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Effect of patient-specific model scaling on hip joint reaction force in one-legged stance – study of 356 hips

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Purpose: Estimation of hip joint loading is fundamental for understanding joint function, injury and disease. To predict patient-specific hip loading, a musculoskeletal model must be adapted to the patient's unique geometry. By far the most common and cost effective clinical images are whole pelvis plain radiographs. This study compared the accuracy of anisotropic and isotropic scaling of musculoskeletal model to hip joint force prediction by taking patient-specific bone geometry from standard anteroposterior radiograms. *Methods*: 356 hips from 250 radiograms of adult human pelvis were analyzed. A musculoskeletal model was constructed from sequential images of the Visible Human Male. The common body position of one-legged stance was substituted for the midstance phase of walking. Three scaling methods were applied: a) anisotropic scaling by interhip separation, ilium height, ilium width, and lateral and inferior position of the greater trochanter, b) isotropic scaling by pelvic width and c) isotropic scaling by interhip separation. Hip joint force in one-legged stance was estimated by inverse static model. *Results*: Isotropic scaling affects all proportions equally, what results in small difference in hip joint reaction force among patients. Anisotropic hip scaling increases variation in hip joint force among patients considerably. The difference in hip joint force estimated by isotropic and anisotropic scaling may surpass patient's body weight. *Conclusions*: Hip joint force estimated by isotropic scaling depends mostly on reference musculoskeletal geometry. Individual's hip joint reaction force estimation could be improved by including additional bone geometrical parameters in the scaling method.

Key words: *hip joint, scaling, radiogram, joint load*

1. Introduction

Estimation of hip joint loading is fundamental for understanding joint function, injury and disease. Since the measurement of forces acting in the hip joint is technically complex and invasive [4], a majority of studies estimate hip forces by mathematical models [7], [19]. Mathematical models use measured external forces, kinematics and electromyography as input and incorporate three-dimensional geometry of muscles and bones [17], [24]. To predict patient-specific hip loading, a musculoskeletal model should be adapted to conform to the particular configuration of the patient. Scaling within the musculoskeletal model is primarily based on three-dimensional positions of anatomical landmarks in most clinical systems [19]. Each bone in the model is rescaled by either a single general rescaling factor or by multiple rescaling factors specific to individual bones [21]. Muscle attachment points are adapted according to the bone scaling.

An alternative method of scaling is based on creation of a patient-specific musculoskeletal model from computer tomography (CT) or magnetic resonance

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(MR) scans. Patient-specific models based on MR scans predict considerably different muscle-tendon lengths [21], muscle moment arms [19], [21], and muscle forces [6] compared to the models based on generic scaling. However, patient-specific CT or MR-based models include a limited number of patients and are seldom implemented in clinical practice due to the high cost of MR scan and processing time [6], [24]. It was suggested that scaling by geometrical parameters obtained from standard anteroposterior radiograms could improve the accuracy of determination of the hip joint reaction force [16]. Anteroposterior radiograms have already been used to adapt musculoskeletal models to patients' geometry in several clinical studies [14], [18]. There are several methods of scaling based on anteroposterior radiograms, the most common is the isotropic (uniform) scaling of bones [21]. Isotropic scaling is based on measuring the characteristic dimension of bone, e.g., pelvic width, and assuming that the same bones are alike in shape in all patients. However, the bone dimensions vary between patients and it should be more accurate to apply anisotropic scaling, e.g., use scaling factors specific to anatomic directions. We hypothesize that the computed joint force would be more sensitive to input geometry at anisotropic scaling than at isotropic scaling. The aim of the study is to verify this hypothesis and to quantify the difference between isotropic and anisotropic scaling using anteroposterior radiograms as reference.

2. Materials and methods

2.1. Patient-specific data

250 radiograms of adult human hips were obtained from the archive of the Department of Orthopaedic Surgery and the Department of Traumatology, Ljubljana University Medical Centre, Ljubljana, Slovenia. 356 hips were included in the study. Only radiograms with the entire pelvis and proximal femora visible and non-luxated hips were included in the study. 10% magnification of all radiograms was assumed. The study did not require approval by the local ethical committee but it was performed in accordance with the ethical standards of the 1964 Declaration of Helsinki as revised in 2000.

The hip joint center (HJC) was chosen at the center of the circle fitting of the bony surface of the femoral head. The geometry of an individual hip was described by the following parameters (Fig. 1): interhip separation l (the distance between the left and right HJC), iliac height H (the vertical distance between the most superior point on the ilium and HJC), iliac width C (the horizontal distance between the most lateral point on the ilium and HJC) and the coordinates of the effective muscle insertion point on the greater trochanter in the femoral coordinate system (T_y and T_z, respectively). The attachment of the effective muscle on the greater trochanter was defined by intersection of the contour of the greater trochanter and the midpoint normal to the straight line connecting the most lateral and the most superior points of the greater trochanter. The values of the pelvic and femoral geometrical parameters l, H, C, T_y, and T_z were determined in the reference frame where the straight line connecting the left and the right HJC defines the horizontal line, by using a computer-assisted technique [14].

2.2. Reference musculoskeletal model

Contours of hip bones were segmented manually from the cryo-section images of a male cadaver of the Visible Human Project [1] using Seg3D software (Scientific Computing and Imaging Institute, Salt Lake City, UT). Bone geometry was visualized by OpenDX,

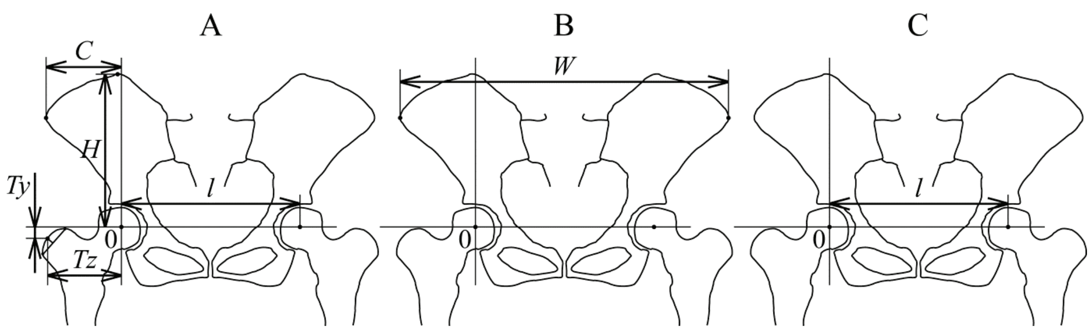

Fig. 1. Scaling methods: A – anisotropic scaling by: interhip separation l, height of the ilium H, width of the ilium C and inferior and lateral position of grater trochanter T_y and T_z, respectively. B – isotropic scaling by the width of the pelvis $W = l + 2C$ and C – isotropic scaling by the interhip separation l

an open-source visualization software package [13]. The reference values of geometrical parameters of the hip l_0, H_0, C_0, T_{y0} and T_{z0} were obtained from the radiogram created by projection of CT slices from Visible Human Project onto the frontal plane.

Cartesian coordinate system after Wu et al, 2002 was adopted. The origin of coordinate system is at left or right HJCs, horizontal z-axis points laterally, y-axis points superiorly and x-axis points anteriorly [25].

Reference muscle modeling is based on the work of Delp et al. [11]. The model includes 27 active muscle units of the hip. Coordinates of proximal $r_p = (x_p, y_p, z_p)$ and distal $r_d = (x_d, y_d, z_d)$ muscle attachment points are defined in the bone model [9]. Muscles are modeled as straight lines connecting the attachment points r_p and r_d.

To avoid unrealistic muscle action, muscle paths are modified by adding wrapping points if necessary. Wrapping points are obtained automatically from extracted bone geometry using a convex wrapping algorithm [12]. Maximum isometric force of each muscle unit is taken from Brand et al, 1986.

2.3. Scaled musculoskeletal model

Reference musculoskeletal model was scaled to an individual using the patient-specific data from anteroposterior radiograms. Any point in the reference model that can describe either bone surface or muscle attachment was adjusted for a given patient by multiplication with the corresponding scaling matrix M in x plane. No scaling was done in the x-direction as the standard anteroposterior radiogram contains no implicit depth information. Muscle insertions on the distal femur were not scaled since the distal femur is not included in the standard anteroposterior radiogram.

Three types of model scaling were defined: Type A (anisotropic scaling, Fig. 1) was proposed by Iglič et al. [14], where the pelvis and femur are scaled separately and anisotropically. Scaling matrix $M_P = \left[\frac{H}{H_0}, \frac{C}{C_0}\right]$ was taken for the pelvis and the proximal muscle attachment points, while scaling matrix $M_F = \left[\frac{T_y}{T_{y0}}, \frac{T_z}{T_{z0}}\right]$ was taken for the proximal femur and distal attachment points. Within method B (isotropic scaling, Fig. 1), the distances are scaled by pelvic width $W = l + 2C$. Pelvic and proximal femur coordinates were scaled equally by setting $M = \frac{W}{W_0}$, where $W_0 = l_0 + 2C$. Within method C (isotropic scaling, Fig. 1), distances are scaled by the interhip separation l. The pelvis and proximal femur were scaled equally by setting $M = \frac{l}{l_0}$.

2.4. Hip joint force calculation

One-legged stance representing midstance phase of walking is conveniently taken as a representative body position [10]. It was assumed that the hip joint reaction force passes through HJC, the coordinate system origin. Muscle activity is required to maintain equilibrium in one-legged stance. Equations expressing equilibrium of forces and torques in one-legged stance were taken into account. Three torque equilibrium equations containing 27 unknown muscle forces present a statically indeterminate problem and muscle forces are estimated by inverse dynamic optimization [23]. The optimal muscle activation is found by minimization of the sum of muscle stresses cubed [8] while the force of each muscle is constrained by imposing an upper bound of isometric force. After computation of the muscle forces, the components of the hip joint reaction force F_R are determined from the force equilibrium. The hip joint reaction force is calculated in relation to body weight F_R/BW.

Statistics

Statistical analyses were conducted using the R Statistical Software (Foundation for Statistical Computing, Vienna, Austria). Sample normality was tested by the Shapiro–Wilk test and inspected using q-q plots. Descriptive statistics were performed for the study, and continuous variables were presented as means ± standard deviation (SD). Differences between groups were determined using the Student's t-test and relationships between variables were described using Pearson coefficient. Level $p < 0.05$ was considered statistically significant.

3. Results

Anisotropic scaling A yields considerably higher variance in hip joint reaction force than isotropic scaling B or C where the variance in hip joint reactions force is minute (Fig. 2). Anisotropic scaling A predicts lower average hip load than isotropic scaling B and C that is statistically significant (paired t-test). The peak difference between the hip joint reaction force was obtained by anisotropic scaling A and by isotropic scaling B in some patients surpassed 1x. Isotropic scaling assumes that bones have mutually proportional dimensions between patients.

Although the hips with larger interhip separation l have significantly (R = 0,42, p = 0.001) larger iliac height H, no correlation was found between the interhip separation l and the iliac width C (R = 0.05, p = 0.05) nor lateral coordinate of greater trochanter T_z (R = –0.01, p = 0.05), Table 1. All pelvic and femoral geometrical parameters l, H, C, T_y, and T_z were normally distributed (Shapiro–Wilk test, $p < 0.05$).

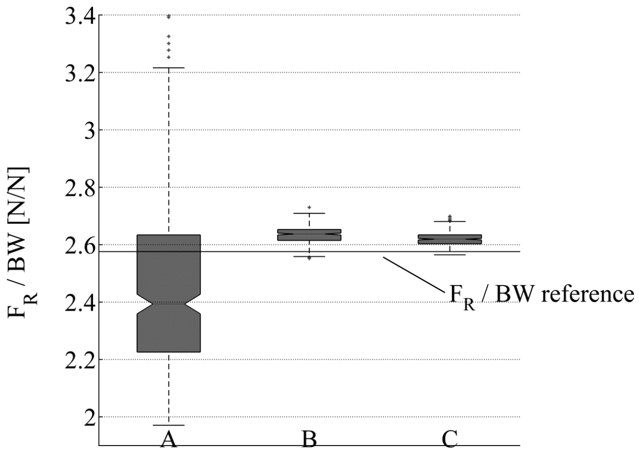

Fig. 2. Ranges of calculated hip joint reaction force for A – anisotropic scaling, B – isotropic scaling by the pelvic width, C – isotropic scaling by the interhip separation

of one-legged stance. It was found that including patient-specific geometry in more detail considerably affects the hip joint reaction force.

The hip joint reaction force computed using isotropically scaled models (type B and C) is nearly insensitive to the changes in input geometrical parameters in one-legged stance.

Isotropic scaling modifies the lever arm of the body weight by increasing or decreasing the interhip separation, however, the iliac width and trochanter position are changed in the same proportion [14]. The change in torque of gravitational forces is compensated by the change in muscle moment arms. Therefore, the level of hip joint force in one-legged stance depends on geometry of reference model mostly if isotropic scaling of both pelvic and femur by the same scaling factor is adopted. Since the same reference model was used in all patients, variation in hip joint force is negligible for isotropic scaling.

Isotropic hip scaling (B and C) yields a higher average magnitude of hip joint reaction force than anisotropic scaling A. However, this could be an effect of reference muscle geometry. A reference muscle model derived from the Visible Human Project has lower muscle moment arms than average values of the population considered (Table 1) and thus it enables prediction of greater hip force.

As shown in Table 1, the overall size of pelvic bone is not directly correlated to the positions of bone prominences serving as muscle attachment points. Hence, scaling based on overall bone size does not enable to accurately predict muscle moment arms and position of the center of rotation as also shown in previous studies [6], [15]. Including the inter-individual variations in pelvic and proximal femur dimensions in anisotropic scaling increases range of predicted hip joint force significantly.

Table 1. Reference values, average values, standard deviations and coefficients of correlations R between different geometrical parameters. Statistically significant correlations are marked by * for $p < 0.05$ and ** for $p < 0.001$

				R				
Parameter	Reference	Mean [mm]	StDev [mm]	l	H	C	T_z	T_y
l	172	202.2	18.2	1.00	0.42**	0.05	–0.01	0.29**
H	158	152.5	11.4		1.00	0.28**	0.19*	0.05
C	45	58.6	10.2			1.00	0.33**	–0.18*
T_z	51	60.9	7.9				1.00	–0.11*
T_y	–23	–10.8	6.2					1.00

4. Discussion

This study analyzes the effects of anisotropic and isotropic scaling methods on calculated hip joint reaction forces subjected to patient-specific geometry. Geometrical parameters were obtained from standard anteroposterior radiograms while the force was calculated by a three dimensional musculoskeletal model

The radiogram is a planar projection and does not allow for anteroposterior adjustment of parameters. Including anteroposterior scaling in general provides more accurate estimation of the hip joint reaction force, however, neglecting anteroposterior adjustment is not likely to have a significant effect on the resultant hip force in one-legged stance. Namely, in this body position, abductor activity is required to compensate gravitational torque acting in the frontal plane [14] while components of moment arms in the frontal plane depend on mediolateral and superior-inferior position of muscle attachment points [14]. In dynamic loading, the torques of muscle forces attains various magnitudes and directions [4] and proper anteroposterior scaling might be more important.

The present study focuses on the effects of patient-specific geometry on hip joint force when the muscle parameters are kept unchanged. To advance the description, the muscle parameters, such as tendon length, optimal length and maximum muscle force should also be scaled [7], [17].

As shown in previous studies [6], [15], [20], [21] and confirmed in the present study, the accuracy of predicting the hip joint reaction force by simplified musculoskeletal scaling is limited. The accuracy of hip joint force estimation could in principle be improved if more information on patient-specific geometry were included. For example, pelvic width obtained from bony landmarks measurements (ASIS) can be used in combination with hip joint center measurements obtained from motion analysis [2]. This method allows for more accurate estimation of iliac width *C*. The same method can be applied to assessment of trochanter position estimated by external or radiographic measurements. The number of measured parameters should be further optimized to achieve accurate and effective scaling of musculoskeletal model.

5. Conclusions

The scaling based on gross bone dimension (isotropic scaling) does not scale muscle moment arms accurately and does not predict hip load accurately. Hip joint force estimated by isotropic scaling depends mostly on reference musculoskeletal geometry. To improve estimation of the hip joint reaction force in an individual, determination of more hip and pelvic geometrical parameters should be included in the scaling method (anisotropic scaling). The difference in estimated hip joint force between isotropic and anisotropic scaling may be as high as patient's body weight.

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References

[1] ACKERMAN M.J., *Accessing the Visible Human Project*, D-Lib Magazin, 1995, 1(4), DOI: 10.1045/october95-ackerman.
[2] ARBABI E., SCHMID J., BOULIC R., THALMANN D., MAGNENAT-THALMANN N., *Sensitivity of hip tissues contact evaluation to the methods used for estimating the hip joint center of rotation*, Medical & Biological Engineering & Computing, 2012, 50(6), 595–604, DOI: 10.1007/s11517-012-0873-0.
[3] ARNOLD A.S., BLEMKER S.S., DELP S.L., *Evaluation of a Deformable Musculoskeletal Model for Estimating Muscle-Tendon Lengths During Crouch Gait*, Annals of Biomedical Engineering, 2001, 29(3), 263–274. DOI: 10.1114/1.1355277
[4] BERGMANN G., *Hip 98: Loading of the hip joint*, 2001.
[5] BRAND R.A., PEDERSEN D.R., FRIEDERICH J.A., *The sensitivity of muscle force predictions to changes in physiologic cross-sectional area*, Journal of Biomechanics, 1986, 19(8), 589–596, DOI: 10.1016/0021-9290(86)90164-8.
[6] CORREA T.A., BAKER R., KERR G.H., PANDY M.G., *Accuracy of generic musculoskeletal models in predicting the functional roles of muscles in human gait*, Journal of Biomechanics, 2011, 44(11), 2096–2105, DOI: 10.1016/j.jbiomech.2011.05.023.
[7] CORREA T.A., PANDY M.G., *A mass–length scaling law for modeling muscle strength in the lower limb*, Journal of Biomechanics, 2011, 44(16), 2782–2789, DOI: 10.1016/j.jbiomech.2011.08.024.
[8] CROWNINSHIELD R.D., BRAND R.A., *A physiologically based criterion of muscle force prediction in locomotion*, Journal of Biomechanics, 1981, 14(11), 793–801, DOI: 10.1016/0021-9290(81)90035-x
[9] DANIEL M., IGLIČ A., KRALJ-IGLIČ V., KONVIČKOVÁ S., *Computer system for definition of the quantitative geometry of musculature from CT images*, Computer Methods in Biomechanics and Biomedical Engineering, 2005, 8(1), 25–29, DOI: 10.1080/10255840500110952.
[10] DEBEVEC H., PEDERSEN D.R., IGLIČ A., DANIEL M., *One-Legged Stance as a Representative Static Body Position for Calculation of Hip Contact Stress Distribution in Clinical Studies*, Journal of Applied Biomechanics, 2010, 26(4), 522–525, DOI: 10.1123/jab.26.4.522
[11] DELP S.L., LOAN J.P., HOY M.G., ZAJAC F.E., TOPP E.L., ROSEN J.M., *An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures*, IEEE Transactions on Biomedical Engineering, 1990, 37(8), 757–767. DOI: 10.1109/10.102791
[12] DESAILLY E., SARDAIN P., KHOURI N., YEPREMIAN D., LACOUTURE P., *The convex wrapping algorithm: A method for identifying muscle paths using the underlying bone mesh*, Journal of Biomechanics, 2010, 43(13), 2601–2607. DOI: 10.1016/j.jbiomech.2010.05.005
[13] *IBM Visualization Data Explorer User's Guide*, International Business Machines Corporation, New York, USA, version 3 edition, 1997.
[14] IGLIČ A., KRALJ-IGLIČ V., DANIEL M., MAČEK-LEBAR A., *Computer Determination of Contact Stress Distribution and Size of Weight Bearing Area in the Human Hip Joint*, Com-

puter Methods in Biomechanics and Biomedical Engineering, 2002, 5(2), 185–192, DOI: 10.1080/10255840290010300.

[15] LENAERTS G., BARTELS W., GELAUDE F., MULIER M., SPAEPEN A., VAN DER PERRE G. et al., *Subject-specific hip geometry and hip joint centre location affects calculated contact forces at the hip during gait*, Journal of Biomechanics, 2009, 42(9), 1246–1251, DOI: 10.1016/j.jbiomech.2009.03.037.

[16] LENAERTS G., DE GROOTE F., DEMEULENAERE B., MULIER M., VAN DER PERRE G., SPAEPEN A. et al., *Subject-specific hip geometry affects predicted hip joint contact forces during gait*, Journal of Biomechanics, 2008, 41(6), 1243–1252, DOI: 10.1016/j.jbiomech.2008.01.014.

[17] MENEGALDO L.L., DE OLIVEIRA L.F., *Effect of muscle model parameter scaling for isometric plantar flexion torque prediction*, Journal of Biomechanics, 2009, 42(15), 2597–2601, DOI: 10.1016/j.jbiomech.2009.06.043.

[18] RECNIK G., KRALJ-IGLIČ V., IGLIČ A., ANTOLIČ V., KRAMBERGER S., RIGLER I. et al., *The role of obesity, biomechanical constitution of the pelvis and contact joint stress in progression of hip osteoarthritis*, Osteoarthritis and Cartilage, 2009, 17(7), 879–882, DOI: 10.1016/j.joca.2008.12.006.

[19] SCHEYS L., VAN CAMPENHOUT A., SPAEPEN A., SUETENS P., JONKERS I., *Personalized MR-based musculoskeletal models compared to rescaled generic models in the presence of increased femoral anteversion: Effect on hip moment arm lengths*, Gait & Posture, 2008, 28(3), 358–365, DOI: 10.1016/j.gaitpost.2008.05.002

[20] SCHEYS L., DESLOOVERE K., SUETENS P., JONKERS I., *Level of subject-specific detail in musculoskeletal models affects hip moment arm length calculation during gait in pediatric subjects with increased femoral anteversion*, Journal of Biomechanics, 2011, 44(7), 1346–1353, DOI: 10.1016/j.jbiomech.2011.01.001.

[21] SCHEYS L., SPAEPEN A., SUETENS P., JONKERS I., *Calculated moment-arm and muscle-tendon lengths during gait differ substantially using MR based versus rescaled generic lower-limb musculoskeletal models*, Gait & Posture, 2008, 28(4), 640–648. DOI: 10.1016/j.gaitpost.2008.04.010

[22] SEIDEL G.K., MARCHINDA D.M., DIJKERS M., SOUTAS--LITTLE R.W., *Hip joint center location from palpable bony landmarks – A cadaver study*, Journal of Biomechanics, 1995, 28(8), 995–998, DOI: 10.1016/0021-9290(94)00149-x

[23] TSIRAKOS D., BALTZOPOULOS V., BARTLETT R., *Inverse Optimization: Functional and Physiological Considerations Related to the Force-Sharing Problem*, Critical Reviews™ in Biomedical Engineering, 1997, 25(4–5), 371–407, DOI: 10.1615/critrevbiomedeng.v25.i4-5.20.

[24] VICECONTI M., TESTI D., TADDEI F., MARTELLI S., CLAPWORTHY G.J., JAN S.V.S., *Biomechanics Modeling of the Musculoskeletal Apparatus: Status and Key Issues*, Proceedings of the IEEE, 2006, 94(4), 725–739, DOI: 10.1109/jproc.2006.871769.

[25] WU G., SIEGLER S., ALLARD P., KIRTLEY C., LEARDINI A., ROSENBAUM D. et al., *ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion*. Part I: *ankle, hip, and spine*, Journal of Biomechanics, 2002, 35(4), 543–548, DOI: 10.1016/s0021-9290(01)00222-6.