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One-Legged Stance as a Representative Static Body Position for Calculation of Hip Contact Stress Distribution in Clinical Studies

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It was shown in several clinical studies that static one-legged stance may be a relevant body position to describe the loads acting at the hip. However, the stress distribution averaged during movement may better describe hip load than hip contact stress distribution in the static body position. Using data on the resultant hip force during walking taken from the measurements of Bergmann (2001), spatial distribution of contact stress over the articular surface was calculated by the HIPSTRESS method and compared with the stress distribution in one-legged stance. It is shown, that the shape of the contact stress distribution during one-legged stance closely resembled the averaged contact stress distribution during the walking cycle (Pearson's correlation coefficient $R^2 = .986$; $p < .001$). This finding presents a link between the hypothesis that the averaged contact stress distribution during a walking cycle is crucial for cartilage development and the results of clinical studies in which the calculated distribution of contact stress in one-legged stance was successfully used to predict the clinical status of the hip.

***Keywords*:** contact pressure, hip biomechanics, mathematical model

Experimental measurements (Krebs et al., 1991; Morrell et al., 2005) and mathematical modeling have been widely used to determine the contact stress distribution in the human hip (Brinckmann et al., 1981; Dalstra & Huiskes, 1995; Hadley et al., 1990; Legal, 1987; Mavčič et al., 2002). Although hip contact stress can also be computed during walking and other daily activities using dynamic loading forces in the human hip as input data (Ipavec et al., 1999; Yoshida et al., 2006), for the sake of simplicity in clinical (AP radiograph) studies only the contact stress in static one-legged stance is usually considered (Genda et al., 2001; Iglič et al., 2002; Legal, 1987; Rečnik et al., 2007; The et al., 2008). Using one-legged stance as a representative body position for determination of contact stress it was shown that the peak contact stress (Mavčič et al., 2002, 2004), as well as the parameters determining hip contact stress distribution (such as hip joint loading) (Hadley et al., 1990; Pompe et al., 2003), can be used to evaluate the biomechanical loading of the hip joint.

It has been suggested recently that the temporal and spatial aspects of contact stress distribution may be more important for hip development than the value of the peak stress itself (Brand, 2005). Based on this assumption, the stress distribution, averaged during hip movement, may better describe biomechanical loading of the hip than the stress distribution in static one-legged stance. On the other hand, it was shown in different clinical studies that the static one-legged stance may be relevant for adequate description of the biomechanical loading of the hip (Genda et al., 2001; Mavčič et al., 2002, 2004; Rečnik et al., 2007). It is the aim of this study to evaluate averaged stress distribution during a walking cycle and compare it with the stress distribution in one-legged stance using the same approach as used previously in the above-mentioned clinical studies.

Methods

The hip joint contact stress over the articular surface was determined using the method HIPSTRESS (Iglič et al., 2002; Mavčič et al., 2002, 2004; Pompe et al., 2003) that was developed previously (Ipavec et al., 1999). The input parameters of the mathematical model HIPSTRESS in both the one-leg stance and walking are: the radius of articular surface r, position of the acetabulum and the hip joint reaction force $\boldsymbol{R}$. The radius of the articular surface

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was taken to be 25 mm. It was assumed that an angle formed between the center of the hip, outer edge of the acetabulum in the frontal plane and vertical, the center-edge angle (ϑ_{CE}) after Wiberg (1939), is 30°, while no acetabular anteversion was taken into account (Figure 1). The contact stresses were normalized relative to body weight (BW).

The values of the hip joint reaction forces measured by using an implanted instrumented endoprosthesis during one step of normal walking (Bergmann, 2001) were used to compute the average stress distribution during the walking cycle (p_{walk}). Hip joint force is defined in 201 equidistant time steps during walking (Bergmann, 2001). The stress over the contact area was computed for each time step, and the average stress (p_{walk}) for a given point at the articular surface was computed by dividing the sum of stresses at that point across all time steps by the number of steps. The stress was normalized relative to body weight (BW).

The force estimated for an average patient in the clinical studies (Mavčič et al., 2002, 2004; Pompe et al., 2003) was used to compute the contact stress in one-legged stance (p_{oneleg}). Using the equilibrium of forces and torques acting on the pelvis in one-legged stance, it was shown that the hip joint loading force lies (nearly) in the frontal plane of the body during one-legged stance (Figure 1) (Iglič et al., 2002). The values of medial inclination of the force $\boldsymbol{R}$ from the sagittal plane of the body ϑ_R and the magnitude of the hip joint reaction force R were taken from the clinical study of healthy patients, ϑ_R equals 8° and R is 2.7 of body weight (BW) (Table 1 in Mavčič et al., 2002).

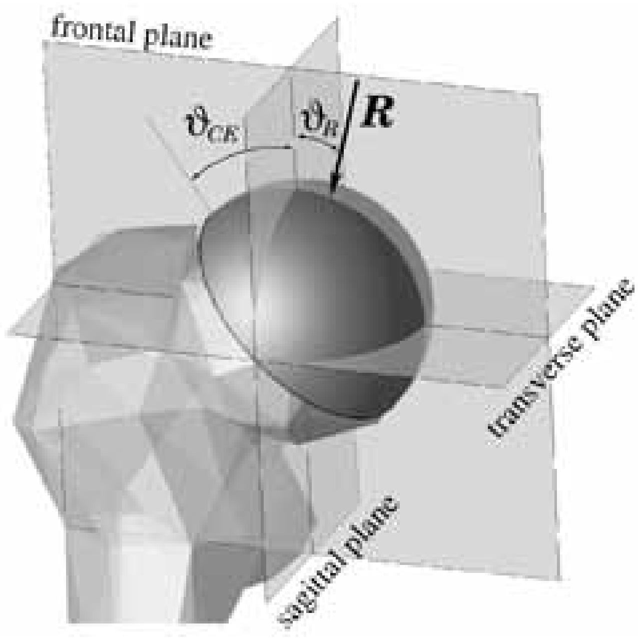

Figure 1 — Geometrical model of the articular cartilage and its position in the body coordinate frame defined by frontal plane, sagittal plane, and transverse plane of the patient. The hip joint reaction force in one-legged stance ($\boldsymbol{R}$) is shown, ϑ_R denotes inclination of the hip joint reaction force, and ϑ_{CE} denotes center-edge angle of Wiberg.

Results and Discussion

Projections of contact stress averaged over the walking cycle and the contact stress distribution in one-legged stance are shown in Figure 2. The peak contact stress in one-legged stance normalized to BW (max p_{oneleg} = 2908 Pa/N) is much higher than the peak value of contact stress averaged over the walking cycle (max p_{walk} = 1406 Pa/N). This can be expected since the walking cycle also includes the swing phase (Legal, 1987) in which hip joint loading is substantially lower than in the stance phase (Brand et al., 1994; Hodge et al., 1986; Krebs et al., 1991).

To compare the spatial distribution of contact stress averaged over walking cycle and the stress in one-legged stance, the stress is computed relative to the correspondent maximal value, $p^{norm} = p / max(p)$. The average difference of relative contact stresses $\Delta p^{norm} = (p^{norm}_{walk} - p^{norm}_{oneleg})$ over the articular surface is +2.5% with standard deviation 3.1% (Figure 2C). One-leg stance underestimates loading of the antero-inferior region of the acetabulum where stress averaged over the walking cycle is relatively higher (Δp^{norm} =13.3%), while it predicts higher contact stresses in the postero-superior acetabulum than stresses averaged over the walking cycle (Δp^{norm}=-5.7%). Results show a strong correlation between the spatial distribution of the p_{walk} and p_{oneleg} (Pearson correlation coefficient R^2 = .986; p < .001) and a small difference in relative contact stresses in the superior acetabulum ($|\Delta p^{norm}|$<3%, Figure 2C), that is known as the principle load-bearing area (Legal, 1987). This finding may present a link between the hypothesis that the averaged contact stress distribution during a walking cycle is relevant for cartilage development and the results of the clinical studies in which the calculated values of contact stress distribution in one-legged stance were successfully used to describe the clinical status of the hip (Hadley et al., 1990; Mavčič et al., 2002; Pompe et al., 2003).

If the hip joint reaction force $\boldsymbol{R}$ during daily activities can be determined (e.g., using implanted endoprosthesis (Brand et al., 1994; Bergmann, 2001; Hodge et al., 1986, Krebs et al., 1991; Morrell et al., 2005) or inverse dynamical analysis (Crowninshield et al., 1978; Brand et al., 1994), it is most appropriate to use these values of $\boldsymbol{R}$ to assess the hip contact stress distribution. However, in many cases these measurements are not possible for a given patient, especially if the data are taken from archives or if the patient's physical condition doesn't allow gait analysis. According to our results, one-legged stance can be chosen not only as a representative body position frequently attained during daily activities (Iglič et al., 2002), but also as a position in which the contact stress distribution corresponds to the average contact stress during walking with a standard deviation of 3.1%. Since the computation of hip contact stress distribution

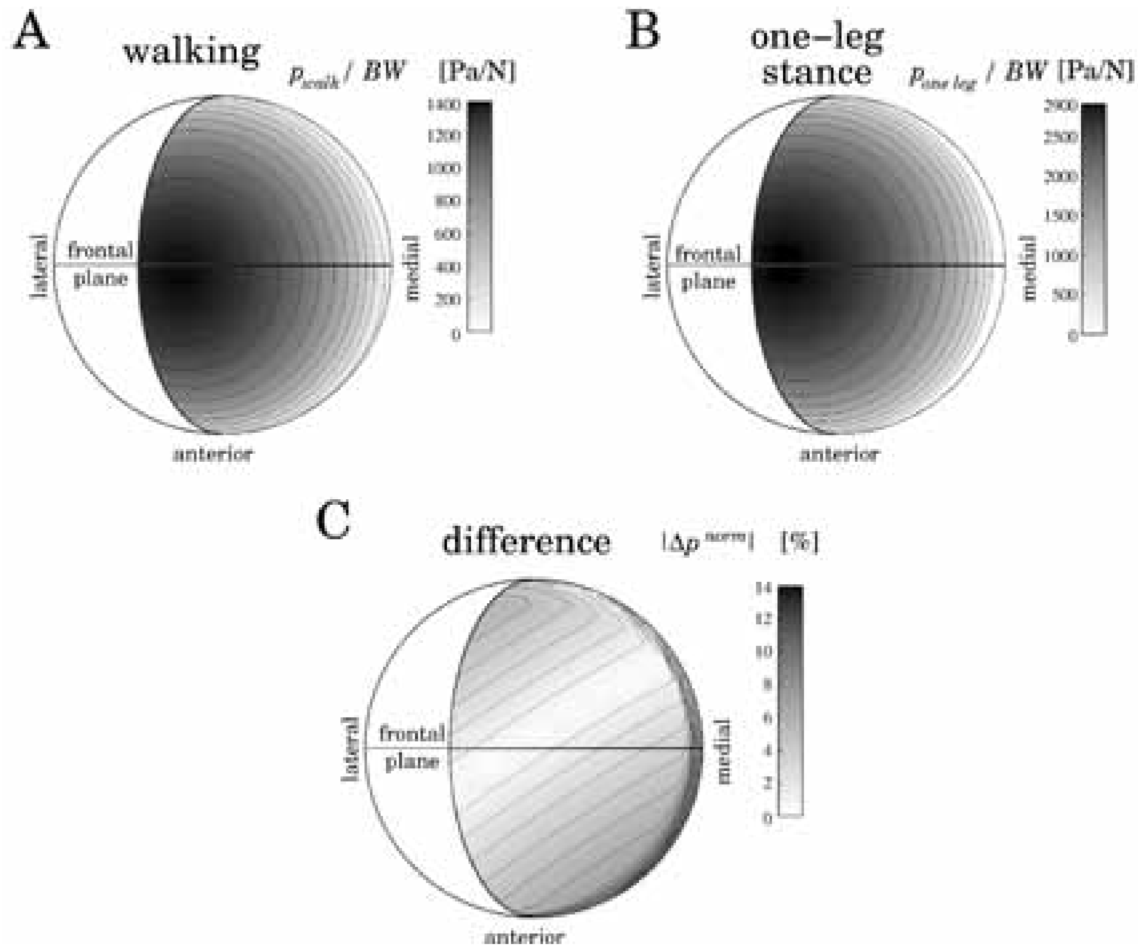

Figure 2 — Projections of the contact stress distribution (A) averaged during the walking cycle p_{walk} and (B) the contact stress distribution in one-legged stance p_{oneleg} in the transverse plane. The values of stress are normalized with respect to body weight (BW). The isolines correspond to tenths of the maximal values of stress. (C) Projection of absolute value of difference in contact stresses relative to their maximal values $\left|\Delta p^{norm}\right| = \left|p_{walk} / \max(p_{walk}) - p_{oneleg} / \max(p_{oneleg})\right|$. The isolines correspond to 1% of difference.

in one-legged posture (Iglič et al., 2002) is much simpler than the computation of the dynamic force during body motion (Crowninshield et al., 1978; Brand et al., 1994), it offers an advantage in everyday clinical practice with respect to inverse dynamic analysis.

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