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Determination of contact hip stress from nomograms based on mathematical model

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Abstract

Nomograms are presented that enable determination of maximal stress on the hip joint weight bearing area if certain geometrical parameters of the hip and pelvis and the body weight are known. The nomograms are calculated by using previously developed mathematical models. It is demonstrated how the maximal stress on the hip joint weight bearing area is determined from the presented nomograms for a hip for which the geometrical parameters were obtained from a standard anteroposterior rentgenograph. This simple and noninvasive method may give insight into the biomechanical status of the hip which should be considered in routine surgical planning and as a part of the routine examination of the patient without the use of any additional tools.

Keywords: Hip; Hip stress; Arthrosis; Weight bearing area; Nomograms

1. Introduction

Contact hip joint stress distribution is an important feature that affects cartilage and bone adaptation [33,28,4,26] and therefore influences the development of the hip joint. It was suggested that a long lasting high contact stress in the hip joint may accelerate the development of arthrosis [10,30,3]. As an unfavourable biomechanical state of the hip, the elevated hip joint contact stress was assumed to be a cause of arthrosis, and different osteotomies involving pelvis and proximal femur were introduced in order to reduce and redistribute contact stress on the weight bearing area in the hip joint [3,12,19].

Increased contact stress in the joint articular surface can result from too small a lateral coverage of the femoral head, from too high a resultant hip joint force or from too vertical resultant hip joint force [3,12,19,23,26]. The direction and magnitude of the resultant hip joint force **R** depends among others on the femoral and pelvic geometry [3,18,20,24,25,33] and body weight. Traditionally, the combination of small lateral coverage of acetabulum, i.e. small centre edge angle ϑ_{CE} (Fig. 1), and obesity is thought to be a risk factor for arthrosis development. However, small angle ϑ_{CE} (i.e. small lateral coverage of femoral head) does not necessarily also implicate high contact stress in the hip joint. For example in the case of large medial inclination of the hip joint resultant force (ϑ_R), the hip joint contact stress may be relatively low even at very small ϑ_{CE} angle [23]. In different acetabular osteotomies the hip joint contact stress may be additionally reduced by medialisation [19] of the acetabulum causing a decrease of the magnitude of the hip joint resultant force [24,17]. Hence it can be suggested that the centre edge angle of Wiberg ϑ_{CE} alone cannot be an adequate criterion for consideration in routine surgical planning and in estimation of the long term success of different osteotomies in dysplastic hips. Consequently, more demanding quantitative methods are needed to help surgeons in preoperative

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planning of different acetabular osteotomies in order to reduce hip joint contact stress. A relevant description of the biomechanical state of the hip could be a valuable tool in deciding the optimal treatment and also in predicting the patterns to avoid arthrosis. The hip joint contact stress and hip joint contact stress gradient were suggested as the relevant parameters to help surgeons in planning femoral and pelvic osteotomies [3,12,19]. For simplicity the maximal value of stress attained on the weight bearing area $p_{\max}$ was used as the measure of the hip joint contact stress [12,19,23].

A direct measurement of the stress distribution was performed in a patient who was implanted with a partial hip endoprothesis with a stress measuring device [13–15]. Valuable information was obtained in the course of rehabilitation and further in different activities of the patient. However, for routine clinical application this method cannot be applied. A combination of external measurements [2] with mathematical models can therefore be used in nonoperated persons [23]. In this case, it is important to have a mathematical model that accounts for the relevant features that determine hip stress. The method should not be time consuming and demanding with respect to the necessary equipment and skills, so that it could be applied as part of the routine examination of the patient and surgical planning. Therefore, simple mathematical models have been developed [27,28,4,19,9] to complement sophisticated models based on finite element methods [36,5,7].

Over the last decade our group developed a three dimensional mathematical model for calculation of the resultant hip joint force **R** in a one legged stance [16,18] and the corresponding model for determination of contact stress distribution over the hip joint articular surface [19,23]. These models were used for calculation of the resultant hip joint force and/or contact stress distribution in different pelvis shapes [18], and for simulation of different postoperative configurations of the pelvis and hip in a one legged stance [17,19,20]. The calculated hip joint resultant force in the one legged stance can be scaled by the individual variations in muscle attachment points [17,18,21] where the femoral and pelvic geometrical parameters (Fig. 1) should be determined from the anteroposterior radiographs.

Recently, the model for stress distribution was generalised [23] and used in combination with a sophisticated opto-electronic system, piezoelectric force plate and mathematical model for determination of the hip joint resultant force during gait [2], in order to monitor the contact stress distribution in the hip joint during gait [23,29].

Several assumptions and simplifications were introduced in the above model for stress distribution [23]. First, the cosine radial stress distribution function [4,19] adopted in the model is derived from the assumption that the radial stress in the hip articular surface is proportional to the radial strain of the cartilage layer [28]. The cartilage layer is assumed to be a homogenous continuum and a linear elastic solid of a uniform thickness [23]. Second, the femoral head and the acetabular shell are taken to be spherical, while the thickness of the cartilage layer before the deformation is assumed to be constant. Since the validity of the cosine stress distribution function used in this work is based on the assumption that the femoral head and the acetabulum have spherical geometry, the deviation from this situation would in general lead to a stress distribution function different from a cosine function. When the deviation from sphericity is small the contact stress distribution could be described as a perturbation to the cosine function [23]. Third, the weight bearing area is overestimated as a result of not accounting for the cotyloid notch. However, this region would not be expected to actually distribute much load. Therefore, the actual underestimation of stress distribution as a result of overestimating contact area is negligible [23].

Due to the described limitations of the model we cannot predict local contact stress in the hip joint in detail. However, the model and the corresponding computer program [23] may predict the global stress distribution in the hip joint in accordance with the relevant in vivo data [13–15] which demonstrate that the contact stress in the hip joint is nonuniform in a way as predicted by our model [23]. Moreover, the measured shape of the weight bearing area [13–15] in the stance period of gait may be considered, in a rough fashion, to correspond to the shape of the weight bearing area obtained by our model [23]. We also found a value of peak stress $p_{\max}$ in the stance period of gait around 3 MPa for the normal values of ϑ_{CE}, femoral head radius and body weight. This correponds well to the in vivo data [14,15].

In this work we present a method for routine estimation of maximal hip stress $p_{\max}$ in one legged stance by using a standard anteroposterior rentgenograph and the presented nomograms. Direct use of the computer is not required. This may be of advantage in routine surgical planning and as a part of the routine examination of the patient since no additional tools and skills are required. If the problem under consideration involves analysis of an individual case or a small number of cases, this method may also be faster. In nomograms presented in this work, the resultant hip force **R** in one-legged stance and the peak stress in the hip joint $p_{\max}$ are calculated by using the previously developed mathematical models and computer programs described below [18,23].

2. Material and methods

The nomograms were calculated by using the program HIPSTRESS [21] that is based on two programs. The first program for determination of the hip joint resultant force [16–18] provides the solution of the equilibrium equations for forces and torques in the one-legged stance. The calculated hip joint resultant force in the one legged stance is scaled by the individual variations in muscle attachmnet points [1,17,18,21]. The second program [23] includes consistently related equations for determination of the stress pole and of the medial border of the weight bearing area. The program for the determination of the hip joint contact stress distribution [23] requires as input data the magnitude and direction of the resultant hip joint force **R**, the centre edge angle ϑ_{CE} and the radius of the hip joint articular surface r (Fig. 1).

As already mentioned above, the reference values of the model muscle insertion points [8,16–18] are rescaled in order to adjust the configuration of the hip and pelvis of the individual person [1,17–21,39]. Thereby the measured values of the hip and pelvic geometrical parameters from the standard anteroposterior rentgenographs for the given patient are taken into account [1,16–19,25,35]. The input hip and pelvic geometrical parameters of the model for determination of the resultant hip force [16–19] are (Fig. 1): half of the distance between the centres of the femoral heads i.e. half of the interhip distance ($l/2$), the vertical distance between the centre of the femoral head and the highest point on the crista iliaca (P_H), i.e. the pelvic height (H), the horizontal distance between the centre of the femoral head and the most lateral point on the crista iliaca (P_C), i.e. the pelvic width (C), the vertical and the horizontal distance from the centre of the femoral head to the effective muscle attachment point (T) on the greater trochanter (z and x, respectively), and the body weight W_b. The point T is determined by the intersection of the contour of the greater trochanter and the normal through the midpoint of the straight line connecting the most lateral point (point 1) and the highest point (point 2) on the greater trochanter [1,20,39] (Fig. 1). The femoral head centre can be determined by template with Mose circles [32]. The geometrical parameters described above are used to scale the respective reference values of the attachment points of the muscles [8] included in the model for individual persons [16–22]. The values of the pelvic and femoral geometrical parameters H, C, l, z and x are determined in the reference frame where the straight line connecting the centres of both femoral heads defines the horizontal line and also the frontal plane of the body. The line determining H is perpendicular to horizontal, while the lines determining the values of C and z are parallel to the horizontal. While taking the rentgenograph both femurs should be in the zero joint configuration where the straight line connecting the femoral head centre and the midpoint between the lateral and medial epicondyle is perpendicular to the straight line through both femoral head centres [8,16]. The abduction or adduction of the legs from this reference configuration would affect the accuracy of the correct values of the coordinates of the point T, i.e. z and x.

As mentioned above, the input parameters of the com-

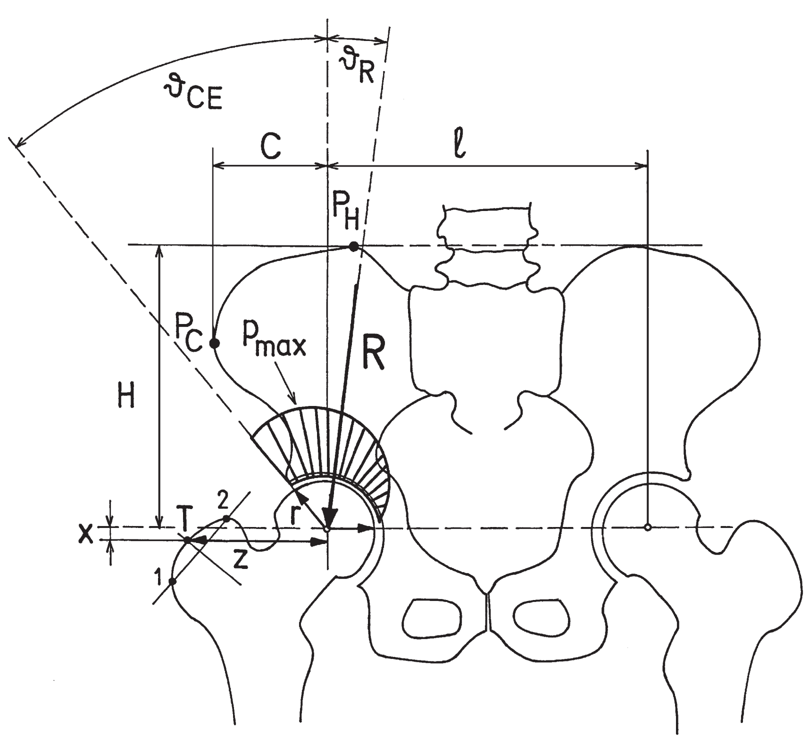

Fig. 1. The geometrical parameters of the hip and pelvis needed for determination of the maximal stress on the weight bearing area. The stress distribution and the resultant hip joint force **R** are also shown schematically. Symbol ϑ_R denotes the inclination of **R** with respect to the vertical, the point T denotes the effective muscle attachment point on the greater trochanter.

puter program for calculation of the hip stress distribution [23] are the magnitude of the resultant hip force **R**, the direction of **R** represented by its inclination with respect to the vertical ϑ_R, the centre edge angle ϑ_{CE} and the radius of the femoral head r.

Since the contact stress distribution in the hip joint depends on the femoral and pelvic geometrical parameters [12,17–25] the accuracy of the measured values of these geometrical parameters (Fig. 1) also influences the accuracy of the contact stress distribution determined from the antero-posterior radiographs [38]. Hence, the correct magnification factor should be determined individually for each patient. The magnification factor can be determined by using the metal lamella of known length L_o which should be placed by a special screw system, to the level of the femoral head centres before making the anteroposterior radiograph. From the length of the lamella on the antero-posterior radiograph image (L) the magnification factor L/L_o can be calculated [38].

Below we present the nomograms and describe how the peak stress on the weight bearing area can be determined by using the standard anteroposterior rentgenographs and the presented nomograms.

Figs. 2–6 present the dependencies of the inclination of the resultant hip force ϑ_R on the half of the interhip distance $l/2$. The figures are characterised by the values of the pelvic width C. Different curves on each graph represent different values of the horizontal distance between the centre of the femoral head and the effective muscle attachment point on the greater trochanter (z) while different values of the pelvis height H are taken into account by presenting three different graphs in each figure.

Fig. 7 presents the dependence of the magnitude of

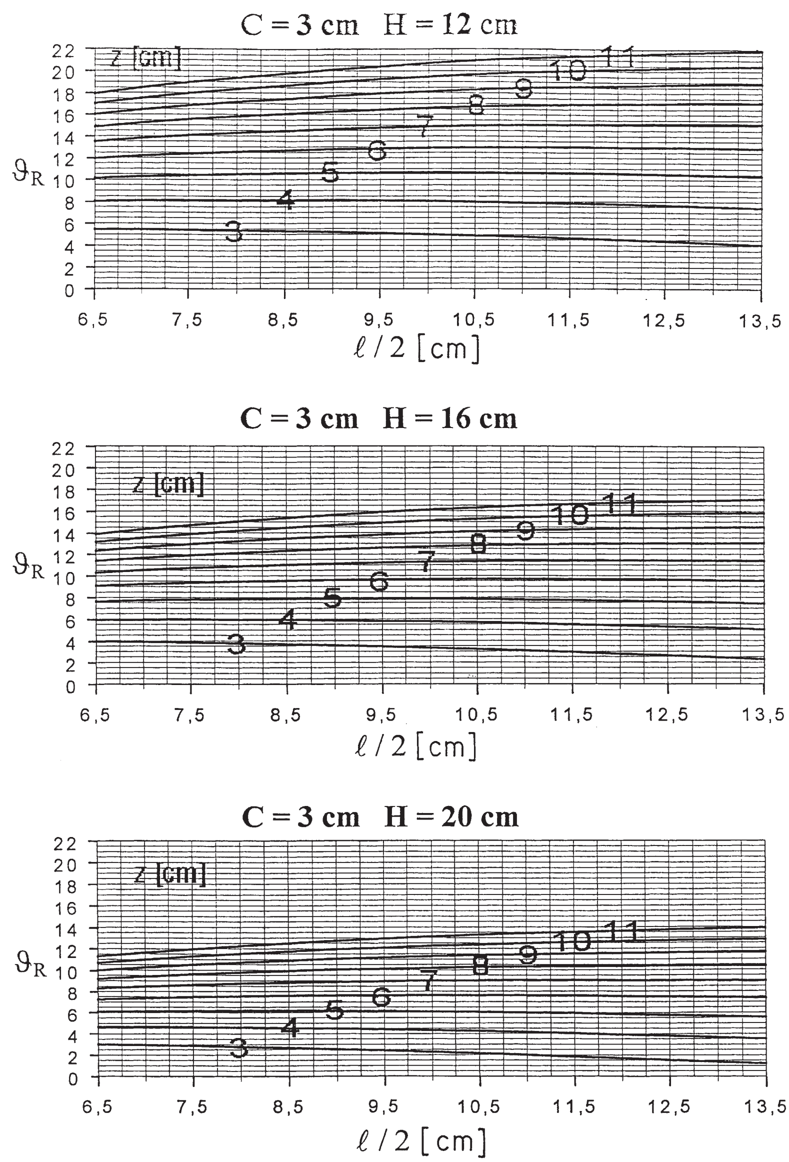

Fig. 2. Nomograms for determination of the inclination of the resultant hip force with respect to the vertical ϑ_R (C=3 cm).

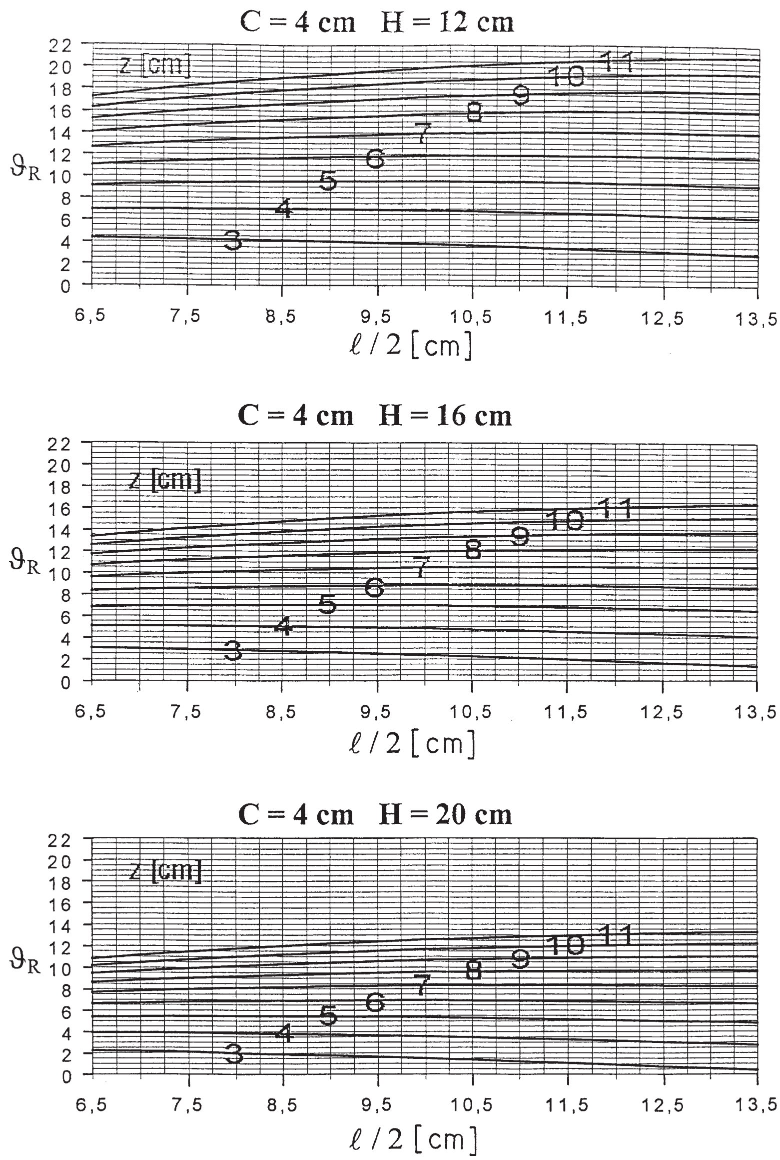

Fig. 3. Nomograms for determination of the inclination of the resultant hip force with respect to the vertical ϑ_R (C=4 cm).

the resultant hip force normalised by the body weight R/W_b on the half of the interhip distance $l/2$. Different curves on the graph represent different values of the horizontal distance between the centre of the femoral head and the effective muscle attachment point on the greater trochanter (z).

Figs. 8–10 present the dependence of the maximal stress on the weight bearing area divided by the body weight and multiplied by the square of the femoral head radius $p_{max}r^2/W_b$ on the sum of the angles ϑ_R and ϑ_{CE}. The figures are characterised by the range of (ϑ_R+ϑ_{CE}): Fig. 8 accounts for the interval of (ϑ_R+ϑ_{CE}) between 10 and 20°, Fig. 9 accounts for the interval of (ϑ_R+ϑ_{CE}) between 20 and 30°, while Fig. 10 accounts for the interval of (ϑ_R+ϑ_{CE}) between 30 and 60°. Different curves on the graph represent different values of R/W_b.

As there are many parameters that influence the maximal value of stress in the hip joint articular surface p_{max}, it is necessary to perform the determination of maximal stress stepwise, as described in Appendix A.

3. Results and discussion

As an example, we would like to determine the maximal stress for a hip with l=17 cm, C=4.5 cm, H=13 cm, z=4 cm, ϑ_{CE}=25°, r=2.5 cm and W_b=750 N. The geometrical parameters l, C, H, z, ϑ_{CE} and r were determined from the antero-posterior radiograph. We chose Fig. 3, the first graph at the top of the figure. It can be seen from the curve labelled by the number 4 (for z=4 cm) that the inclination ϑ_R corresponding to a value of $l/2$=8.5 cm is 7°. Then we considered Fig. 7, chose the curve labelled by the number 4 (for z=4 cm) and

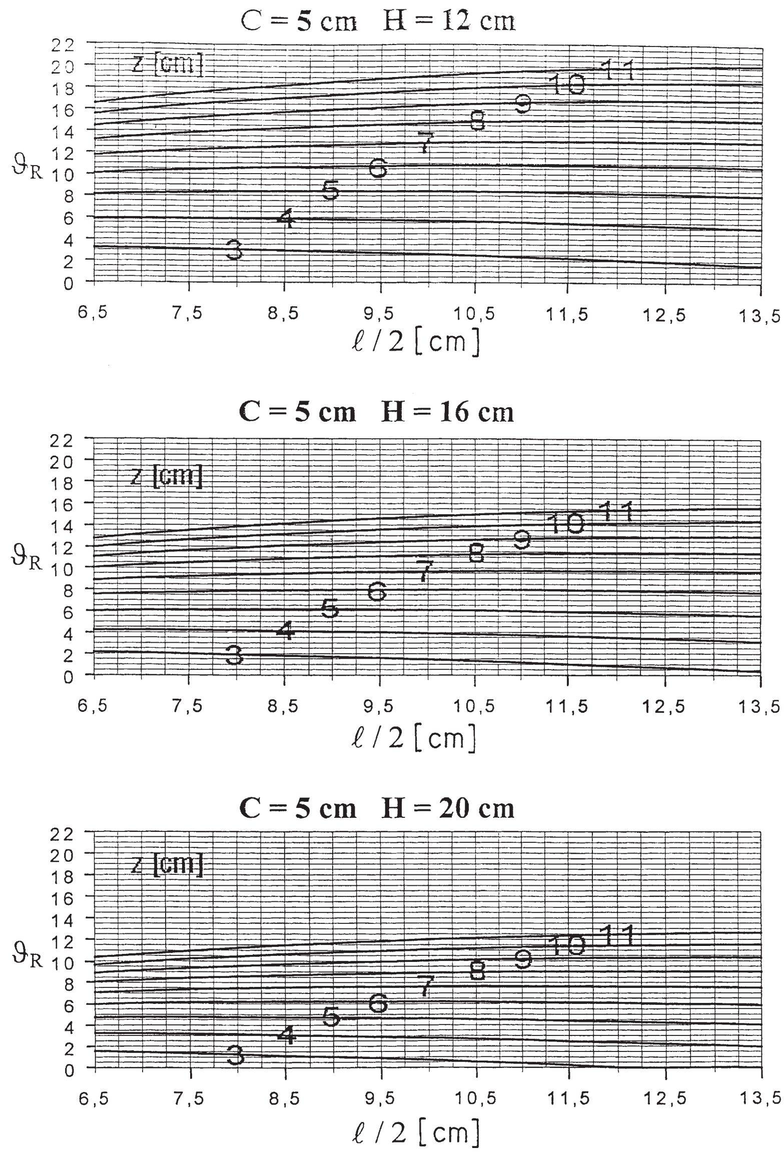

Fig. 4. Nomograms for determination of the inclination of the resultant hip force with respect to the vertical ϑ_R (C=5 cm).

obtained the value of the normalised force R/W_b to be 3.1. The sum $\vartheta_R+\vartheta_{CE}$ is apparently 7°+25°=32°, therefore we chose Fig. 10 to determine the normalised maximal stress. In Fig. 10 we chose the curve labelled by the number 3. The value of $p_{max}r^2/W_b$ that corresponds to the sum $\vartheta_R+\vartheta_{CE}$=32 is 2.5. To obtain the maximal stress p_{max} we multiply the result by the body weight W_b and divide it by the square of the femoral head radius r^2: (2.5·750 N)/(6.25 cm^2) = 3.0 MPa.

The value of p_{max} obtained directly by the program HIPSTRESS [21] for the above data is 3.2 MPa, therefore the error made by using the nomograms instead of the computer program is in this particular case about 7%. For illustration Fig. 11 shows the distribution of the difference between the values of p_{max} determined by the computer using the program HIPSTRESS and p_{max} determined manually by using the nomograms (N=38). The average difference between the values of p_{max} determined by the computer program HIPSTRESS and manually using the nomograms was 4.6%. It can therefore be concluded that within this precision there was no need for presenting additional nomograms to account for different values of the vertical coordinate of the effective muscle attachment point on the greater trochanter $x\in(-1$ cm, 1 cm).

It has been shown that vertical floor reaction force is nearly constant for a period during the stance phase of gait [6]. Therefore the static one legged stance body position is important not only due to itself but also due to its resemblance to the midstance phase of slow walking gait [31]. Consequently, the static one legged stance can be used as a representative body position [6,12] for patients who usually walk slowly [31]. In addition, it was also indicated that the peak hip stress during the

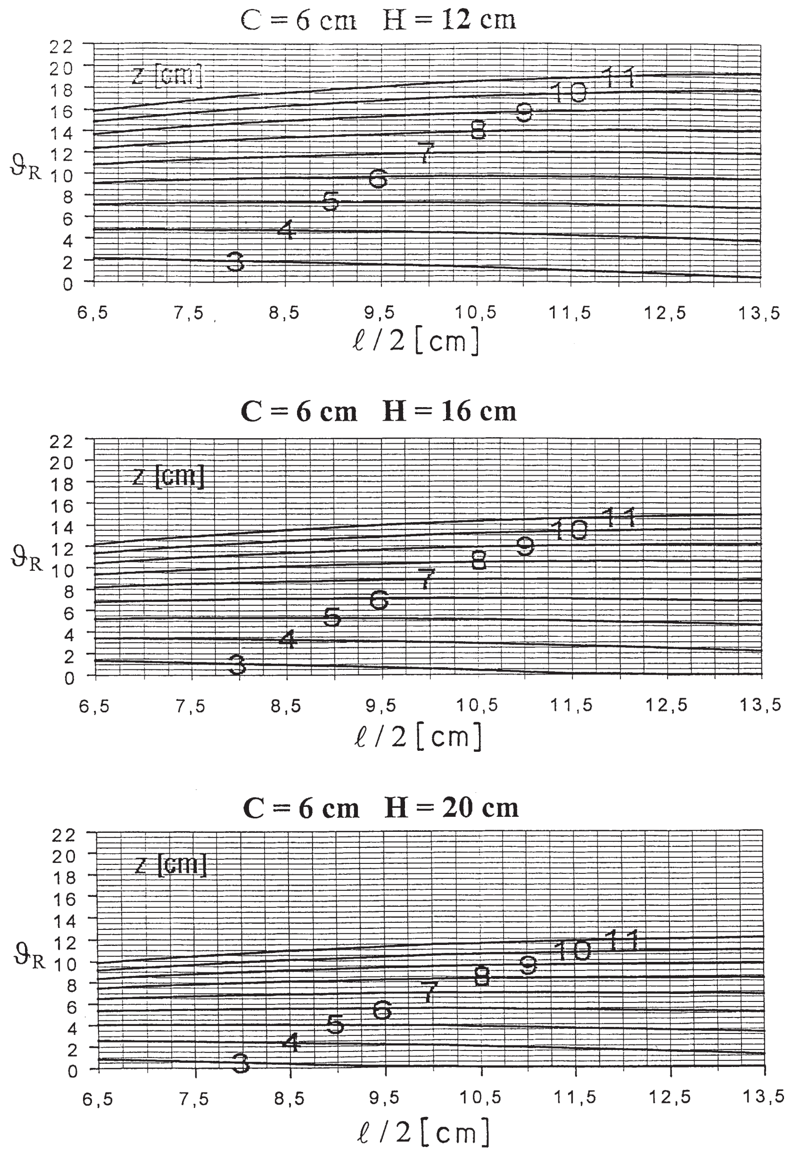

Fig. 5. Nomograms for determination of the inclination of the resultant hip force with respect to the vertical ϑ_R (C=6cm).

midstance phase of gait is related linearly to peak hip stresses in all phases of gait, as well as to some other similar activities such as adduction, external rotation and flexion [12]. This implies that the peak hip stress for all phases of slow gait can be estimated from known values of the peak hip stress p_{max} in static one legged stance by linear interpolation [12].

It should be noted that the nomograms for determination of maximal stress in the hip joint p_{max} from the known values of R, r, W_b and $\vartheta_R+\vartheta_{CE}$ given in Figs. 8–10 can also be used for estimation of p_{max} in successive phases of gait if the values R and ϑ_R are determined separately by using the opto-electronic system, piezoelectric force plate and inverse dynamics mathematical model with optimisation algorithm, as described in detail elsewhere [2,23,24]. If the resultant hip force is obtained directly from these external laboratory measurements [2,24,23,34] the peak stress may be determined from Figs. 8–10 using only steps 4 and 5 (see Appendix A).

Sometimes, especially in studies of data from the archives, information about body weight is not available. In such cases the normalised quantity p_{max}/W_b is also useful in the evaluation of the biomechanical status of the hip.

The geometrical parameters can be obtained from the rentgenographs directly by using a ruler (for the distances) and Mose circles [32] (for the radius of the hip joint articular surface). If the required tools are available, the image can also be represented by the profiles which are then transformed into a digital form and processed by the computer [25].

Determination of stress in the hip joint can be of help in deciding for a treatment and in planning the operation [3,12,19]. In order to propose an optimal outcome, clini-

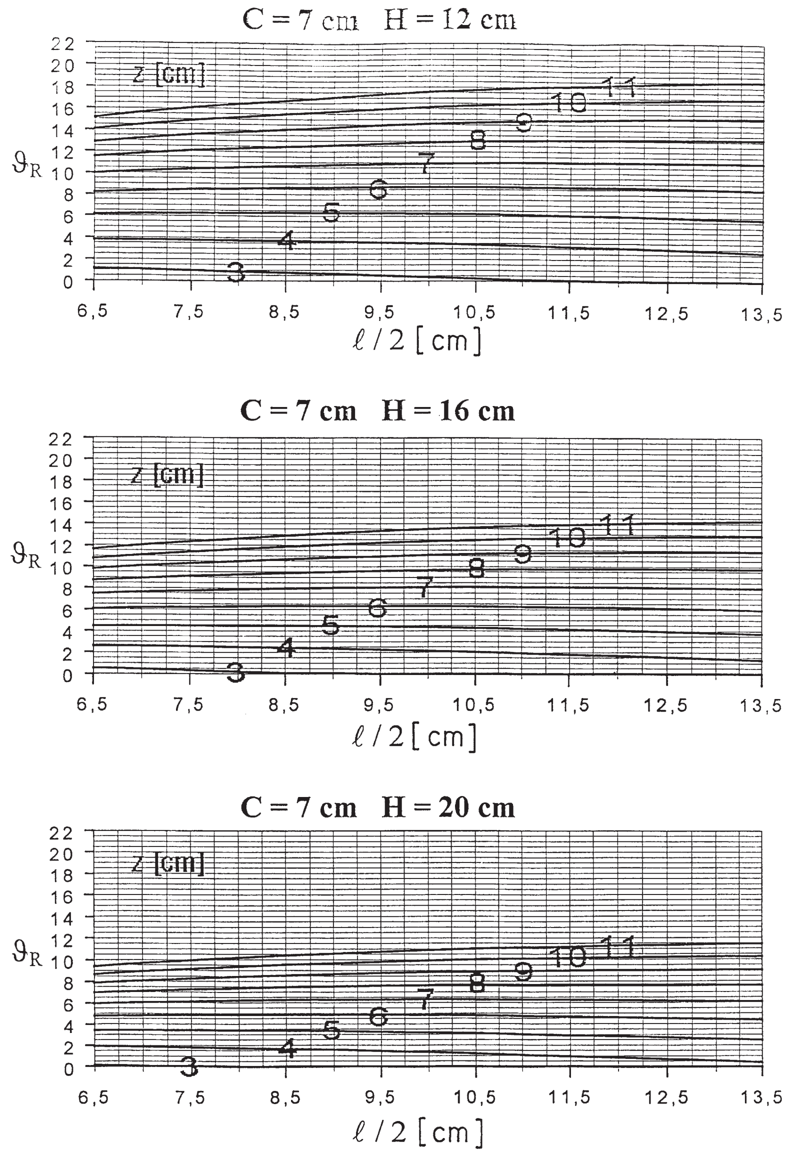

Fig. 6. Nomograms for determination of the inclination of the resultant hip force with respect to the vertical ϑ_R (C=7 cm).

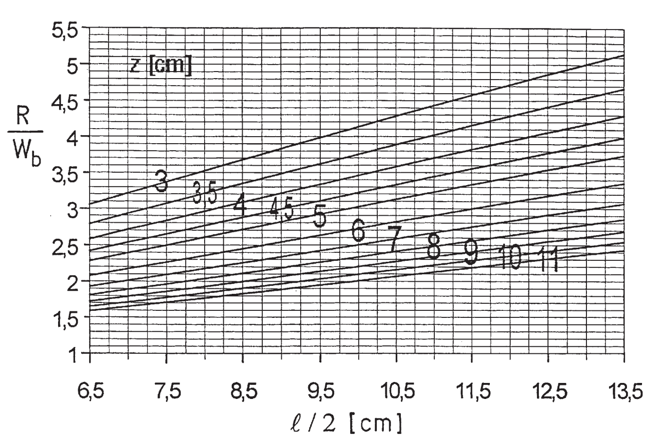

Fig. 7. Nomogram for determination of the magnitude of the resultant hip force normalised by body weight R/W_b.

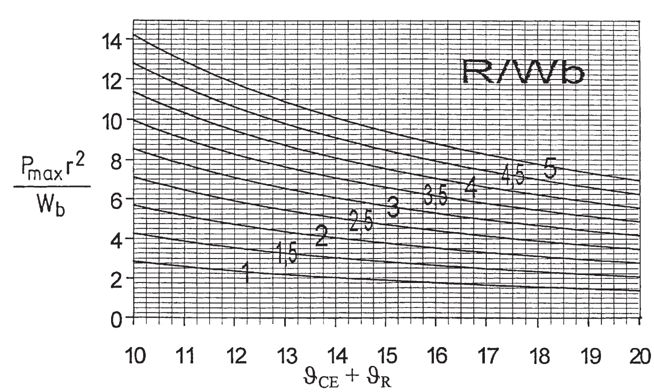

Fig. 8. Nomogram for determination of the maximal stress on the weight bearing area divided by body weight and multiplied by the square of the femoral head radius $p_{max}r^2/W_b$ (the angle (ϑ_R+ϑ_{CE}) is between 10 and 20°).

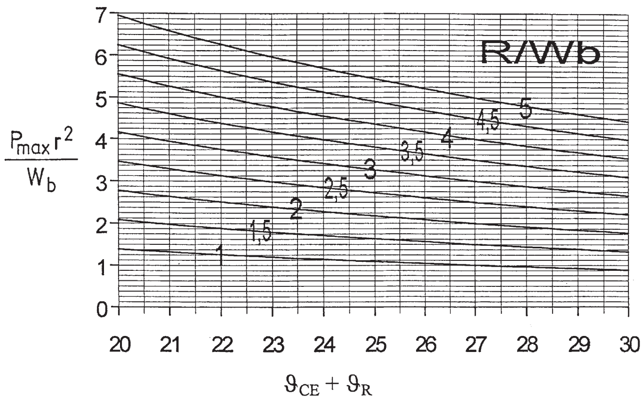

Fig. 9. Nomogram for determination of the maximal stress on the weight bearing area divided by body weight and multiplied by the square of the femoral head radius $p_{max}r^2/W_b$ (the angle (ϑ_R+ϑ_{CE}) is between 20 and 30°).

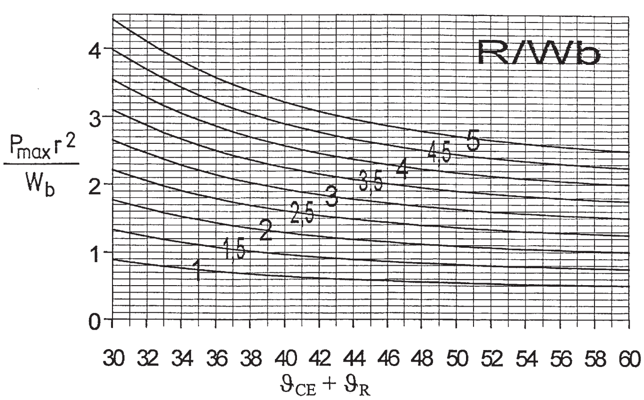

Fig. 10. Nomogram for determination of the maximal stress on the weight bearing area divided by body weight and multiplied by the square of the femoral head radius $p_{max}r^2/W_b$ (the angle (ϑ_R+ϑ_{CE}) is between 30 and 60°).

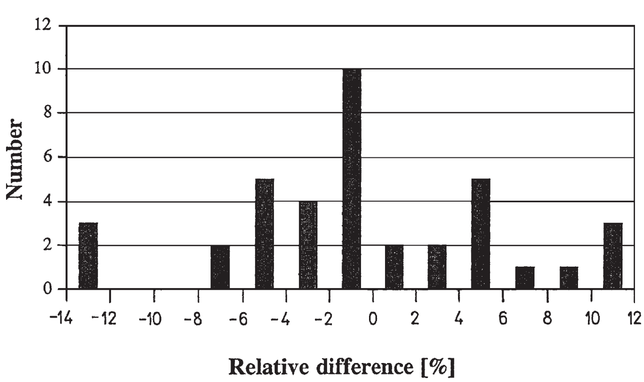

Fig. 11. Histogram of the relative difference between the values of p_{max} determined by the computer program HIPSTRESS and p_{max} determined manually using the nomograms given in Figs. 2–10. The analysis was performed for 38 persons.

cal studies have to be performed to evaluate the relevance of a particular mathematical model used in the determination of stress. Preliminary results indicate that the maximal stress p_{max}, calculated by the method proposed in this work is useful in determining the biomechanical status of the hip. We found recently [35] that the relative maximal stress p_{max}/W_b is considerably higher in the population of dysplastic hips than in the population of healthy hips, which is consistent with the results of other studies [10,30]. Also, the relative maximal stress p_{max}/W_b was found to be higher in healthy women than in healthy men [22]. As women have a higher incidence of arthrosis [11,37] these results favour the hypothesis that elevated stress in the hip joint can be one of the reasons for development of arthrosis in the female population [25]. As the effect of body weight is a parameter that can be to some extent regulated in healthy subjects, by avoiding obesity [11] the female population could take advantage of a factor that is in favour of keeping the peak stress as low as possible [22].

In conclusion, we present the nomograms from which the peak stress on the weight bearing area in one legged stance can be determined if the geometrical parameters (given in Fig. 1) and the body weight are known. The geometrical parameters can be obtained from a standard antero-posterior rentgenograph. We outlined this method because there is a large amount of data in the form of standard antero-posterior rentgenographs available from the archives.

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Appendix A

Detailed step-by-step instructions for determination of the maximal value of stress in the hip joint from the presented nomograms are given below.

1. In the first step we determine the geometrical parameters C, H, z, $l/2$, ϑ_{CE} and r (Fig. 1). A standard antero-posterior rentgenograph can be used. We determine the body weight W_b.
2. In the second step we determine the inclination of the resultant hip force ϑ_R from Figs. 2–6. We chose the figure corresponding to a pelvic width C that is closest to the measured value of C, and a graph in this figure that corresponds to a pelvic height H that is closest to the measured value of H. We chose the curve on the graph that corresponds to the horizontal coordinate of the effective muscle attachment point on the

greater trochanter z that is closest to the measured value of z. We take into account the value of the interhip distance $l/2$ on the abscissa of the graph and determine the angle ϑ_R on the ordinate.
3. In the third step we determine the magnitude of the resultant hip force normalised with respect to body weight R/W_b from Fig. 7. We chose the curve on Fig. 7 that corresponds to the horizontal coordinate of the effective muscle attachment point on the greater trochanter z that is closest to the measured value of (z). We take into account the value of the interhip distance $l/2$ on the abscissa of the graph and determine the relative magnitude of the force R/W_b on the ordinate.
4. In the fourth step we determine the value of $p_{max}r^2/W_b$ from Figs. 8–10. We add the angle ϑ_R obtained in step 2 to the centre-edge angle ϑ_{CE} and chose the figure corresponding to the range that encloses the obtained sum ($\vartheta_R+\vartheta_{CE}$). On the graph we chose the curve corresponding value of R/W_b obtained in step 3. We take into account the value of the sum ($\vartheta_R+\vartheta_{CE}$) on the abscissa of the graph and determine the relative maximal stress $p_{max}r^2/W_b$ on the ordinate.
5. In the fifth step we determine the maximal hip joint stress p_{max}. We multiply the normalised value obtained in step 4 by the body weight W_b and divide it by the square of the femoral head radius (r).

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