTHOPAEDIC CLINICAL PRACTICE

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ABSTRACT

There exist several invasive and noninvasive methods to measure the contact hip stress but due to their complexity only few have so far been tested in clinical trials with large numbers of participating subjects. Consequently, the use of contact hip stress measurements in orthopaedic clinical practice is still in its experimental phase.

Biomechanical studies of human hips based on the analysis of 2-D pelvic radiographs have turned out to be a reasonable compromise between the measurement accuracy and the feasibility in clinical setting. Clinical studies have shown significantly higher values of hip stress in adult dysplastic hips when compared to normal hips. It has been found that the cumulative hip stress independently predicts the WOMAC score after 29-years of follow up in dysplastic hips and does so better than morphological radiographic parameters of hip dysplasia or the resultant hip force alone. The preoperative value of the contact hip stress and the magnitude of its operative correction have been found predictors of the long term success of the Bernese periacetabular osteotomy. Elevated shear stress in femoral neck, but not elevated hip contact stress, has been found to be a risk factor for slipping of the capital femoral epiphysis. A statistically significant correlation between the contact hip stress and the age at the total arthroplasty has been shown in a group of hips with idiopathic hip osteoarthritis.

Through advances in 3-D imaging with MRI and CAT scan, visualisation of the femoral head coverage and pelvic muscle attachment points has improved considerably. However, the need to supplement the morphological hip status with biomechanical analysis remains. The current trend is to combine the kinetic gait measurements of the resultant hip force with 3-D imaging of the hip weight-bearing surface in order to better estimate the contact hip stress for a given activity/body position. The added value of such measurements over 2-D pelvic radiograph analysis has not been established yet in clinical trials.

INTRODUCTION

Hip stress depends on the magnitude/direction of the resultant hip force, the size of the weight-bearing surface and the stress distribution across this surface. The interest of orthopaedic surgeons in possible relationship between the contact stress and the pathological cartilage conditions has been particularly large for weight-bearing joints with relatively simple kinematics such as the hip joint. A review of the contact hip stress measurement methods has been published by Brand et al. [1]. While many methods have been developed only the noninvasive methods could be used in clinical practice. The aim of this paper is to review the current trends in the clinical use of methods of contact hip stress assessment.

NONINVASIVE DETERMINATION OF THE RESULTANT HIP FORCE

The resultant hip force can be estimated in static or dynamic conditions. Static biomechanical models estimate the resultant hip force for a given body position by solving the static equations for the equilibria of forces and torques [2-4]. The muscles in the static biomechanical models are assumed to be force generators with fixed coordinates. Interindividual variability of muscle attachment points can be achieved by linear scaling of the pelvic configuration in plain anterior-posterior radiographs. In this setting, the standing anterior-posterior pelvic radiographs are assumed to represent the body position of one-legged stance. One of the static biomechanical models applied to a large number of subjects in clinical studies is part of the HIPSTRESS method [4-7]. In this model the one-legged stance is considered to be the representative position for slow gait as the most frequent activity in everyday life [8]. In the one-legged stance, abductor activity is needed to maintain balance on the load bearing leg. The following radiographic parameters are measured on the anterior-posterior radiographs manually [7]: the interhip distance, the pelvic height, the pelvic width, the coordinates of the insertion point of abductors on the greater trochanter (Figure 1). The three-dimensional reference coordinates of the muscle attachment points are taken from a prototype specimen and they are adjusted by linear scaling with regard to the radiographic pelvic parameters for each individual hip. The solution of the vector equations for the equilibria of forces and torques yields the three components of the resultant hip force and the tensions in the abductor muscles.

Noninvasive estimation of the resultant hip force during dynamic activities in different body positions requires the use of the dynamic biomechanical models.
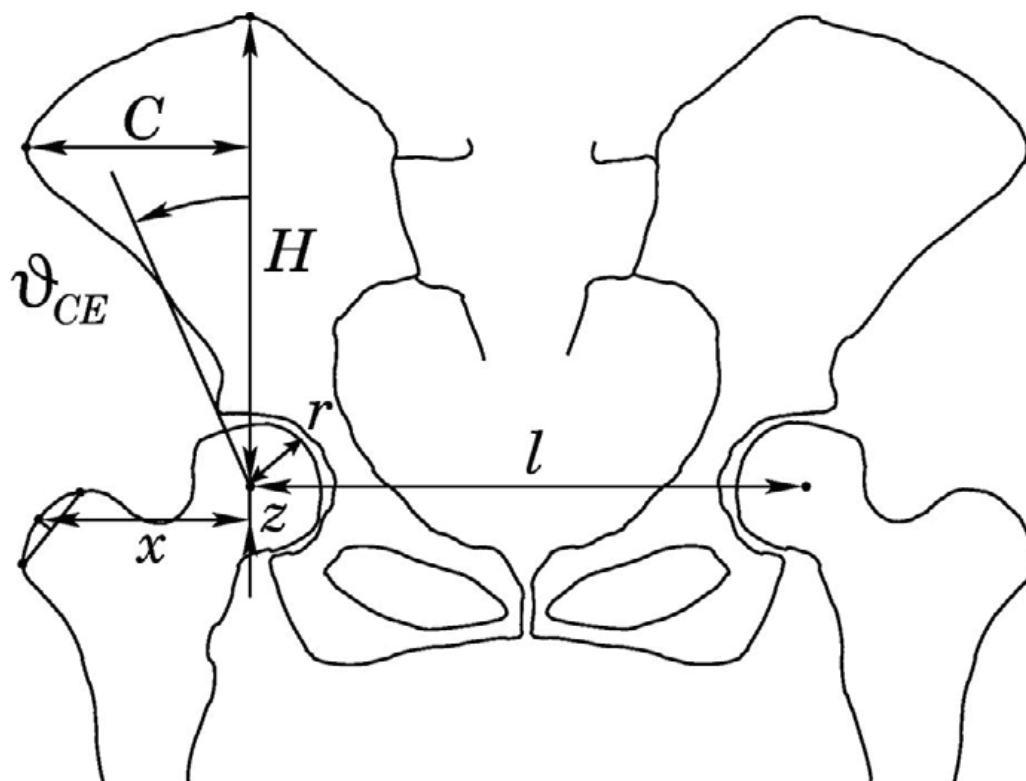
Figure 1. In the HIPSTRESS method of the resultant hip force computation, the following radiographic parameters are measured from the anterior-posterior radiographs manually: the interhip distance l, the pelvic height H, the pelvic width C, the coordinates of the insertion point of abductors on the greater trochanter (x, z). The average of acetabular and femoral head radius r and the Wiberg lateral center-edge angle ϑ_{CE} are used subsequently to estimate the weight-bearing surface and the contact hip stress distribution.

They are based on laboratory measurements with contact force plates, kinematic data of body segment motion and subsequent inverse dynamic analysis of the moving segments [9, 10]. Such approach necessarily includes complex muscle models for several body segments and an appropriate optimization technique to solve the model equations with reduntant muscle forces.

An example of a dynamic biomechanical model based on gait analysis has been developed for use in pre-clinical testing [10]. A computer model of the bones and muscles of the human lower extremities (CT-data, Visible Human, NLM, USA) was scaled to match the anatomies of four THR patients with telemeterized femoral components. Gait analysis data for walking and stair climbing were determined simultaneously with in vivo hip contact forces.

The gait data and the individual musculoskeletal models were then used to calculate the intersegmental resultant joint forces at the ankle, hip and knee joint, as well as the muscle and joint contact forces throughout each gait cycle. The calculated hip joint contact forces were finally validated against the contact forces measured in vivo for both activities.

The latest trend in the noninvasive hip force estimation of individual patients is the polyamid reversed engineering model based on a computed tomography dataset [11]. Pelvis and femur of an individual patient are reproduced in polyamid by selective laser sintering. Hip joint forces can be measured using an experimental setup in which an industrial robot is exerting hip joint forces and moments representing one-legged stance. Hip extensor and

abductor actuator forces are measured which counterbalanced the joint moments. The resulting bony model is geometrically accurate, but it does not take into account the joint incongruencies due to the neglected cartilaginous structures in the model [11].

NONINVASIVE DETERMINATION OF THE HIP WEIGHT-BEARING SURFACE

In the past when three-dimensional imaging was not yet widely available in the clinical setting, the femoral head radius and the anterior and/or lateral coverage were the main parameters to be used in the biomechanical models for the estimation of the weight-bearing surface. The basic method used for the noninvasive determination of the hip weight-bearing has been plain radiography in the anterior-posterior projection and the false profile view. Many radiographic indices have been described in order to evaluate the femoral head coverage by the acetabulum and to indirectly estimate the weight-bearing surface. The most commonly used parameters include the Wiberg lateral centre-edge angle (CE) [12], the vertical anterior centre-edge angle of Lequesne and de Sèze (VCA) [13], the femoral head extrusion index and the acetabular index [14]. Thus, it was assumed femoral head is a perfect spherical surface with variable lateral and/or anterior coverage by the acetabulum. Soon it became clear that in most physiological conditions the weight-bearing surface is *not* equal in size to the entire articular surface. Variability in some parts of the joint (e.g. lateral coverage) may greatly influence the hip loading while the medial part of the joint bears only small loads [6]. In the HIPSTRESS method [4-7], the weight-bearing surface is assumed to make part of a perfect articular sphere limited on the lateral side by the coverage of the acetabulum. Its medial border depends on the location of the pole of stress distribution. Thus, the size of the weight bearing surface is computed from the average of acetabular and femoral head radius and the Wiberg lateral center-edge angle (Figure 1). The biomechanical role of the horseshoe geometry of the acetabular cartilage has also described using a three-dimensional mathematical model. It has been shown the characteristic horseshoe shape of the articular cartilage in the human acetabulum optimizes the contact stress distribution in the hip joint [15].

Recent advances in the three-dimensional imaging technologies (CAT scan, MRI) have greatly improved the ability to estimate the hip articular surface [16]. There exist contact hip stress studies where the weight-bearing surface was determined with a very precise MRI imaging albeit the resultant hip force magnitude/direction was assumed to be constant for all 82 hips examined [17]. MRI is particularly valuable for the non-invasive estimation of the articular cartilage and the labrum. The disadvantage of classic MRI imaging is that (unlike plain radiography) it cannot be performed with the patient standing and thus the cartilage deformation in different weight-bearing positions cannot be estimated directly. Although the three-dimensional image of the *articular* surface is much more precise it does not suffice to estimate the *weight-bearing* surface directly and again biomechanical models must be used to estimate the importance of different parts of the joint in hip loading. With the development of MRI and hip arthroscopy, some authors have suggested that impingement due to local irregularities near the weight-bearing surface may be more important than the contact hip stress on the weight-bearing surface itself. This has lead to the theory of femoroacetabular

impingement, emphasizing the importance of joint incongruencies and the need to identify regions of the femoral head surface with locally increased hip stress values due to the cam/pincer type of impingement [18].

ESTIMATION OF THE CONTACT HIP STRESS IN THE CLINICAL SETTING

Theoretically, any noninvasive method of the resultant hip force determination could be combined with any noninvasive method of the hip weight-bearing surface determination (Figure 2). In the initial research phases the studies mostly focused on the relationship between the biomechanical parameters based on the analysis of individual cases [1]. In such setting it was possible to apply very precise methods, although costly in terms of time and financial resources. However, when the methods were to be applied in clinical studies with larger numbers of patients (over 100) the aspect of timely/financial feasibility turned out to be one of the critical factors. Thus, there exist only few studies of contact hip stress performed in the clinical setting.
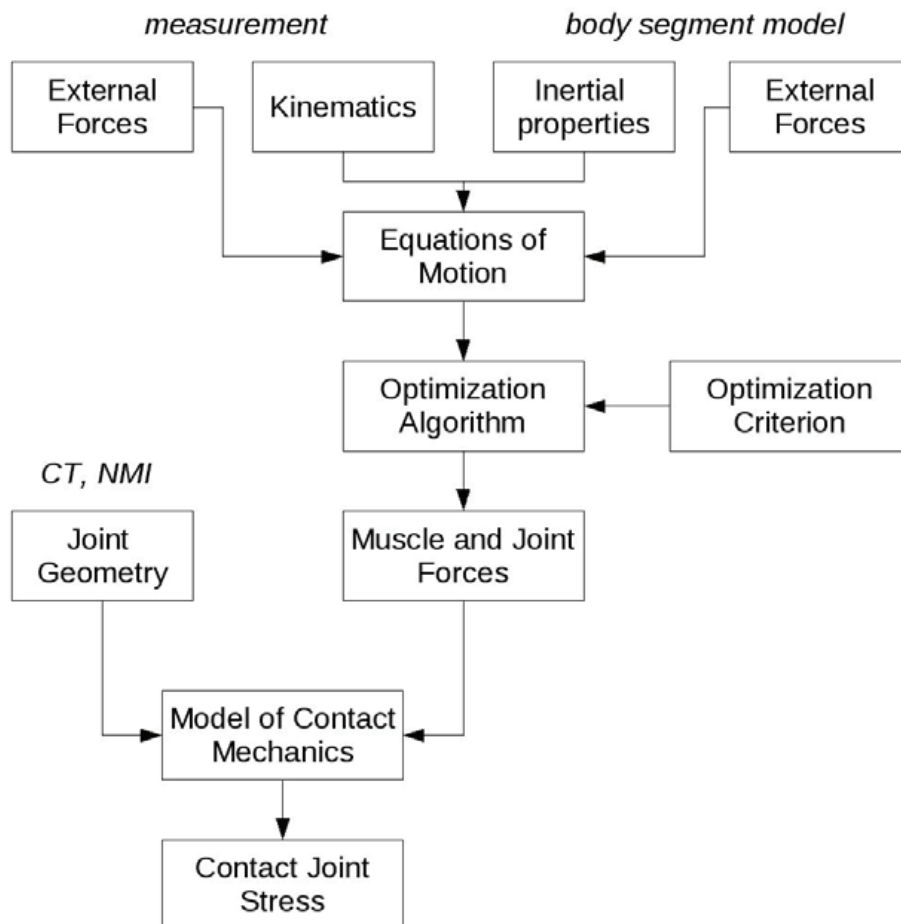

Figure 2. The general algorithm of contact hip stress computation includes the determination of the resultant hip force (combining measurements with the body segment model) and assessment of the weight-bearing surface with morphological imaging of the joint geometry.

As Brand et al. [1] pointed out reports on contact hip stress in the literature may describe different stress distributions. Some authors have estimated spatially averaged values of the contact hip stress (average stress = net force / total weight-bearing surface). Other authors report values of the peak contact stress, i.e. the maximal contact stress value on the weight-bearing surface. Several mathematical approaches to estimate the weight-bearing surface (and consecutively the contact stress distribution) have been proposed [1, 7]. Legal developed a practical method for calculation of the contact hip joint stress for a specific case based on the frontal plane equilibrium force analysis [3]. According to this method, the resultant hip joint force is calculated in static one-legged stance body position assuming one effective abductor muscle with effective attachment point on the greater trochanter and certain inclination towards the horizontal plane. Hadley et al. [19] followed the approach outlined by Legal where the calculation of the hip contact joint stress distribution is restricted to the simplest case of uniform contact stress distribution. One of the few methods applied to more than two hundred patients with different types of hip pathology is the HIPSTRESS method [4-7], developed by the authors of this chapter (Figure 3). The major advantage of the HIPSTRESS method for evaluation of the stress distribution in the hip joint is that it takes into account the non-uniform stress distribution over the weight-bearing surface. This could be of special importance as the gradient of contact stress distribution seems to be even more important than the magnitude of the contact stress [1]. In the HIPSTRESS model the weight-bearing surface is not fixed in advance. The hip geometry affects the resultant hip force and the size and the shape of the weight-bearing surface in a self-consistent manner [6]. These theoretical predictions are based on the assumption of Hooke's law, where the cartilage is described macroscopically as a homogeneous continuum and linear elastic solid. This means that the radial stress in the articular surface is taken to be proportional to the radial strain in the cartilage layer [2].

In most biomechanical models the femoral head is considered to be a perfect sphere. This condition may correspond well to the hips with spherical femoral heads, but dysplastic hips in advanced stages of osteoarthritis have incongruent femoral heads with reduced joint space width, which leads to radiographic overestimation of the femoral head radius and the Wiberg lateral center-edge angle and therefore results in underestimation of the peak contact stress [7]. It is therefore reasonable to expect that incorporation of joint congruity assessment in the biomechanical model would result in even higher values of peak contact hip stress in dysplastic hips and therefore improve the predictive value of these biomechanical parameters. Further, conclusions of biomechanical analyses may be misleading if the relative mathematical importance of individual biomechanical parameters for the contact hip stress computation is assumed to be equally important in explaining contact hip stress variability of the general population. The squared value of the femoral head radius, for example, was found to have direct inverse correlation to the contact hip stress; yet a clinical study proved the femoral head radius had small variability between different individuals and did not account for large contact hip stress differences between normal and dysplastic hips [7].

When critically evaluating the results we should also consider that the statistical significance of biomechanical computations from anterior-posterior radiographs is limited by the data dispersion caused by the error in magnification. It was found that the magnification may vary substantially, but the distribution of the magnification is normal [20].
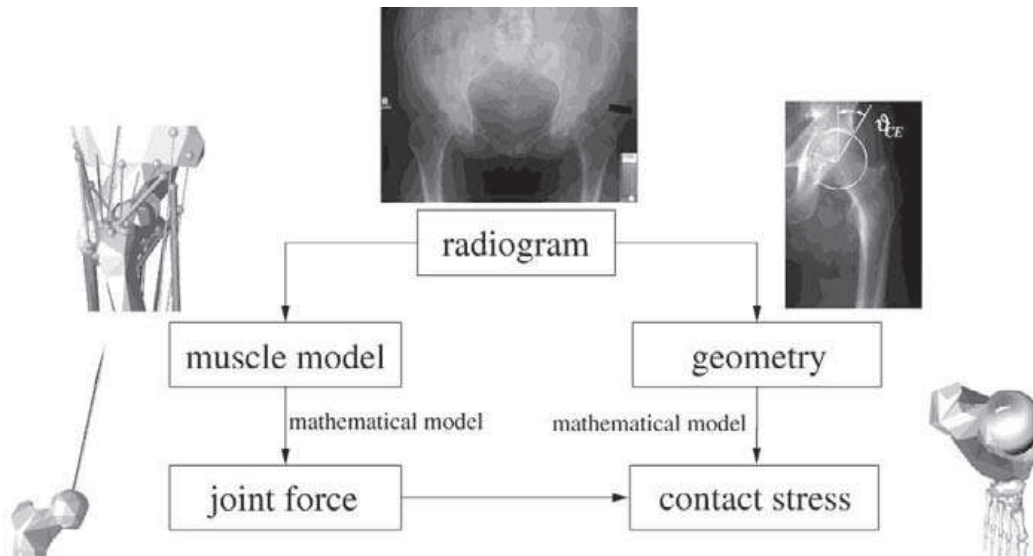
Figure 3. The HIPSTRESS method is based on the general algorithm of contact hip stress computation, as shown in Figure 2: the resultant hip force is computed with a static muscle model of one-legged stance and the joint geometry is assessed by measuring the femoral head radius r and the Wiberg lateral center-edge angle ϑ_{CE}.

Therefore in population studies the magnification should not affect the relative difference between the average or median values but rather increase the data dispersion and decrease the statistical significance of the difference between the considered populations. Also, in determining peak stress, not all the geometrical parameters are equally important as the functional relations between them are nonlinear. It ensues from the mathematical model that stress first depends on the radius of the femoral head, then on the interhip distance, subsequently on the lateral extension of the effective attachment point on the greater trochanter and at last on the pelvic height and width. For example, if the error made in determination of the pelvic height is about 15 per cent, this only yields a 2 per cent error in the peak stress [21].

Three-dimensional imaging techniques have further improved the assessment of the contact hip stress distribution. Hip morphology data can be applied to the finite element models in order to simulate the hip contact stress distribution [22]. Patient-specific, non-linear, contact finite element models of the hip, constructed from computed tomography arthrograms using a custom-built meshing program, were subjected to normal gait cycle loads.

There were significant differences found between the normal control and the asymptomatic hips, and a trend towards significance between the asymptomatic and symptomatic hips of patients afflicted with developmental dysplasia of the hip. The magnitudes of peak cumulative contact pressure differed between apposed articular surfaces. Bone irregularities caused localized pressure elevations and an upward trend between chronic over-pressure exposure and increasing severity of hip dysplasia [22]. However, the method has not been applied to larger numbers of patients in the clinical studies.

Contact Hip Stress in the Assessment of Developmental Hip Dysplasia and Hip Osteoarthritis

Contact hip stress measurements have turned out to be particularly useful in the evaluation of the developmental dysplasia of the hip [1, 7, 23]. With advances in knowledge, it has been established that the developmental dysplasia of the hip (previously called "congenital hip dislocation") is not a uniform clinical entity but rather a broad continuous spectrum ranging from asymptomatic dysplastic acetabula to dislocated hips. Because insufficient acetabular coverage implies the usual hip loads are distributed on a smaller weight-bearing surface compared with normal hips, biomechanical research has focused on estimation of the contact stress in the hip rather than simply morphologic evaluation [24]. In hips with more severe hip dysplasia, several epidemiologic cross-sectional surveys suggested increased incidence of hip osteoarthritis [25, 26]. Furthermore, hip dysplasia was found to be one of the independent risk factors for hip osteoarthritis in addition to age and body mass index [27]. A higher incidence of hip osteoarthritis together with higher average values of contact stress in dysplastic hips have led to the hypothesis that contact hip stress may be one of the key parameters involved in cartilage degeneration [1, 2]. Direct clinical assessment of the predictive value of contact hip stress was reported in two clinical studies of patients with hip dysplasia who were treated with closed reduction and followed up to the average age of 31 years. These authors concluded increased cumulative stress exposure bears higher risk for an unfavorable clinical outcome or osteonecrosis [19, 28]. Although the most severe cases of hip dysplasia are clearly associated with early degeneration, reports on patients with borderline dysplastic hips have been more controversial. In a study with 10-year follow-up of age-matched patients with residual dysplasia without subluxation and normal hips, the authors reported no differences in the reduction of the joint space width or in self-reported hip pain [29]. A recent systematic review found little evidence for a relationship between hip dysplasia and late hip osteoarthritis discovered in patients older than 50 years of age [26]. However, the authors recognized the relationship for the subsequent risk of osteoarthritis in persons diagnosed with dysplasia at a young age compared with the subsequent risk of young patients with osteoarthritis without dysplasia. Some authors have speculated most of the cases of "idiopathic" hip osteoarthritis in fact arise as a result of subtle abnormalities in the anatomic structure of the hip that remained unrecognized during childhood and adolescence and only began to cause clinical symptoms in old age [30].

A clinical study was conducted on the role of the contact hip stress for the development of osteoarthritis in initially asymptomatic human hips, either dysplastic or normal [23]. In the study nonoperatively treated nonsubluxated hips with developmental dysplasia without symptoms at skeletal maturity were identified and compared to adult hips without any disease. Peak contact hip stress was computed with the HIPSTRESS method [4-7] using anterior-posterior pelvic radiographs at skeletal maturity. This method enabled computation of the peak contact hip stress for every individual hip from known values of the body weight, the femoral head radius, the Wiberg center-edge angle, the magnitude of the resultant hip force and the inclination of the resultant hip force with respect to the vertical. The cumulative contact hip stress was determined by multiplying the peak contact hip stress by age at follow-up. The WOMAC scores [31] and radiographic indices of osteoarthritis at a minimum follow-up of 20 years were compared. Dysplastic hips had higher mean peak contact hip stress and

higher mean cumulative contact hip stress than normal hips. Mean WOMAC scores and percentage of asymptomatic hips in the study group at the average age of 51 years were equal to the control group at the average age of 68 years. After adjusting for gender and age, the cumulative contact hip stress, Wiberg center-edge angle, body mass index, but not the peak contact hip stress, independently predicted the final WOMAC score in dysplastic hips. Cumulative contact hip stress predicted early hip osteoarthritis better than the Wiberg center-edge angle [23].

The relative maximum hip joint contact stress was found to be higher in healthy women than in healthy men [32]. As women have a higher incidence of the hip osteoarthritis [33], such epidemiological results support the hypothesis that the increased contact hip stress can be one of the risk factors of hip osteoarthritis [32]. This hypothesis was tested by two studies of standard anterior-posterior pelvic radiographs with no or subtle radiological signs of hip osteoarthritis of patients, who underwent hip arthroplasty for primary osteoarthritis years later [34, 35]. In the population of subjects with unilateral osteoarthritis, average peak contact hip stress was significantly higher in hips with arthroplasty than in contralateral hips. In the population of subjects with bilateral osteoarthritis, average peak contact hip stress was significantly higher in hips with earlier arthroplasty than in contralateral hips [34]. Younger age at hip arthroplasty was associated with higher body weight, higher peak contact hip stress normalized to the body weight, higher resultant hip force and larger peak contact hip stress, but not with body mass index [35].

CONTACT HIP STRESS IN PREOPERATIVE PLANNING OF ORTHOPAEDIC PROCEDURES

Osteoarthritis can develop as an idiopathic disease, however, subtle abnormalities could be detected in the hip joint prior to the development of symptoms. Origins of the development of osteoarthritis are ascribed to metabolic resorption of cartilage and/or deformations of anatomical structures. The deviations in the size, shape, mutual proportions or orientation of the acetabulum and/or the femoral head occur frequently. Such deviations are described as hip dysplasia [19]. Although mostly a pediatric problem, hip dysplasia can persist in untreated or unsuccessfully treated cases into adulthood as residual hip dysplasia and may eventually lead to degeneration of the cartilage, presumably due to the pathologically increased stress within the joint. Therefore hip dysplasia represents an important indication for operative procedures that should reduce or redistribute the hip joint stress, thereby stopping or slowing down the pathological processes in the hip cartilage. The hypothesis of secondary osteoarthritis due to decreased femoral head coverage and consecutively higher contact hip stress [28, 36] has led to the invention of surgical procedures of acetabular reorientation in order to prevent the progression of osteoarthritis. Several operative procedures to increase femoral head coverage have been described [37-40] with limited possibilities of correction. Because of such limitations the 'Bernese' periacetabular osteotomy was developed in 1984 and published in 1988 [41]. The procedure has become widely used because it allows optimal correction with minimal exposure and low complication rate, although the operation is technically demanding and its learning curve is long [42]. Ideally this procedure would be indicated in younger adults with concentric hip motion, spherical joint surfaces and no secondary osteoarthritis, but

many patients who present with symptomatic hip dysplasia do not meet these criteria [43]. Many attempts have been made to use additional clinical, radiographic or biomechanical factors to carefully select patients that would benefit from the joint preserving surgery [44]. Biomechanical parameters have also been used in the preoperative planning in order to determine the optimal degree of correction in different planes and to achieve the subtle balance between improvement of femoral coverage and restriction of range of motion [5, 16, 17].

In patients who underwent the Bernese periacetabular osteotomy there have been many retrospective analyses published, but only two of them have so far included the pre- and postoperative computations of the contact hip stress [45, 46]. Both studies used the HIPSTRESS method applied to 170 hips altogether. The periacetabular osteotomy was shown to improve the lateral and anterior coverage of the femoral head and accordingly to reduce the normalized peak contact stress in all studied hips. Biomechanical results were consistent with previous studies that have shown contact pressure elevation in dysplastic hips [7, 19] and contact pressure reduction by simulated acetabular reorientation [17]. The postoperative values were in most cases reduced to the level observed in healthy adult hips [47]. Although the Bernese periacetabular osteotomy allowed medialization of the hip joint in addition to improvement of lateral and anterior coverage [48], some studied hips showed lateralization of the center of rotation on the postoperative AP radiographs. Accordingly, the magnitude of the resultant hip force increased in some stratification groups, which is consistent with theoretical predictions of the relationship between the center of rotation and the resultant hip force [5]. These clinical results therefore confirmed theoretical findings [5] that the resultant hip force itself is not a sufficient biomechanical parameter for preoperative planning and that rotational osteotomy can effectively decrease the contact hip stress although the resultant hip force may be slightly increased. The conclusions of this study have shown further reduction of the contact hip stress is only possible through medialization of the femoral head and not through excessive lateral coverage. Accordingly, the procedure was modified with curved periacetabular osteotomy that enables better medialization of the femoral head [46].

Contact hip stress has been tested as a risk factor for slipped capital femoral epiphysis, but with less success [49]. Many hypotheses about the etiology of slippage have been examined, yet the underlying mechanisms have not yet been fully elucidated. Hips contralateral to the slipped ones were compared to age- and gender-matched healthy hips with respect to the shear stress and the contact hip stress. The characteristics of individual hips were incorporated by means of geometrical parameters determined from standard anterior-posterior radiographs. Shear stress was calculated by using a mathematical model where the femoral neck was considered to function as an elastic rod. Contact hip stress was calculated by the HIPSTRESS method [4-7]. Hips contralateral to the slipped ones had higher average shear stress in the femoral neck and more vertically inclined physeal angle in comparison to healthy hips, shear stress in the contralateral hips to the slipped ones remained significantly higher even when normalized to the body weight. However, there was no significant difference in the average contact hip stress [49].

The long-term effect of contact hip stress on the clinical outcome was also studied in the hips operated on by various intertrochanteric osteotomies due to avascular necrosis of the femoral head [50]. The hypothesis stated the hips with a more favorable postoperative distribution of contact hip stress had better clinical outcome. For each hip the peak contact hip stress before/after the operation was determined with the modification of the HIPSTRESS

method that took into account the non-weight-bearing necrotic part of the femoral head. The hips were evaluated clinically 9-26 years after the operation and divided into a successful and an unsuccessful group. In the successful group the operation caused an average decrease of the peak hip stress of about 10%, while in the unsuccessful group the operation caused an average increase of the peak hip stress of about 4%, the difference between the respective changes of the peak stress due to the operation being statistically significant [50].

CONCLUSION

The development of noninvasive models to estimate contact hip stress and clinical application of these models have been running hand in hand. Biomechanical studies of human hips based on the analysis of the two-dimensional pelvic radiographs have turned out to be a reasonable compromise between the measurement accuracy and the feasibility in clinical setting. Through wider availability of the three-dimensional imaging techniques, visualization of the femoral head coverage and pelvic muscle attachment points has improved considerably. However, the need to supplement the morphological hip status with biomechanical analysis remains. One of the limitations of the present biomechanical contact hip stress computations is the inability to estimate the patients' physical activity levels throughout their lifetime as such data cannot be reliably acquired retrospectively. The potential confounding effect of differences in physical activity is difficult to estimate because there is no clear evidence lifelong standing, walking, or lifting in the population with normal hips is a risk factor for osteoarthritis. The current trend in biomechanics is to combine the kinetic gait measurements of the resultant hip force with three-dimensional imaging of the hip weight-bearing surface in order to better estimate the contact hip stress for a given activity/body position. Nevertheless, the added value of such measurements over the two-dimensional pelvic radiograph analysis still needs to be established by clinical trials with larger numbers of participating subjects. Eventually, the clinical use of the contact hip stress methods will depend both on their accuracy and feasibility.

REFERENCES

[1]   Brand RA, Iglič A, Kralj-Iglič V. Contact stress in the human hip, implications for disease and treatment. *Hip. Int.* 2001;11:117-26.

[2]   Brinckmann P, Frobin W, Hierholzer E. Stress on the articular surface of the hip joint in healthy adults and person with idiopathic osteoarthrosis of the hip joint. *J. Biomech.* 1981;14:149-56.

[3]   Legal H: Introduction to the Biomechanics of the Hip. In: *Congenital Dysplasia and Dislocation of the Hip*. Ed by Tönnis D. Berlin, Springer-Verlag, 1987: pp 26-57.

[4]   Iglič A, Srakar F, Antolič V. Influence of the pelvic shape on the biomechanical status of the hip. *Clin. Biomech.* 1993;8:223-4.

[5]   Iglič A, Kralj-Iglič V, Antolič V, Srakar F, Stanič U. Effect of the periacetabular osteotomy on the stress on the human hip joint articular surgace. *IEEE Trans. Rehab. Engr.* 1993;1:207-12.

[6]   Ipavec M, Brand R A, Pedersen D R, Mavčič B, Kralj-Iglič V, Iglič A. Mathematical modelling of stress in the hip during gait. *J. Biomech.* 1999;32:1229-35.

[7]   Mavčič B, Pompe B, Antolič V, Daniel M, Iglič A, Kralj-Iglič V. Mathematical estimation of stress distribution in normal and dysplastic human hips. *J. Orthop. Res.* 2002;20:1025-30.

[8]   McLeish RD, Charnley J. Abduction forces in the one-legged stance. *J. Biomech.* 1970;3:191-209.

[9]   Hashimoto N, Ando M, Yayama T, Uchida K, Kobayashi S, Negoro K, Baba H. Dynamic analysis of the resultant force acting on the hip joint during level walking. *Artif. Organs.* 2005;29:387-92.

[10]   Heller MO, Bergmann G, Kassi JP, Claes L, Haas NP, Duda GN. Determination of muscle loading at the hip joint for use in pre-clinical testing. *J. Biomech.* 2005;38:1155-63.

[11]   Pfeifer R, Hurschler C, Ostermeier S, Windhagen H, Pressel T. In vitro investigation of biomechanical changes of the hip after Salter pelvic osteotomy. *Clin. Biomech.* 2008;23:299-304.

[12]   Wiberg G. Studies on dysplastic acetabula and congenital subluxation of the hip joint-with special reference to the complication of osteoarthritis. *Acta. Orthop. Scand.* 1939;83(Suppl 58):1-135.

[13]   Lequesne M, de Sèze S. Le faux profil du bassin: Nouvelle incidence radiographique pour l'étude de la hanche: Son utilité dans les dysplasies et les différentes coxopathies. *Rev. Rhum. Mal. Osteoartic.* 1961;28:643-52.

[14]   Delaunay S, Dussault RG, Kaplan PA, Alford BA. Radiographic measurements of dysplastic adult hips. *Skeletal. Radiol.* 1997;26:75-81.

[15]   Daniel M, Iglič A, Kralj-Iglič V. The shape of acetabular cartilage optimizes hip contact stress distribution. *J. Anat.* 2005;207:85-91.

[16]   Mechlenburg I, Nyengaard J R, Romer L, Søballe K. Changes in load-bearing area after Ganz periacetabular osteotomy evaluated by multislice CT scanning and stereology. *Acta. Orthop. Scand.* 2004;75:147-53.

[17]   Hipp JA, Sugano N, Millis MB, Murphy SB. Planning acetabular redirection osteotomies based on joint contact pressures. *Clin. Orthop. Relat. Res.* 1999;364:134-43.

[18]   Ganz R, Leunig M, Leunig-Ganz K, Harris WH. The etiology of osteoarthritis of the hip: an integrated mechanical concept. *Clin. Orthop. Relat. Res.* 2008;466:264-72.

[19]   Hadley NA, Brown TD, Weinstein SL. The effects of contact pressure elevations and aseptic necrosis on the long-term clinical outcome of congenital hip dislocation. *J. Orthop. Res.* 1990;8:504-13.

[20]   Vengust R, Drobnič M, Daniel M, Antolič V, Pernus F, Iglič A, Kralj-Iglič V. Role of magnification of standard anterior-posterior radiographs in determination of contact hip joint stress. *Biomed. Eng. Appl. Basis. Comm.* 2000;12:242-4.

[21]   Daniel M, Antolič V, Iglič A, Kralj-Iglič V. Determination of contact hip stress from nomograms based on mathematical model. *Med. Eng. Phys.* 2001;23:347-57.

[22]   Russell ME, Shivanna KH, Grosland NM, Pedersen DR. Cartilage contact pressure elevations in dysplastic hips: a chronic overload model. *J. Orthop. Surg. Res.* 2006;3:1-6.

[23] Mavčič B, Iglič A, Kralj-Iglič V, Brand RA, Vengust R. Cumulative hip contact stress predicts osteoarthritis in DDH. *Clin. Orthop. Relat. Res.* 2008;466:884-91.

[24] Brand RA, Pedersen DR, Davy DT, Kotzar GM, Heiple KG, Goldberg VM. Comparison of hip force calculation and measurements in the same patient. *J. Arthroplasty* 1994;9:45-51.

[25] Jacobsen S, Sonne-Holm S. Hip dysplasia: a significant risk factor for the development of hip OA. A cross-sectional survey. *Rheumatology* 2005;44:211-8.

[26] Lievense AM, Bierma-Zeinstra SMA, Verhagen AP, Verhaar JAN, Koes BW. Influence of hip dysplasia on the development of OA of the hip. *Ann. Rheum. Dis.* 2004;63:621-6.

[27] Jacobsen S, Sonne-Holm S. Increased body mass index is a predisposition for treatment by total hip replacement. *Int. Orthop.* 2005;29:229-34.

[28] Maxian TA, Brown TD, Weinstein SL. Chronic stress tolerance levels for human articular cartilage: two non-uniform contact models applied to long term follow up of CDH. *J. Biomech.* 1995;28:159-66.

[29] Jacobsen S, Sonne-Holm S, Søballe K, Gebuhr P, Lund B. Joint space width in dysplasia of the hip. *J. Bone Joint Surg. Br.* 2005;87:471-7.

[30] Murphy SB, Kijewski PK, Millis MB, Harless A. Acetabular dysplasia in the adolescent and young adult. *Clin. Orthop. Relat. Res.* 1990;261:214-23.

[31] Bellamy N, Buchanan WW, Goldsmith CH, Campbell J, Stitt LW. Validation study of WOMAC: a health status instrument for measuring clinically important patient relevant outcomes to antirheumatic drug therapy in patients with osteoarthritis of the hip or knee. *J. Rheumatol.* 1988;15:1833-40.

[32] Iglič A, Daniel M, Kralj-Iglič V, Antolič V, Jaklič A. Peak hip joint contact stress in male and female populations. *J. Musculoskeletal. Res.* 2001;5:17-21.

[33] Heliovara M, Makela M, Impivaara O, Knekt P, Aromaa A, Sievers K. Association of everweight, trauma and workload with coxaarthrosis. A health survey of 7217 persons. *Acta. Orthop. Scand.* 1993;64:513-8.

[34] Rečnik G, Kralj-Iglič V, Iglič A, Antolič V, Kramberger S, Vengust R. Higher peak contact hip stress predetermines the side of hip involved in idiopathic osteoarthritis. *Clin. Biomech.* 2007;22:1119-24.

[35] Rečnik G, Kralj-Iglič V, Iglič A, Antolič V, Kramberger S, Rigler I, Pompe B, Vengust R The role of obesity, biomechanical constitution of the pelvis and contact joint stress in progression of hip osteoarthritis. *Osteoarthritis Cartilage* 2009 (in print).

[36] Michaeli DA, Murphy SB, Hipp JA. Comparison of predicted and measured contact pressures in normal and dysplastic hips. *Med. Eng. Phys.* 1997;19:180-6.

[37] Salter RB. Innominate osteotomy in the treatment of congenital dislocation and subluxation of the hip. *J. Bone Joint Surg. Br.* 1961;43-B:518-39.

[38] LeCoeur P. Corrections des défauts d'orientation de l'articulation coxo-fémorale par ostéotomie de l'isthme iliaque. *Rev. Chir. Orthop.* 1965;51:211-2.

[39] Chiari K. Medial displacement osteotomy of the pelvis. *Clin. Orthop. Relat. Res.* 1974;98:55-71.

[40] Tönnis D. *Congenital dysplasia and dislocation of the hip in children and adults*. New York, Springer 1987.

[41] Ganz R, Klaue K, Vinh TS, Mast JW. A new periacetabular osteotomy for the treatment of hip dysplasia. Technique and preliminary results. *Clin. Orthop. Relat. Res.* 1988;232:26-36.

[42] Søballe K. Pelvic osteotomy for acetabular dysplasia. *Acta. Orthop. Scand.* 2003;74:117-8.

[43] Murphy S, Deshmukh R. Periacetabular osteotomy: preoperative radiographic predictors of outcome. *Clin. Orthop. Relat. Res.* 2002;405:168-74.

[44] Trousdale RT, Cabanela ME. Lessons learned after more than 250 periacetabular osteotomies. *Acta. Orthop. Scand.* 2003;74:119-26.

[45] Kralj M, Mavčič B, Antolič V, Iglič A, Kralj-Iglič V. The Bernese periacetabular osteotomy: clinical, radiographic and mechanical 7-15-year follow-up of 26 hips. *Acta. Orthop.* 2005;76:833-40.

[46] Teratani T, Naito M, Shiramizu K, Nakamura Y, Moriyama S. Modified pubic osteotomy for medialization of the femoral head in periacetabular osteotomy: a retrospective study of 144 hips. *Acta. Orthop.* 2008;79:474-82.

[47] Mavčič B, Slivnik T, Antolič V, Iglič A, Kralj-Iglič V. Hip contact stress is related to the development of hip pathology with increasing age. *Clin. Biomech.* 2004;19:939-43.

[48] Crockarell J Jr, Trousdale RT, Cabanela ME, Berry DJ. Early experience and results with the periacetabular osteotomy. The Mayo Clinic experience. *Clin. Orthop. Relat. Res.* 1999;363:45-53.

[49] Zupanc O, Krizančič M, Daniel M, Mavčič B, Antolič V, Iglič A, Kralj-Iglič V. Shear stress in epiphyseal growth plate is a risk factor for slipped capital femoral epiphysis. *J. Pediatr. Orthop.* 2008;28:444-51.

[50] Dolinar D, Antolič V, Herman S, Iglič A, Kralj-Iglič V, Pavlovčič V. Influence of contact hip stress on the outcome of surgical treatment of hips affected by avascular necrosis. *Arch. Orthop. Trauma. Surg.* 2003;123:509-13.