学位論文

# Femoral head translation in dysplastic hips

# (臼蓋形成不全股における股関節不安定性の評価)

旭川医科大学大学院医学系研究科博士課程
医学専攻感覚器運動器病態学領域

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# Dynamic femoral head translations in dysplastic hips

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ABSTRACT

*Background:* Developmental dysplasia of the hip is an important disease leading to osteoarthritis. Recently, researchers have focused on hip instability as a potentially important dynamic factor for osteoarthritis, but the detailed kinematics of dysplastic hips during weight-bearing gait have not been reported. The purpose of this research is to contrast femoral translation in contralateral healthy hips and dysplastic hips during weight-bearing stepping.

*Methods:* Twelve dysplastic hips and eight healthy hips were investigated. Hip joint kinematics were analyzed using 3D-2D model-image registration with dynamic fluoroscopic images of each hip during a stepping-in-place activity. Femoral translation relative to the acetabular center was quantified as instability.

*Findings:* Total femoral head translations were significantly different between dysplastic and contralateral healthy hips. Mean translation was 1.0 mm in dysplastic hips and 0.4 mm in contralateral healthy hips during swing-phase, and consisted of inferior translation during early swing phase with a complementary superior translation just before foot strike. Total femoral translation was significantly correlated to several radiographic indices of hip dysplasia.

*Interpretation:* Superior translations of the femur during the end of swing phase may result in altered articular contact mechanics, abnormal stresses on the labrum and lost lubricant sealing. All of these factors may contribute to joint degeneration and osteoarthritis in dysplastic hips.

## 1. Introduction

Developmental dysplasia of the hip (DDH) is an important disease leading to osteoarthritis (OA) (Hasegawa et al., 1992; Jacobsen and Sonne-Holm, 2005; Lane et al., 2000), and it is well established that high stress concentrations during weight-bearing ambulation cause OA (Mavcic et al., 2008; Pompe et al., 2007). Historically, static geometric factors on radiography, e.g. center-edge (CE) angle (Wiberg, 1939), Sharp angle (Sharp, 1961), acetabular roof obliquity (ARO)(Massie and Howorth, 1950), and acetabular head index (AHI)(Heyman and Herndon, 1950) have been used to quantify risk of OA in hips with DDH. Recently, researchers have focused on hip instability as a potentially important dynamic factor for OA (Akiyama et al., 2011; Maeyama et al., 2008; Philippon et al., 2007), but detailed kinematics of DDH hips during weight-bearing gait have not been reported. The aim of this study was to contrast femoral translation in DDH and contralateral healthy hips during weight-bearing stepping. We performed dynamic radiographic imaging of dysplastic and contralateral healthy hips and quantified their kinematics using model-image registration techniques. We wished to test two hypotheses: First, because the DDH hip has less geometric coverage, we assumed femoral head translations in DDH hips would be larger than contralateral healthy hips. Second, we assumed the direction of femoral translation would be superolateral during stance-phase and inferoposterior during swing-phase due to the net joint forces during those phases of gait (Ferguson et al., 2000; Fessler, 1957; Henak et al., 2011; Klaue et al., 1991; McCarthy et al., 2003; Shu and Safran, 2011).

## 2. Methods

### 2.1. In vivo study

Thirteen DDH subjects who were scheduled to have osteotomy in our hospital, including ten females, gave informed consent to participate in this IRB approved study. Subjects' mean age was 34 years (range: 19 to 44 years). Five subjects had bilateral dysplastic hips and eight had unilateral dysplastic hips. Diagnosis of DDH was based on anteroposterior radiography with a CE angle < 20° (Wiberg, 1939).

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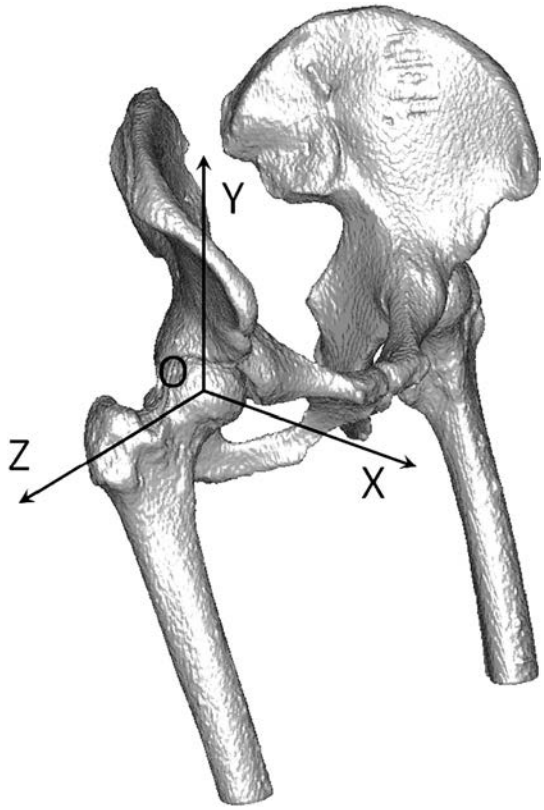


**Fig. 1.** Femur and pelvis 3D models were reconstructed from CT scans. A coordinate system was established so that the origin was at the center of the acetabulum, and the femur and pelvis were coincident in the non-weightbearing supine position.

Sharp angle, ARO, and AHI also were measured by anteroposterior radiography. Exclusion criteria included pregnancy, history of systemic disease, hip injury, hip operation, any other hip disease except DDH, hip joint contracture, radiographical evidence of osteoarthritis (Tonnis grade > 1) (D, 1987), or a deformed femoral head. After exclusions, 12 dysplastic hips and 8 contralateral healthy hips were available for study.

Pelvis and femur geometry models were developed from computed tomography (CT) images of each hip (Aquilion, Toshiba Medical Systems, Japan; 1 mm slice thickness, 512 × 512 acquisition matrix, 0.732 mm pixel spacing). Three-dimensional mesh models were created using open-source segmentation software (itk-SNAP, Penn Image Computing and Science Laboratory, Philadelphia, PA) (Yushkevich et al., 2006). Standard coordinate systems were placed in the pelvic models following published conventions (Geomagic Studio, Geomagic Inc.) (Wu, 2002) and the same reference coordinate system was applied to femur (Fig. 1). The femoral head center was determined as the center of a best fit sphere (Pratt, 1987), and the acetabular center was defined coincident with the femoral head center in the CT images. Therefore, the distance between acetabular and femoral center was zero at the position for CT scan.

Each hip was imaged using anteroposterior fluoroscopy (ARCADIS Avantic, Siemens AG, German) at 15 frames per second, which was the maximum speed of our system, while the subject walked in place (Imaging parameters: 8 ms pulse width, 250 mA tube current, 70 kVp voltage, 1000 mm source-image-distance and 300 mm diameter field of view). Subjects repeatedly simulated a gait motion, including full weight-bearing and swing phases, without forward progression. No subject claimed hip discomfort or pain during imaging. We extracted images corresponding to one gait cycle step. A gait cycle and fluoroscopy images were synchronized using a digital camera movie in order to determine how the stance phase events corresponded to the appearance of the bones in the fluoroscopic images. Images were expressed in normalized time from one foot strike (0% step cycle) to the next strike of the same foot (100% step cycle).

Hip joint kinematics were quantified using open-source software for 3D-to-2D model-image registration (JointTrack, www.sourceforge.net/projects/jointtrack) (Banks and Hodge, 1996; Dennis et al., 2001). Briefly, bone models were superimposed on distortion-corrected digitized fluoroscopic images and moved in a virtual 3D space until their projections matched the recorded images. A single experienced user interactively adjusted the bones' silhouettes to match their fluoroscopic silhouettes, and could then refine the registration using global or local numerical optimization. It took 3–4 min for both manual measurement and numerical optimization per each image. The measurement software provides a 3D rendering of the bones, and graphs of the kinematics over time (Fig. 2), providing the user contextual information helping to

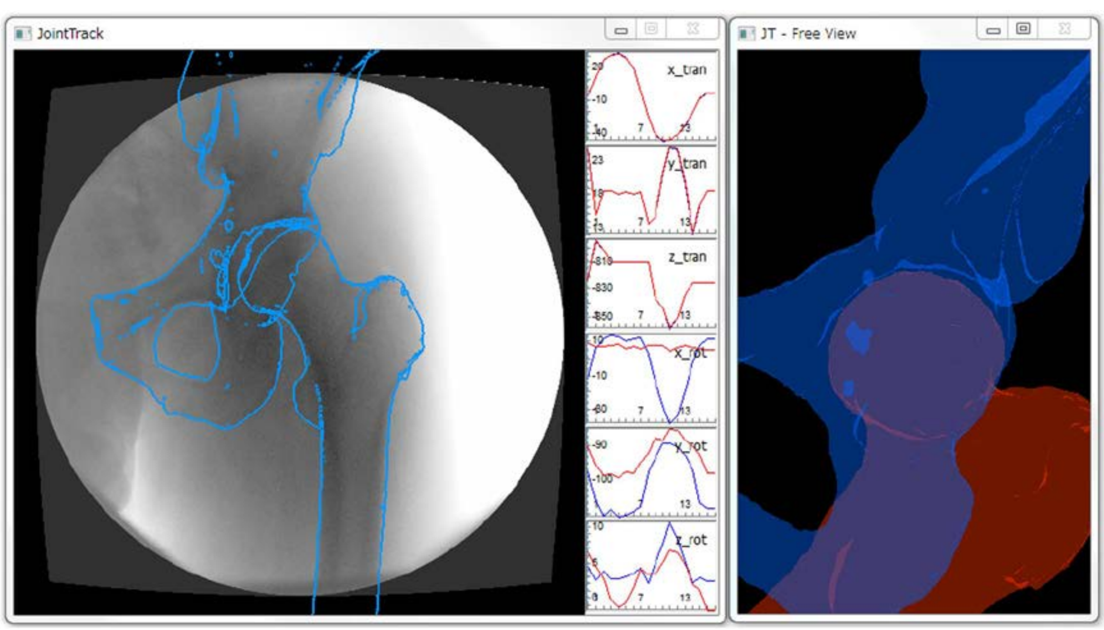


**Fig. 2.** The model-image registration software shows views of the radiographic image with the superimposed bone models, graphs of the (x, y, z) translations and rotations, and a user adjustable view providing any aspect and zoom of the current joint configuration.

avoid obvious bone inter-penetration. Measured translations were expressed as the femur moving with respect to the pelvis, referenced to their relative alignment in the CT scan. Femoral translation, a potential measure of instability, was quantified as the displacement of the femoral head center relative to the acetabular center in each image. We report both the instantaneous distance between femoral head and acetabular centers, as well as the maximum displacement of the femoral head between any two images during the stepping cycle. Group averaging was performed by resampling each subject's data at each 5% increment of the stepping cycle using linear interpolation.

Kinematic results from DDH and contralateral healthy hips were compared using two-way repeated-measures ANOVA with post hoc Tukey tests for pair-wise comparisons (Dr. SPSS II for Windows version 11.0.1J, SPSS JAPAN Inc., Tokyo, Japan). Linear regression analysis was performed to compare hip translations with standard static geometric parameters: CE angle, Sharp angle, ARO and AHI. The significance level for all analyses was $p < 0.05$.

### 2.2. In vitro study

An in vitro validation study was performed to determine the accuracy of single-plane model-image registration for this application (see Supplementary Materials). A cadaver specimen was prepared with metallic beads in the acetabulum and femur, and stereo radiographic views were obtained. One of the views was the same AP view used in the in vivo study. Comparing marker-based stereophotogrammetry and model-based single-plane registration, we found RMS errors for coronal plane translations of the hip were 0.2 mm, anterior/posterior translations 0.5 mm, and 1.6° for joint rotations.

## 3. Results

There was no difference in age or sex between the DDH group and the contralateral healthy hip group (Table 1). CE angle, Sharp angle, ARO, and AHI were all significantly different between the two groups.

Femoral head translations were significantly different between DDH and contralateral healthy hips (Fig. 3a). Maximum translations averaged 1.0 mm in DDH hips and 0.5 mm in contralateral healthy hips during swing-phase. The DDH femoral head translated inferiorly immediately after toe off and remained in the same position until just before heel strike. Translations were predominantly in the superior/inferior direction, with average anteroposterior translations of 0.3 mm in each group ($p > 0.05$) (Fig. 3b,c,d). There were no significant pair-wise differences in translation components during stance phase.

Maximum translations between any two points in the step cycle were significantly greater in the DDH hips (Table 1, $p < 0.001$). Regression analyses showed significant correlations between maximum femoral head translation and CE angle, AHI and ARO. Smaller CE and AHI corresponded to larger femoral translations, and larger ARO corresponded to larger femoral translations (Fig. 4a,b,c). We did not observe significant correlations between maximum femoral translations and the Sharp angle (Fig. 4d).

## 4. Discussion

The pathogenesis of OA in DDH hips is thought to be mainly a function of reduced geometric stability and load-bearing area in the acetabulum. A variety of radiographic parameters have been used as surrogates for predicting acetabular load concentration, but the relationship between these parameters and OA causation remains unclear. Numerous studies have reported correlation between the CE angle and OA (Chung et al., 2010; Jacobsen and Sonne-Holm, 2005; Lane et al., 2000; Murphy et al., 1995; Reijman et al., 2005), while others have concluded there is no relationship (Johnsen et al., 2009; Lau et al., 1995). It is not possible currently to predict degenerative progression to OA based upon commonly accepted definitions of radiographic parameters. Dynamic hip joint instability, also thought to be characteristic of DDH, has been less studied because it is difficult to measure, and any relationships between instability and OA remain speculative. We used dynamic radiographic images and model-image registration to quantify femoral translations during a dynamic weight-bearing stepping activity. Contrary to hypothesized translations during both swing and stance phases, we observed femoral head translation only during swing phase, and we did not observe femoral superolateral translation during stance. Hips with the most abnormal radiographic indices demonstrated the largest hip translation. In these DDH hips without OA, there is femoral translation during swing phase which may relate to subsequent joint degeneration.

Several groups have reported superolateral directed forces and accelerations in DDH hips during weight bearing gait. In vivo and in vitro studies have shown hip joint forces during gait are directed superoposterior at heel strike (Bergmann et al., 2001; Endo et al., 2003;

**Table 1**
Subject demographics, radiographic hip characteristics, and maximum femoral displacement.

| Group | Subject | Sex | Side | Age | CE angle (deg) | Sharp angle (deg) | ARO[a] (deg) | AHI[b] (deg) | Maximum displacement (mm) |
|---|---|---|---|---|---|---|---|---|---|
| DDH hip | 1 | Male | Right | 38 | 4.1 | 49.3 | 26.3 | 55.2 | 2.48 |
| | 2 | Female | Right | 44 | 4.9 | 48.8 | 22.7 | 65.0 | 1.63 |
| | 3 | Female | Right | 26 | 11.6 | 51.1 | 20.7 | 77.3 | 1.46 |
| | 4 | Female | Left | 26 | 13.2 | 51.7 | 19.7 | 66.0 | 1.33 |
| | 5 | Female | Right | 28 | 13.7 | 51.6 | 23.0 | 54.0 | 1.71 |
| | 6 | Male | Right | 41 | 13.9 | 49.0 | 18.5 | 68.2 | 1.99 |
| | 7 | Female | Left | 44 | 14.2 | 46.2 | 19.2 | 71.5 | 2.26 |
| | 8 | Female | Left | 19 | 15.0 | 52.6 | 22.3 | 74.5 | 2.32 |
| | 9 | Male | Left | 35 | 15.1 | 44.1 | 15.3 | 66.7 | 2.27 |
| | 10 | Female | Right | 33 | 15.5 | 49.2 | 19.5 | 64.0 | 1.94 |
| | 11 | Female | Left | 26 | 15.9 | 43.6 | 15.2 | 69.1 | 1.63 |
| | 12 | Female | Left | 29 | 17.3 | 49.1 | 15.4 | 75.7 | 1.45 |
| Healthy contralateral hip | 1 | Male | Right | 35 | 20.2 | 42.4 | 12.8 | 76.0 | 1.21 |
| | 2 | Female | Right | 44 | 20.4 | 44.7 | 15.6 | 77.0 | 0.93 |
| | 3 | Female | Right | 29 | 22.5 | 47.0 | 17.8 | 78.0 | 0.94 |
| | 4 | Male | Left | 41 | 23.2 | 45.9 | 9.0 | 74.5 | 0.88 |
| | 5 | Female | Left | 44 | 27.0 | 43.4 | 9.7 | 87.2 | 0.92 |
| | 6 | Female | Left | 25 | 27.2 | 44.1 | 12.7 | 83.0 | 0.79 |
| | 7 | Female | Right | 42 | 27.5 | 45.7 | 18.1 | 82.1 | 1.27 |
| | 8 | Female | Right | 26 | 28.5 | 43.2 | 8.7 | 84.2 | 1.28 |
| | *t*-Test | | | $p = 0.19$ | $p < 0.001$ | $p < 0.001$ | $p < 0.001$ | $p < 0.001$ | $p < 0.001$ |

[a] Acetabular roof obliquity.
[b] Acetabular head index.

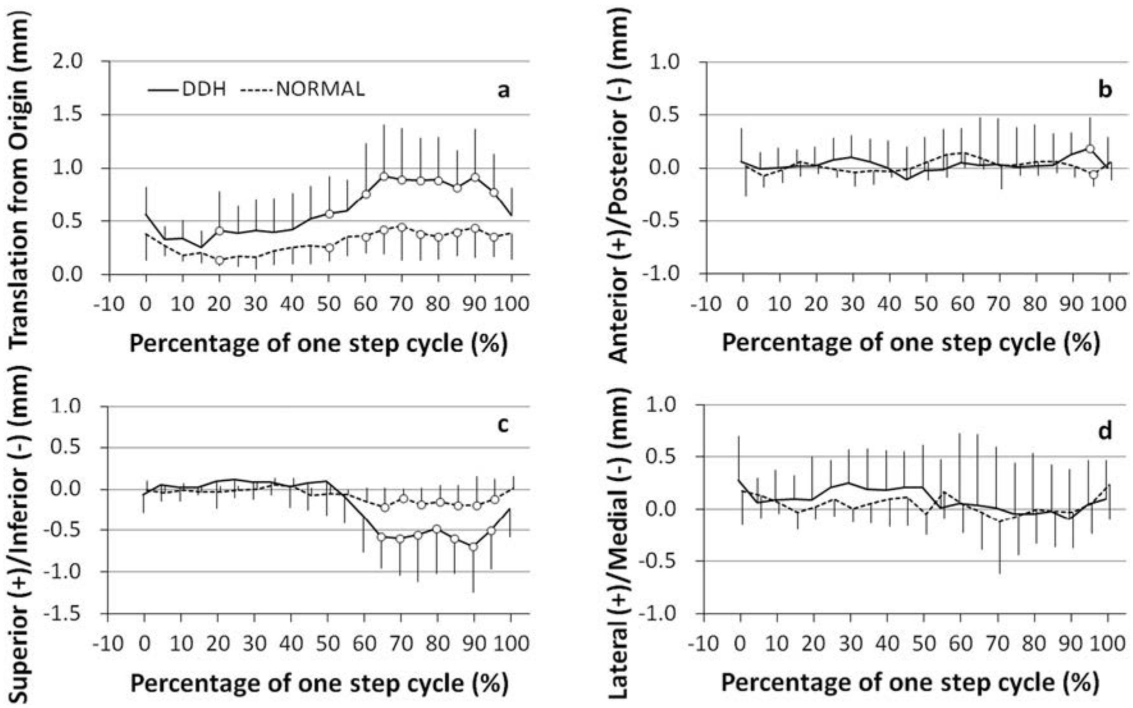


**Fig. 3.** Femoral translations from the acetabular origin during stepping-in-place in DDH and contralateral healthy hips: (a) translation magnitude, (b) anterior/posterior translation, (c) superior/inferior translation, (d) lateral/medial translation. Open circles show significant pair-wise differences between the two groups.

Yoshida et al., 2006), although we are unaware of corresponding reports of femoral translations. Maeyama et al. suggested the possibility of superolateral translation based upon skin-surface triaxial accelerometer measurements of superior, posterior, and laterally directed accelerations during gait (Maeyama et al., 2008). However, these investigators did not specify the gait cycle phase in which the accelerations were observed. Several studies have reported inferior femoral translation during swing phase in subjects with a variety of hip arthroplasty devices (Dennis et al., 2001; Komistek et al., 2002; Lombardi et al., 2000), but we are unaware of similar reports on dysplastic hips. We observed inferior femoral translations during early swing phase with proximal return prior to foot-strike. Total translation was greater in DDH hips (1.9 ± 0.4 mm) than in contralateral healthy hips (1.0 ± 0.2 mm). We did not observe superolateral translation during stance in either group of hips.

Several radiographic indices are used to evaluate DDH and to quantify acetabular coverage after osteotomy. The CE angle is an objective measure used to diagnose DDH (Iglic et al., 2002; Mavcic and Pompe, 2002; Mavcic et al., 2000). AHI quantifies relative acetabular support, and ARO and Sharp angles express acetabular inclination. These radiographic indices are not independent, but supplement DDH grading (Hasegawa et al., 1992; Ito et al., 2004; Nicholls et al., 2011). Nakamura et al. correlated these radiographic indicators and primary OA and found the CE angle and ARO were strongly correlated to primary OA with superolateral migration (Nakamura et al., 1989). Our data show strong correlations between hip translation and CE angle, AHI and ARO in DDH hips without OA. Although we did not observe superolateral translation during stance phase in our study, greater swing-phase femoral translations in more severely malformed hips would appear predictive of superolateral translation and progression to OA, consistent with Nakamura et al.'s findings.

The fact that femoral head translation occurs in DDH hips may be important to understand the etiology of OA. Previous studies with pressure (Hodge et al., 1986) and force (Bergmann et al., 2001) instrumented hip arthroplasties have shown hip joint forces and acetabular pressures increase significantly in advance of foot-strike during gait. Thus, observed superior femoral translations during late swing-phase likely accompany significant joint forces and hip contact pressures. These motion and loading conditions likely are not typical of homeostatic joint mechanics and may precipitate a mechanobiologic cascade leading to OA (Andriacchi et al., 2004; Carter et al., 2004). In the knee, Andriacchi et al. have shown changes in knee kinematics after anterior cruciate ligament damage predict the location of arthritic degeneration (Andriacchi et al., 2004). In a conforming joint like the hip, changes in acetabular pressures will be much more sensitive to altered joint kinematics, so it is reasonable to predict arthritic degeneration based upon abnormal femoral translations. In addition to altered articular contact mechanics, femoral translations likely will affect stresses in the labrum and, consequently, joint sealing and lubrication. Several studies have highlighted the role of the acetabular labrum in stabilizing the femoral head and preventing joint fluid from flowing into the extraarticular space (Crawford et al., 2007; Ferguson et al., 2003). Henak et al. used finite element analysis to find larger stresses in the labrum with superolateral translations in a DDH model (Henak et al., 2011). Studies to correlate subject-specific hip kinematics and patterns of osteoarthritic degeneration will be required to support these etiologic hypotheses.

Prearthritis or early degeneration are typical triggers for performing acetabular osteotomy in patients with DDH (Yasunaga et al., 2001). Labral tears typically are not considered as a surgical indication in our protocol, and we did not evaluate the existence of labral tears in the DDH hips. Our results, therefore, must consider both labral tears and capsular laxity as contributing to greater hip translations in the preoperative DDH patients. Measures to evaluate labral tears and capsular laxity separately are needed to determine responsibility for greater translations in DDH hips.

This study has several limitations. First, our selection criteria were quite restrictive and excluded hips with more severe subluxation or diagnosed OA. We would expect, therefore, femoral translations in the DDH hips studied to be closer to normal and less than might be observed in cases with more severe subluxation or OA. Second, we observed a stepping-in-place motion in order to permit detailed observation using single-plane fluoroscopy. Stepping-in-place lacks

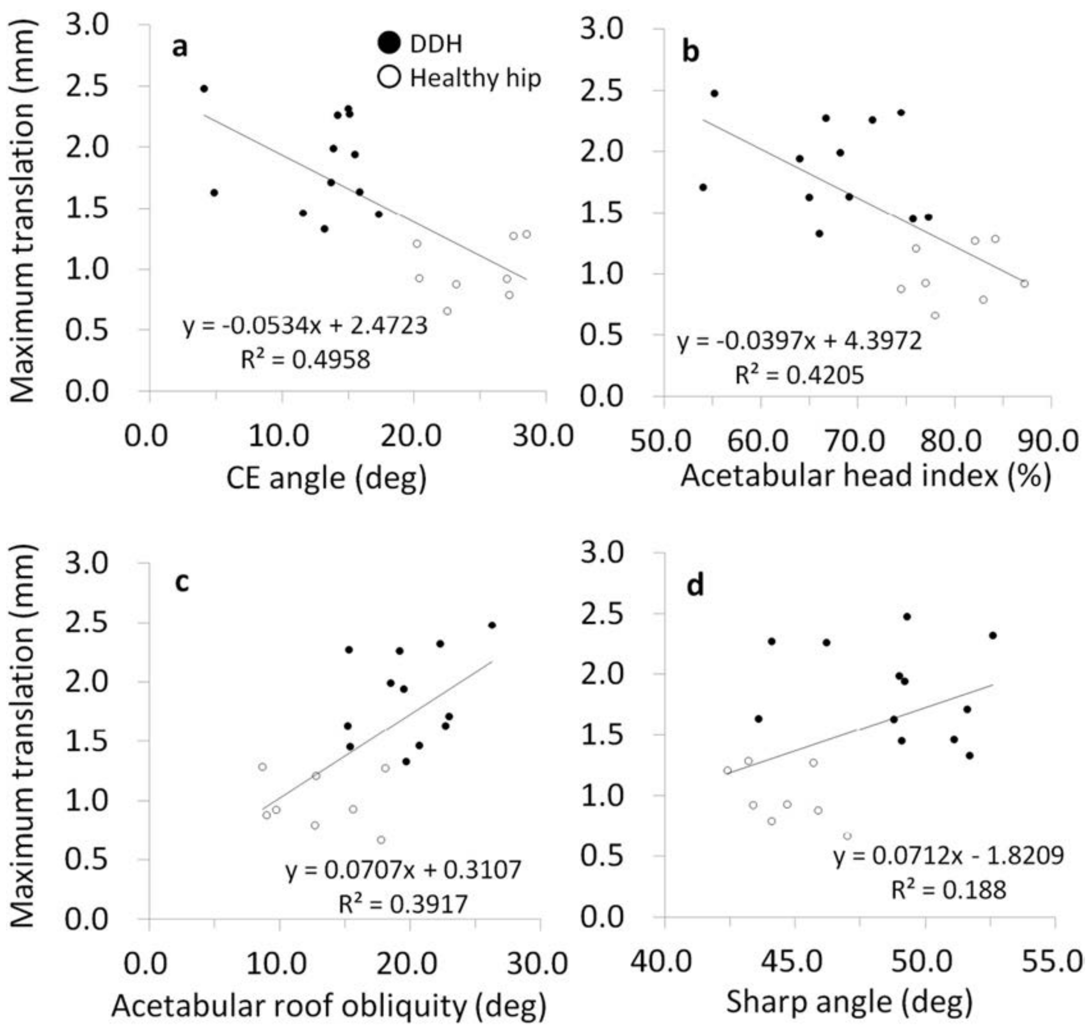


**Fig. 4.** Maximum femoral head translation as a function of radiographic measures of hip dysplasia: (a) CE angle, (b) Acetabular Head Index, (c) Acetabular Roof Obliquity, (d) Sharp angle.

the dynamics and hip range of motion of overground gait, so we can reasonably expect greater hip translations during true walking. This may be especially important in terminal stance with the hip extended, where the pubofemoral ligament and iliofemoral ligament inhibit external rotation (Martin et al., 2008). If DDH hips develop ligament laxity or dysfunction, greater anterior femoral translation may be observed. Third, the sampling rate for fluoroscopy was 15 Hz, which is slower than typically used for gait studies. A faster sampling rate allows more detailed kinematics to be captured, including peak translations, which might have revealed a larger range of total hip translations than was observed in this study. Fourth, MR images were not available for all subjects, so labral integrity could not be related to joint kinematics, especially in the superior direction. Fifth, the accuracy of single-plane registration is lower for AP (out of plane) translations. We did not observe any differences in AP femoral translation between the DDH and contralateral healthy hips, possibly because of inadequate measurement sensitivity. Finally, the measurement validation study was performed using static imaging conditions which were different from our clinical study. However, 8 ms exposures with small pelvis translation velocities likely minimize motion blur-related measurement artifacts (Ellingson et al., 2016).

In conclusion, DDH hips exhibit greater swing-phase femoral translations during walking-in-place than contralateral healthy hips. These excess motions may be predictive of future degenerative changes, due to abnormal contact pressures and locations, overload of the labrum, lost joint sealing and lubrication, and related mechanical changes. Significant correlations between hip geometry (CE angle, AHI, ARO) and dynamic femoral head translation suggest a causal relationship for hip OA that will require longitudinal studies to confirm. Measurements of femoral translation in DDH hips provide information that may lead to a better understanding of the pathogenesis of hip OA and related treatment strategies.

## Conflict of interest

There is no conflict of interest with regard to this article.

## Acknowledgments

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## Appendix A. Supplementary data

Supplementary data to this article can be found online at http://dx.doi.org/10.1016/j.clinbiomech.2017.05.003.

## References

Akiyama, K., Sakai, T., Koyanagi, J., Yoshikawa, H., Sugamoto, K., 2011. Evaluation of Translation in the Normal and Dysplastic Hip Using Three-dimensional Magnetic

Resonance Imaging and Voxel-based Registration. (Osteoarthritis and Cartilage).
Andriacchi, T.P., Mundermann, A., Smith, R.L., Alexander, E.J., Dyrby, C.O., Koo, S., 2004. A framework for the in vivo pathomechanics of osteoarthritis at the knee. Ann. Biomed. Eng. 32, 447–457.
Banks, S.A., Hodge, W.A., 1996. Accurate measurement of three-dimensional knee replacement kinematics using single-plane fluoroscopy. IEEE Trans. Biomed. Eng. 43, 638–649.
Bergmann, G., Deuretzbacher, G., Heller, M., Graichen, F., Rohlmann, A., Strauss, J., Duga, G.N., 2001. Hip contact forces and gait patterns from routine activities. J. Biomech. 34, 859–871.
Carter, D.R., Beaupre, G.S., Wong, M., Smith, R.L., Andriacchi, T.P., Schurman, D.J., 2004. The mechanobiology of articular cartilage development and degeneration. Clin. Orthop. Relat. Res. S69–S77.
Chung, C.Y., Park, M.S., Lee, K.M., Lee, S.H., Kim, T.K., Kim, K.W., Park, J.H., Lee, J.J., 2010. Hip osteoarthritis and risk factors in elderly Korean population. Osteoarthritis and cartilage/OARS, Osteoarthritis Research Society 18, 312–316.
Crawford, M.J., Dy, C.J., Alexander, J.W., Thompson, M., Schroder, S.J., Vega, C.E., Patel, R.V., Miller, A.R., Mccarthy, J.C., Lowe, W.R., Noble, P.C., 2007. The 2007 Frank Stinchfield award. The biomechanics of the hip labrum and the stability of the hip. Clin. Orthop. Relat. Res. 465, 16–22.
D, T., R, G., 1987. General radiography of the hip joint. In: D, T., H, L. (Eds.), Congenital Dysplasia and Dislocation of the Hip in Children and Adults. Springer-Verlag, Berlin Heidelberg.
Dennis, D.A., Komistek, R.D., Northcut, E.J., Ochoa, J.A., Ritchie, A., 2001. In vivo determination of hip joint separation and the forces generated due to impact loading conditions. J. Biomech. 34, 623–629.
Ellingson, A.M., Mozingo, J.D., Magnuson, D.J., Pagnano, M.W., Zhao, K.D., 2016. Characterizing fluoroscopy based kinematic accuracy as a function of pulse width and velocity. J. Biomech. 49, 3741–3745.
Endo, H., Mitani, S., Senda, M., Kawai, A., McCown, C., Umeda, M., Miyakawa, T., Inoue, H., 2003. Three-dimensional gait analysis of adults with hip dysplasia after rotational acetabular osteotomy. Journal of Orthopaedic Science: Official Journal of the Japanese Orthopaedic Association 8, 762–771.
Ferguson, S.J., Bryant, J.T., Ganz, R., Ito, K., 2000. The influence of the acetabular labrum on hip joint cartilage consolidation: a poroelastic finite element model. J. Biomech. 33, 953–960.
Ferguson, S.J., Bryant, J.T., Ganz, R., Ito, K., 2003. An in vitro investigation of the acetabular labral seal in hip joint mechanics. J. Biomech. 36, 171–178.
Fessler, H., 1957. Load distribution in a model of a hip joint. J. Bone Joint Surg. Br. Vol. 39-B, 145–153.
Hasegawa, Y., Iwata, H., Mizuno, M., Genda, E., Sato, S., Miura, T., 1992. The natural course of osteoarthritis of the hip due to subluxation or acetabular dysplasia. Arch. Orthop. Trauma Surg. 111, 187–191.
Henak, C.R., Ellis, B.J., Harris, M.D., Anderson, A.E., Peters, C.L., Weiss, J.A., 2011. Role of the acetabular labrum in load support across the hip joint. J. Biomech. 44, 2201–2206.
Heyman, C.H., Herndon, C.H., 1950. Legg-Perthes disease; a method for the measurement of the roentgenographic result. J. Bone Joint Surg. Am. 32, 767–778.
Hodge, W.A., Fijan, R.S., Carlson, K.L., Burgess, R.G., Harris, W.H., Mann, R.W., 1986. Contact pressures in the human hip joint measured in vivo. Proc. Natl. Acad. Sci. U. S. A. 83, 2879–2883.
Iglic, A., Kralj-Iglic, V., Daniel, M., Macek-Lebar, A., 2002. Computer determination of contact stress distribution and size of weight bearing area in the human hip joint. Computer Methods in Biomechanics and Biomedical Engineering 5, 185–192.
Ito, H., Matsuno, T., Minami, A., 2004. Chiari pelvic osteotomy for advanced osteoarthritis in patients with hip dysplasia. J. Bone Joint Surg. Am. 86-A, 1439–1445.
Jacobsen, S., Sonne-Holm, S., 2005. Hip dysplasia: a significant risk factor for the development of hip osteoarthritis. A cross-sectional survey. Rheumatology (Oxford) 44, 211–218.
Johnsen, K., Goll, R., Reikeras, O., 2009. Acetabular dysplasia as an aetiological factor in development of hip osteoarthritis. Int. Orthop. 33, 653–657.
Klaue, K., Durnin, C., Ganz, R., 1991. The acetabular rim syndrome. A clinical presentation of dysplasia of the hip. J. Bone Joint Surg. Br. Vol. 73, 423.
Komistek, R.D., Dennis, D.A., Ochoa, J.A., Haas, B.D., Hammill, C., 2002. In vivo comparison of hip separation after metal-on-metal or metal-on-polyethylene total hip arthroplasty. J. Bone Joint Surg. 84A, 1836–1841.
Lane, N.E., Lin, P., Christiansen, L., Gore, L.R., Williams, E.N., Hochberg, M.C., Nevitt, M.C., 2000. Association of mild acetabular dysplasia with an increased risk of incident hip osteoarthritis in elderly white women: the study of osteoporotic fractures. Arthritis Rheum. 43, 400–404.
Lau, E.M., Lin, F., Lam, D., Silman, A., Croft, P., 1995. Hip osteoarthritis and dysplasia in Chinese men. Ann. Rheum. Dis. 54, 965–969.
Lombardi Jr., A.V., Mallory, T.H., Dennis, D.A., Komistek, R.D., Fada, R.A., Northcut, E.J., 2000. An in vivo determination of total hip arthroplasty pistoning during activity. J. Arthroplast. 15, 702–709.
Maeyama, A., Naito, M., Moriyama, S., Yoshimura, I., 2008. Evaluation of dynamic instability of the dysplastic hip with use of triaxial accelerometry. J. Bone Joint Surg. Am. 90, 85–92.
Martin, H.D., Savage, A., Braly, B.A., Palmer, I.J., Beall, D.P., Kelly, B., 2008. The function of the hip capsular ligaments: a quantitative report. Arthroscopy: The Journal of Arthroscopic & Related Surgery: Official Publication of the Arthroscopy Association of North America and the International Arthroscopy Association 24, 188–195.
Massie, W.K., Howorth, M.B., 1950. Congenital dislocation of the hip. Part I. Method of grading results. J. Bone Joint Surg. Am. 32-A, 519–531.
Mavcic, B., Antolic, V., Brand, R., Iglic, A., Ipavec, M., Kralj-Iglic, V., Pedersen, D.R., 2000. Weight bearing area during gait in normal and dysplastic hips. Pflugers Arch. - Eur. J. Physiol. 439, R213–R214.
Mavcic, B., Iglic, A., Kralj-Iglic, V., Brand, R.A., Vengust, R., 2008. Cumulative hip contact stress predicts osteoarthritis in DDH. Clin. Orthop. Relat. Res. 466, 884–891.
Mavcic, B., Pompe, B., 2002. Mathematical estimation of stress distribution in normal and displastic human hips. J. Orthop. Res. 20, 1025–1030.
McCarthy, J., Noble, P., Aluisio, F.V., Schuck, M., Wright, J., Lee, J.A., 2003. Anatomy, pathologic features, and treatment of acetabular labral tears. Clin. Orthop. Relat. Res. 38–47.
Murphy, S.B., Ganz, R., Muller, M.E., 1995. The prognosis in untreated dysplasia of the hip. A study of radiographic factors that predict the outcome. J. Bone Joint Surg. Am. 77, 985–989.
Nakamura, S., Ninomiya, S., Nakamura, T., 1989. Primary osteoarthritis of the hip joint in Japan. Clin. Orthop. Relat. Res. 190–196.
Nicholls, A.S., Kiran, A., Pollard, T.C.B., Hart, D.J., Arden, C.P.A., Gill, T., Murray, D.W., Carr, A.J., Arden, N.K., 2011. The association between hip morphology parameters and nineteen-year risk of end-stage osteoarthritis of the hip: a nested case-control study. Arthritis Rheum. 63, 3392–3400.
Philippon, M., Zehms, C., Briggs, K., Manchester, D., Kuppersmith, D., 2007. Hip instability in the athlete. Operative Techniques in Sports Medicine 15, 189–194.
Pompe, B., Antolic, V., Mavcic, B., Iglic, A., Kralj-Iglic, V., 2007. Hip joint contact stress as an additional parameter for determining hip dysplasia in adults: comparison with Severin's classification. Medical Science Monitor: International Medical Journal of Experimental and Clinical Research 13, CR215–CR219.
Pratt, V., 1987. Direct Least-squares Fitting of Algebraic Surfaces. Proceedings of the 14th Annual Conference on Computer Graphics and Interactive Techniques ACM.
Reijman, M., Hazes, J., Pols, H., Koes, B., Bierma-Zeinstra, S., 2005. Acetabular dysplasia predicts incident osteoarthritis of the hip: the Rotterdam study. Arthritis Rheum. 52, 787–793.
Sharp, I.K., 1961. Acetabular dysplasia: the acetabular angle. J. Bone Joint Surg. Br. Vol. 43, 268.
Shu, B., Safran, M.R., 2011. Hip instability: anatomic and clinical considerations of traumatic and atraumatic instability. Clin. Sports Med. 30, 349–367.
Wiberg, G., 1939. Studies on dysplastic acetabula and congenital subluxation of the hip joint: with special reference to the complication of osteoarthritis. Acta Orthop. Scand. 83 (Suppl. 58), 1–135.
Wu, G., 2002. ISB recommendation on definition of joint coordinate system of various joints for the reporting of human joint motion. J. Biomech. 35, 543–548.
Yasunaga, Y., Ikuta, Y., Kanazawa, T., Takahashi, K., Hisatome, T., 2001. The state of the articular cartilage at the time of surgery as an indication for rotational acetabular osteotomy. J. Bone Joint Surg. Br. Vol. 83, 1001–1004.
Yoshida, H., Faust, A., Wilckens, J., Kitagawa, M., Fetto, J., Chao, E.Y.S., 2006. Three-dimensional dynamic hip contact area and pressure distribution during activities of daily living. J. Biomech. 39, 1996–2004.
Yushkevich, P.A., Piven, J., Hazlett, H.C., Smith, R.G., Ho, S., Gee, J.C., Gerig, G., 2006. User-guided 3D active contour segmentation of anatomical structures: significantly improved efficiency and reliability. NeuroImage 31, 1116–1128.

**Supplementary Material**

*Measurement Validation Study,*

We performed an in vitro study to establish the accuracy of single plane 3D-2D registration with AP projections of natural hips. We compared the single plane model-image registration method, as used in the in vivo study, with x-ray stereophotogrammetery using implanted metal beads as the reference measure. Several studies have reported bead-based stereophotogrammetric methods provide superior measurement accuracy compared to single plane model-image registration methods, especially for translations perpendicular to the single-plane image (You et al., 2001, Li et al., 2004, Li et al., 2008).

We used an adult male cadaver with an 85cm waist circumference. Four and five 2mm-diameter steel beads were embedded in the pelvic and femoral cortices, respectively. CT scans were performed to develop 3D models of the bones and bead clusters (SOMATOM Sensation 16, Siemens AG; 140kVp, 0.5 mm slice thickness, 512 × 512 acquisition matrix, 0.94mm pixel spacing).

The specimen was fixed to a turntable with fixed 45° rotation stops, permitting repeatable positioning of the specimen in anteroposterior (AP) and 45° oblique views using a single fluoroscopy system (AXIOM-Artis, Siemens AG, Germany, Fig. A1). The specimen was fixed in seven positions, including joint distraction, covering the range of motion for the stepping activity, and AP and oblique views were captured. Two experienced software users, including the individual who analyzed all of the in vivo images, analyzed the biplane and single-plane image sets three times at least three days apart; the biplane image pairs using composite 3D models of the bones and metal bead clusters, and the single plane AP views using 3D models of the femur and pelvis. Root mean square (RMS) differences between joint translations and rotations measured with single- and biplane techniques were computed and represent lower bounds for measurement errors of the single-plane technique.

RMS errors for coronal plane translations were 0.2 mm, anterior/posterior translations 0.5 mm, and 1.6° for joint rotations.

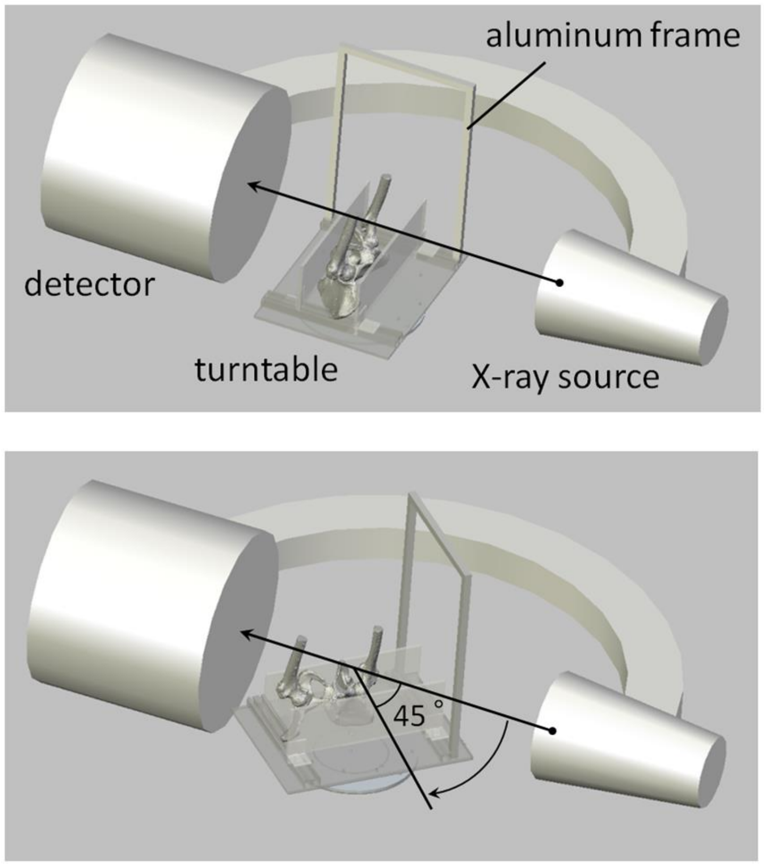


Fig.A1 system setup for bi-plane fluoroscopic image

LI, G., VAN DE VELDE, S. K. & BINGHAM, J. T. 2008. Validation of a non-invasive fluoroscopic imaging technique for the measurement of dynamic knee joint motion. *Journal of Biomechanics,* 41**,** 1616-1622.

LI, G., WUERZ, T. H. & DEFRATE, L. E. 2004. Feasibility of using orthogonal fluoroscopic images to measure in vivo joint kinematics. *Journal of biomechanical engineering,* 126**,** 314-8.

YOU, B. M., SIY, P., ANDERST, W. & TASHMAN, S. 2001. In vivo measurement of 3-D skeletal kinematics from sequences of biplane radiographs: application to knee kinematics. *IEEE Transactions on Medical Imaging,* 20**,** 514-25.