***Manuscript (including title page)**

Title:

Estimation of tibiofemoral static zero position during dynamic drop landing

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Abstract (248/250 words):

The aim of this study was to assess the *in vivo* knee secondary motions intrinsic to flexion in isolation from actual displacements during a landing activity. For this purpose a “static zero position”, which denotes the normal tibiofemoral position to the static flexion angle, was introduced to describe the intrinsic secondary motion. The three-dimensional motion data of the healthy knee were collected for 13 male and 13 female young adults by using an auto motion analysis system and point cluster technique. First, the relationship between flexion and secondary motion in the static state was determined during a single-leg quasistatic squat. The static zero position during a single-leg drop landing was then calculated by substituting the flexion angle into the flexion-secondary relational expression obtained. The results showed that after the foot-ground contact, the estimated static zero positions shifted monotonically in valgus, internal rotation, and anterior translation in the case of both the male and female groups. For the time-course change, noticeable differences between the actual displacement and estimated static zero position were found from the foot-ground contact up to 25 ms after the contact for the valgus/varus and external/internal rotation, and between 20 and 35 ms after the contact for the anterior/posterior translation. In summary, the static zero position demonstrated relatively modest but not negligible shift in comparison with the actual displacement. The intrinsic tibiofemoral motion, or baseline shift, would be worth taking into account when examining the fundamental function and injury mechanics of knee during an impulsive activity.

Main Text:

Introduction

In the tibiofemoral joint, an involuntary axial rotation occurs as the knee flexes or extends. This is known as a "screw-home movement" (SHM) and was first demonstrated using a cadaver specimen [1-4]. The SHM describes the coupled property in which the tibia automatically rotates externally relative to the femur during passive extension of the knee, and the reverse occurs during knee flexion. Regardless of whether this property is coupled, the rotational motions, *i.e.*, valgus/varus and external/internal rotation, and the translational motions, *i.e.*, lateral/medial translation, anterior/posterior translation, and superior/inferior translation, have been investigated as the knee motions that are intrinsic to flexion, recently termed secondary motions [5, 6]. As reported previously, application of minimal resistance to the motion of a cadaver knee model by using a rig showed that all of the secondary motions were coupled to passive flexion [7]. The geometric structure of the femoral epicondyle and mechanical stress of ligaments have been considered anatomical determinants of the coupled motion [8-10]. In contrast, several studies have described the secondary motion as it lies within a relatively wide range called the envelope of passive flexion [11]. Although the notion of coupled motion and that of the envelope conceptually oppose one another, both of these ideas share the view that the motion path is highly sensitive to the external load [7]. Numerous studies have assessed secondary motion, because it is closely linked to the fundamental knee function in terms of mobility, stiffness, and stability [12]. However, at present, *in vivo* behaviors of these secondary motions remain poorly understood owing to not only the technical difficulty of measurement but also the multifold effects of possible neuromuscular control depending on the type of motor activity.

The behavior of secondary motion can be analyzed on the basis of three-dimensional (3D) measurement. *In vivo* 6-degree-of-freedom (6DOF) motion of the tibiofemoral joint, sometimes with the patella, has been evaluated to study the function of the living knee by using various apparatuses and devices: computerized electrogoniometers [13], intracortical traction marker pins fixed on the bone [14], clusters of skin marker sets attached to a body segment [5], and image matching with 3D bone models and projected images

[15-18]. These studies attempted to reveal the typical kinematic patterns of healthy or injured knees during gross motor activities such as gait, lunge, and squat. Nevertheless, little is known about the nature of secondary motion during high-impact activities such as drop landing. The investigation of these strenuous motions will lead to better understandings of the knee function to control an impulsive activity.

The analysis of 3D motion has also been adopted in studies of injury mechanisms, *e.g.,* identification of the kinematic risk factors for noncontact anterior cruciate ligament (ACL) injury. Several previous studies have examined the rotational and translational displacements of the tibiofemoral joint during risk motions, such as drop landing or cutting, and have shown that the valgus and external/internal rotation of the tibia at slight knee flexion were kinematic risk factors for ACL injury [19-21]. The kinematic data obtained, however, could not be separated into intrinsic motion under static conditions, such as quiet weight-bearing, and extrinsic motion due to dynamic effects, such as landing impact or rapid muscle contraction. It has been shown that the knee frontal plane angle during several landing tasks, *i.e.*, stepping down, single-leg landing and drop vertical jumping, were substantially different from that during static standing [22]. The findings of this previous study also suggested that neuromuscular mechanisms contribute to the consistent control across these dynamic activities. Examination of the discrepancies between the static and dynamic effects by using a 3D analysis might provide further insight into the injury mechanism. For this purpose, an effective approach would be to separate the intrinsic secondary motion in a static state from the net displacements during an impulsive activity.

The objective of this study was to assess the static secondary motions of the healthy knee intrinsic to flexion in isolation from the total displacements during a dynamic landing activity. The authors introduced the notion of “static zero position” that represents the normal position of the tibiofemoral joint in a state of static equilibrium. If the static zero position shifts with the knee flexion angle as typically demonstrated by the SHM, it is regarded as a baseline shift in the tibia-femur relative motion. A comparison between the actual displacement and estimated static zero position during a gross motor activity could provide knowledge about not only fundamental knee motor function but also knee injury mechanism.

Materials and Methods

Thirteen healthy male young adults, mean age 23.2 years (range 20-27), and 13 healthy female counterparts, mean age 22.8 years (range 18-27), participated in this study. Informed consent was obtained from the subjects, and the study was approved by an ethics committee of the National Rehabilitation Center for Persons with Disabilities. The right knee was studied if a history of leg injury was absent; otherwise, the left knee was chosen for testing. Twelve right and 1 left knee for the male subjects, and 11 right and 2 left knees for the female subjects were investigated. Because complete data were not obtained for 2 female subjects after the estimation of the static zero position was started, these 2 subjects were excluded from all of the subsequent analyses. Numerous studies have shown significant sex-based influences on knee kinematics, thus the assessments of the static zero position were basically performed separately for the male and female groups.

In this study, a point cluster technique (PCT) that outputs the *in vivo* 6DOF rotational and translational displacements of the tibiofemoral joint was used to collect the 3D kinematic data [23, 24]. The PCT has been used to estimate ACL strain [25], identify the kinematic risk factors for ACL injury, [19, 20] compare the rotational stability between single- and double-bundle ACL reconstructions [26], and calculate the knee joint moment [27]. Among the 6DOF data, the present study examined the valgus/varus, external/internal rotation, and anterior/posterior translation that were discussed in a relevant previous study [5].

The subjects were asked to perform 3 motor tasks barefoot: quiet standing, single-leg quasistatic squat, and single-leg drop landing. First, the subjects were instructed to maintain a quiet upright standing stance with their feet shoulder-width apart for 1 s, as the rest position trial of the PCT procedure. Second, single-leg quasistatic squat was performed 3 times with their foot in the neutral position; the test knee was slowly flexed over 5 s from full extension to deep flexion (Fig. 1), simulating the motion of the subsequent drop landing task. A fluoroscopic study has shown that a slow squatting motion is likely to produce equivalent knee kinematics compared to a series of completely static squats [18]. Finally, the subjects executed the single-leg drop landing 3 times in neutral foot position from a 30-cm-high platform to a 30-cm forward landing point with their hands on their hips.

The 3D coordinate data of the 3 motor tasks were collected using a 6-camera retroreflective auto motion capture system (Hawk Digital RealTime System, Motion Analysis, Inc., Santa Rosa, CA, USA) at a sampling rate of 200 Hz. The vertical ground reaction force was also monitored using a force platform (9287A, Kistler Japan Co., Ltd., Tokyo, Japan) at a sampling rate of 1 kHz during quasistatic squat and drop landing. The force platform was synchronized with the auto motion capture system.

The arrangement of skin markers attached to the test leg followed the empirical pattern of our PCT procedure (Fig. 1). In total, 24 markers included 10 and 6 on the thigh and shank segments, respectively, as a cluster marker set, and markers on the great trochanter (1), the lateral and medial epicondyles of the femur (2), the lateral and medial edges of the tibia plateau (2), the lateral (fibula) and medial malleoli (2), and the fifth metatarsophalangeal joint (1) as bone landmarks.

For the rest position trial, the PCT algorithm carried out a principal axis transformation on the assumption that each cluster marker had a unit mass, and the cluster coordinate system obtained was transformed to a bone coordinate system fixed on the bone landmarks [23]. A bone coordinate system was also derived for each frame of quasistatic squat and drop landing with optimized masses. The displacements of the tibiofemoral joint were defined with the vector projection onto a joint coordinate system [28]. This approach has the advantage of reporting the rotational and translational displacements in clinical terminology [29]. First, the longitudinal axis of the shank was chosen as the superior(+)/inferior(-) axis ($\boldsymbol{e}_3$ in Grood & Suntay). The anterior(+)/posterior(-) axis ($\boldsymbol{e}_2$) was then obtained by computing the cross product of the superior/inferior axis and a tentative transverse axis from the medial to lateral epicondyle of the femur. Finally, the lateral(+)/medial(-) axis ($\boldsymbol{e}_1$) was determined as the cross product of the anterior/posterior axis and superior/inferior axis. All the displacements were described as the tibial displacement relative to the femur.

The estimation accuracy of our PCT algorithm was assessed using a dummy leg model for which the thigh and shank segments were shaped out of Styrofoam, and an aluminum precision stage was installed as the knee joint. The installed precision stages were the gonio stage, rotation stage, and XY stage, which is equivalent to valgus/varus, external/internal rotation, and anterior/posterior translation, respectively. The minimum scales of these stages were 0.1 deg, 0.08 deg, and 0.01 mm, respectively. The root mean square errors between the stage scales and values calculated by PCT were 0.079 deg for valgus of 0 to 12 deg, 0.24 deg for rotation of 0 to 12 deg, and 0.38 mm for anterior translation of 0 to 12 mm [24].

For each secondary motion, the average curve of the 3 quasistatic squatting trials within the individual subject was plotted versus the flexion angle. This averaged curve was considered the static flexion-secondary relation ($f_{\text{flexion-secondary}}$) that converted the flexion angle (θ_{static}) into one of the secondary motion (α_{static}), assuming a static equilibrium state under the weight-bearing condition:

$$\alpha_{\text{static}} = f_{\text{flexion-secondary}}(\theta_{\text{static}})$$

where the conversion relation was determined for every 1 degree of the flexion angle. One of the static zero position during the drop landing ($\alpha_{\text{static zero position}}$) was estimated by substituting the flexion angle ($\theta_{\text{drop landing}}$) during the task into the static flexion-secondary relations obtained above:

$$\alpha_{\text{static zero position}} = f_{\text{flexion-secondary}}(\theta_{\text{drop landing}})$$

The flexion angle ($\theta_{\text{drop landing}}$) was substituted for every 1 frame (5 ms) of the motion capture data from the foot-ground contact (T = 0) to 80 ms after the contact.

Statistical tests were performed using statistical software (SPSS 17.0, SPSS Japan Inc., Tokyo, Japan). Independent samples *t*-test was conducted to test the difference between the male and female groups for the vertical ground reaction force and the actual displacement of the secondary motion during the drop landing. Cohen's *d* was also used to estimate effect size of the difference. The significance level was set at alpha = 0.05. The difference between the actual displacement and estimated static zero position within each sex group was tested using paired samples *t*-test for every 1 frame in a time-course curve. For these frame-by-frame multiple comparisons (total, 17 frames), a Bonferroni correction was used to avoid inflation of the alpha level (0.05/17 = 0.0029)

Results

Figure 2 shows typical data of the knee flexion angle and vertical ground reaction force during quasistatic squat. The vertical ground reaction force was almost constant, or had no typical trend, over the task duration. These were observed for all subjects and indicated that all of them successfully performed the quasistatic squat as required. Therefore, quasistatic squat was regarded as the static state sequence at the corresponding flexion angle; *i.e.,* no dynamic effect on the secondary motion was observed.

Figure 3 shows the typical secondary motion curve plotted versus the flexion angle during quasistatic squat. These secondary motions were similar in time-course tendency among the 3 trials for an individual subject, while remarkable variation was observed among the subjects by visual inspection. Among the male subjects (n = 13), 9 showed valgus, 12 showed internal rotation, and 11showed anterior translation of the tibia as the knee flexed; the remaining subjects were opposite. Similarly, among the female subjects (n = 11; 2 others had been excluded), 10 showed valgus, 8 showed internal rotation, and 11 showed anterior translation of the tibia. It should be noted that the position was defined relative to the neutral tibia-femur position derived from the rest position trial (quiet standing).

During drop landing, the vertical ground reaction force showed a peak of mean 362%BW (*SD* 61) at T = 38.8 ms (*SD* 11.6) and 327%BW (*SD* 48) at T = 47.3 ms (*SD* 8.8) for the male and females groups, respectively. No significant differences were observed between the sex groups with regard to normalized peak value ($t(22) = 1.55$, $P = 0.135$, $d = 0.66$), and peak time ($t(22) = 1.98$, $P = 0.060$, $d = 0.85$).

The solid line in Figure 4 shows the actual displacement of secondary motion from the foot-ground contact to 80 ms after the contact during drop landing. The position of the secondary motion at the foot-ground contact is shown in Table 1. None of these landing positions showed significant differences between the male and female groups. In addition, the value and time of typical peaks are shown in Table 2. The valgus peak time was significantly earlier in the male group than in the female group, but the other peak values and peak times did not show any significant differences between the sex groups.

In addition to the actual displacement, the estimated static zero position during the drop landing is shown in Figure 4. The estimated static zero position at the foot-ground contact was valgus, internally rotated, and anterior translated position in the mean value for both the male and female groups. Thereafter, the estimated static zero position shifted monotonically in valgus, internal rotation, and anterior translation for both the sex groups. The P value of Student's t-tests between the actual displacement and estimated static zero position is shown for every 1 frame at the bottom of each graph. Low P values (< 0.0029 with Bonferroni correction) were mainly found in $T \leq 25$ ms for the valgus/varus and external/internal rotation, and between T = 20 and 35 ms for the anterior/posterior translation (filled circles in Fig. 4).

Discussion

This study examined the actual rotational and translational displacements of the tibiofemoral joint during a drop landing task compared with the shift of the static zero position in which the knee was assumed to be in static equilibrium. The static zero position was estimated by an extrapolative substitution of the flexion angle into flexion-secondary relations derived from quasistatic weight-bearing squat. This procedure allowed direct comparisons between the static (intrinsic) and dynamic (extrinsic) effects on the joint motion (Fig. 4). The results of this study have 2 potential contributions in clinical practice. First, they deepen the knowledge of the function of knee secondary motion during an impulsive activity aside from the SHM. Second, they offer suggestions about the kinematic mechanism of knee injury.

Previous studies showed that secondary motion was coupled to knee flexion when static weight-bearing tasks were performed by healthy subjects [15, 17] and ACL-deficient subjects [16], although the kinematic pattern significantly changed depending on whether the subjects were healthy. In general, quantitative comparisons with other reports might elicit misunderstanding about joint kinematics because of differences in motor tasks and definitions of coordinate systems [17]. However, the secondary motion plotted versus the flexion angle during quasistatic squat (Fig. 3) qualitatively agreed with the conventions of kinematic pattern, *i.e.*, valgus, internal rotation of the tibia (as the SHM describes), and anterior translation of the tibia. High repeatability in the time-course trend among 3 trials for each subject supported the coupled property of secondary motion. On the other hand, apparently large variability between subjects did not contradict the *in vitro* result that the shape of the coupled path varies between knees [7]. Furthermore, the fact that not all male subjects and not all female subjects showed the same displacement pattern for each secondary motion, except for the anterior translation in the female group, also supported the result of the *in vitro* study.

Meanwhile, a previous study has shown that during the stance phase of walking there are substantial offsets in the secondary external/internal rotation and anterior/posterior translation at identical flexion angles [5]. The findings of this previous study suggested that the secondary motions possibly occur within a range of motion envelope during a dynamic weight-bearing activity such as walking. This previous study also showed that the secondary motions during a non-weight-bearing active leg extension-and-flexion cycle task had no such offsets but followed the same pathways for both of the reciprocating phases. In the current study, discrepancies between the actual displacement and estimated static zero position were observed during the drop landing (Fig. 4). These discrepancies could be attributed to the broadened motion within an envelope due to the dynamic weight-acceptance control during landing activity. The envelope appeared depending on the external load, muscular activation, and type of gross activity as well as the knee flexion angle, and the boundaries of this envelope would be determined by passive bone and soft tissue structures [5].

Interestingly, even at the foot-ground contact, the actual displacement and estimated static zero position were different from each other in valgus/varus and external/internal rotation in the male group and in external/internal rotation in the female group (see P value at $T = 0$ in Fig. 4). This might have been caused by a preparatory muscle contraction before foot-ground contact. Many studies have shown that the intensive muscle contraction of the quadriceps occurred in the pre-contact phase [30, 31]. However, the anterior/posterior translation at the foot-ground contact, which could be directly related to the preparatory quadriceps contraction, showed no differences between the actual displacement and estimated static zero position for both the male and female groups. This suggests that other complicating factors, such as the presence of a ground reaction force or the hamstrings/quadriceps ratio, masked the effect of quadriceps contraction on anterior translation.

A kinematic mechanism of ACL injury has been estimated with 3D motion analysis *in vivo* [19, 20, 22]. For example, internal tibial rotation combined with valgus rotation of the knee has been proposed as the injury mechanism during a single limb drop landing. However, the measured displacement of secondary motion thus far has not separated the motion induced by a dynamic effect from that induced by a static effect. Because in some cases, the secondary motion was intrinsically coupled to flexion in static weight-bearing activity [17], not only the quality of overall displacement but also the quantity of net displacement should be evaluated to discuss the risky kinematics. One way to obtain the net displacement is to subtract the estimated static zero position from the actual displacement.

The result of actual valgus, internal rotation, and anterior translation displacement during drop landing supports the findings of several studies [19, 21]; in contrast, slight varus has been reported in a study using biplane fluoroscopy [32]. In all secondary motions, the actual displacement changed drastically, while the estimated static zero position changed monotonically and moderately after the foot-ground contact. An *in vivo* case study showed that the ACL strain peaked around the peak of ground reaction force [33]. In this study, the vertical ground reaction force of the male group peaked at mean T = 38.8 ms (*SD* 11.6) during drop landing. For the valgus/varus and external/internal rotation, wide gaps (low P value) between the actual displacement and estimated static zero position were observed at an earlier phase than the peak time of the vertical ground reaction force, typically at $T \leq 25$ ms (Fig. 4). The anterior translation, however, showed such a gap from T = 25 to 35 ms that it was relatively close to the reaction force peak. In particular, a steep peak of the anterior translation was also observed in this duration. These results suggest that the vertical ground reaction force can be more strongly related to the anterior translation of the tibia than the other secondary motions. Similar patterns were also observed in the female group, although the peak time of the vertical ground reaction force, mean T = 47.3 ms (*SD* 8.8), approached significance later than it did that in the male group (P = 0.060).

The procedure of estimating the static zero position can also be applied to activities under non-weight-bearing conditions. The coupled property was also reported for the non-weight-bearing task of leg extension on a chair [5]. A pilot study insisted that the timing of the peak ACL strain was estimated at 55 ms prior to the foot-ground contact during a jump landing [25]. Examination of the differences between static and dynamic conditions might also serve as the basis for future studies on non-weight-bearing activities.

The limitation of this study was the task-dependency of the measured activities. The quasistatic squat performed by the subjects was exclusively the static simulation of the single-leg drop landing used in this study; thus, the other dynamic task, such as cutting or double-leg landing, may need another quasistatic task to obtain intended flexion-secondary motion expressions. On the other hand, in terms of control of the ground reaction force, it must be taken into account that quasistatic squat was an activity bearing constant weight (body weight), while drop landing was performed under an inconsistent ground reaction force. For strict simulation of the drop landing situation, the ground reaction force should also be duplicated. In addition, although this study showed the anterior/posterior translation as Dyrby and Andriacchi (2004) did, this parameter might potentially have a large error [6]. Finally, the change in the shape of the thigh and shank muscles might contaminate the data of the secondary motions.

In conclusion, for the knee secondary motion, the actual displacement and estimated static zero position during drop landing differ from each other both in temporal pattern and in absolute amplitude. Although the shift of the static zero position was relatively modest as compared with the actual displacement, it appeared not to be negligible in magnitude. The discrepancies between static and dynamic effects can be attributed to the conceptually contradicting nature, *i.e.*, coupled and envelope.

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Table Titles:

Table 1. Position of secondary motion at foot-ground contact during drop landing

Table 2. Peak value and peak time of secondary motion during drop landing

Figure Legends:

Figure 1. Single-leg quasistatic squat.

Figure 2. Typical flexion angle and vertical ground reaction force during quasistatic squat (male subject #08).

Figure 3. Typical phase plot of secondary motion versus the flexion angle during quasistatic squat (a-c: male subject #08, d-f: female subject #09).

Figure 4. Mean time-course change (error bar: *SD*) of actual displacement (solid line) and estimated static zero position (dotted line) during drop landing for male (a-c) and female (d-f) subjects, and *P* value (right axis) at each frame. Filled circle (•) indicates significant difference with Bonferroni correction ($P < 0.05/17$(frames) $= 0.0029$).

Table(s)

Table 1. Position of secondary motion at foot-ground contact during drop landing

	Male		Female			
	M	*SD*	*M*	*SD*	*P*	*d*
Valgus(+)/varus(-) (deg)	-2.3	5.5	0.8	5.4	0.180	0.59
External(+)/internal(-) rotation (deg)	-0.6	7.0	-2.6	6.5	0.368	0.39
Anterior(+)/posterior(-) translation (cm)	0.51	0.61	0.24	0.75	0.349	0.41

Table(s)

Table 2. Peak value and peak time of secondary motion during drop landing

	Male		Female			
	M	*SD*	*M*	*SD*	*P*	*d*
Varus peak value (deg)	-1.6	1.3	-3.1	3.6	0.210	0.61
Varus peak time (ms)	23.1	4.3	27.3	9.3	0.192	0.62
Valgus peak value (deg)	7.5	3.2	11.1	5.5	0.060	0.85
Valgus peak time (ms)*	56.5	12.6	70.9	6.6	0.003	1.45
Internal rotation peak value (deg)	-15.9	5.3	-14.8	6.6	0.641	0.20
Internal rotation peak time (ms)	56.2	15.0	55.5	7.6	0.885	0.06
Anterior translation peak value (cm)	0.76	0.32	0.84	0.40	0.581	0.24
Anterior translation peak time (ms)	29.2	5.7	29.1	6.6	0.956	0.02

*: $P < 0.05$

Figure(s)

Figure 1

Figure(s)
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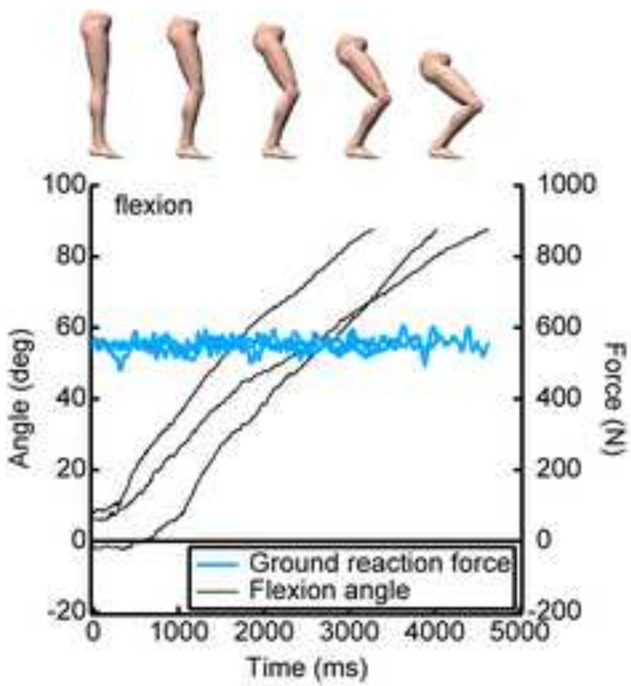

Figure 2

Figure(s)

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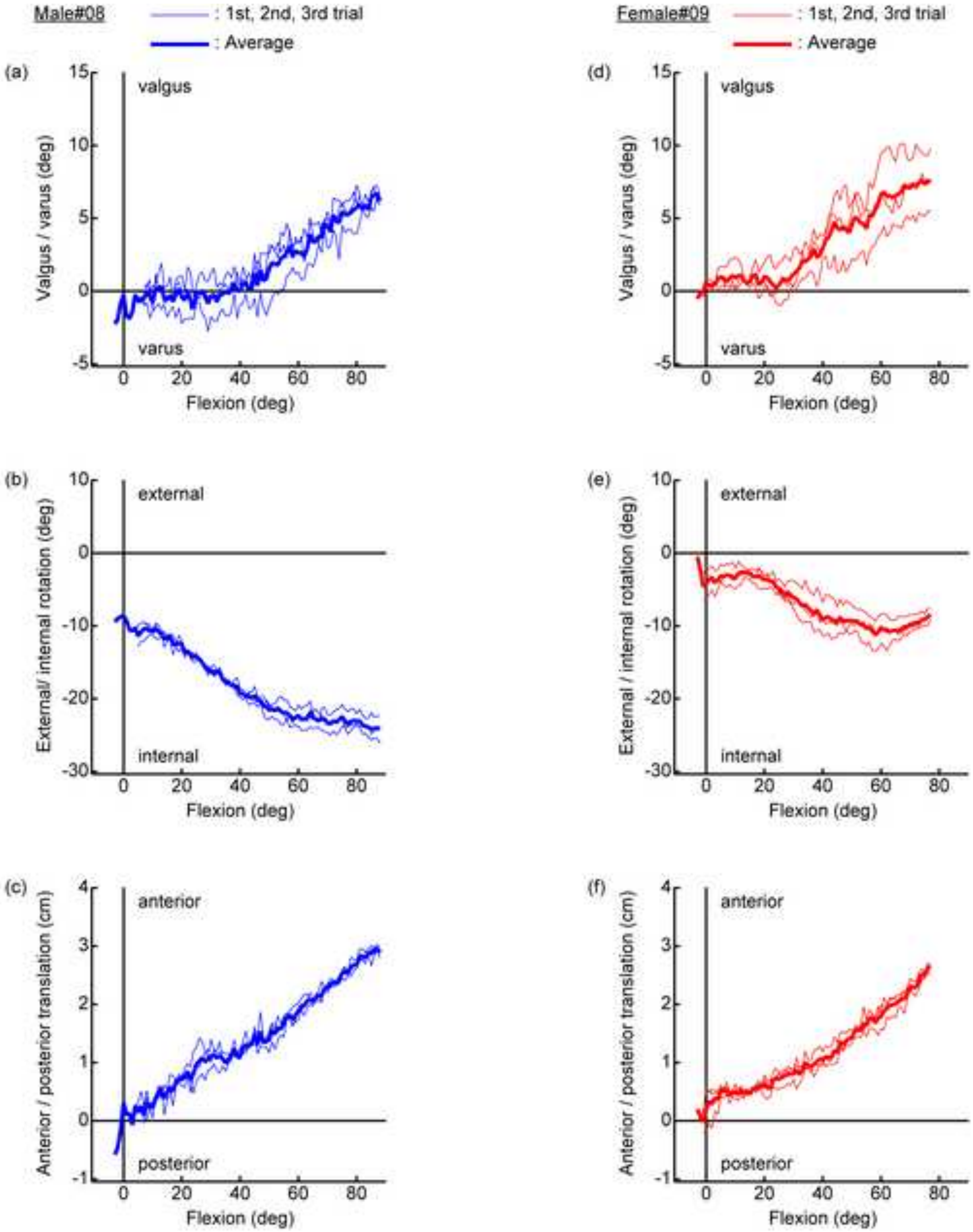

Figure 3

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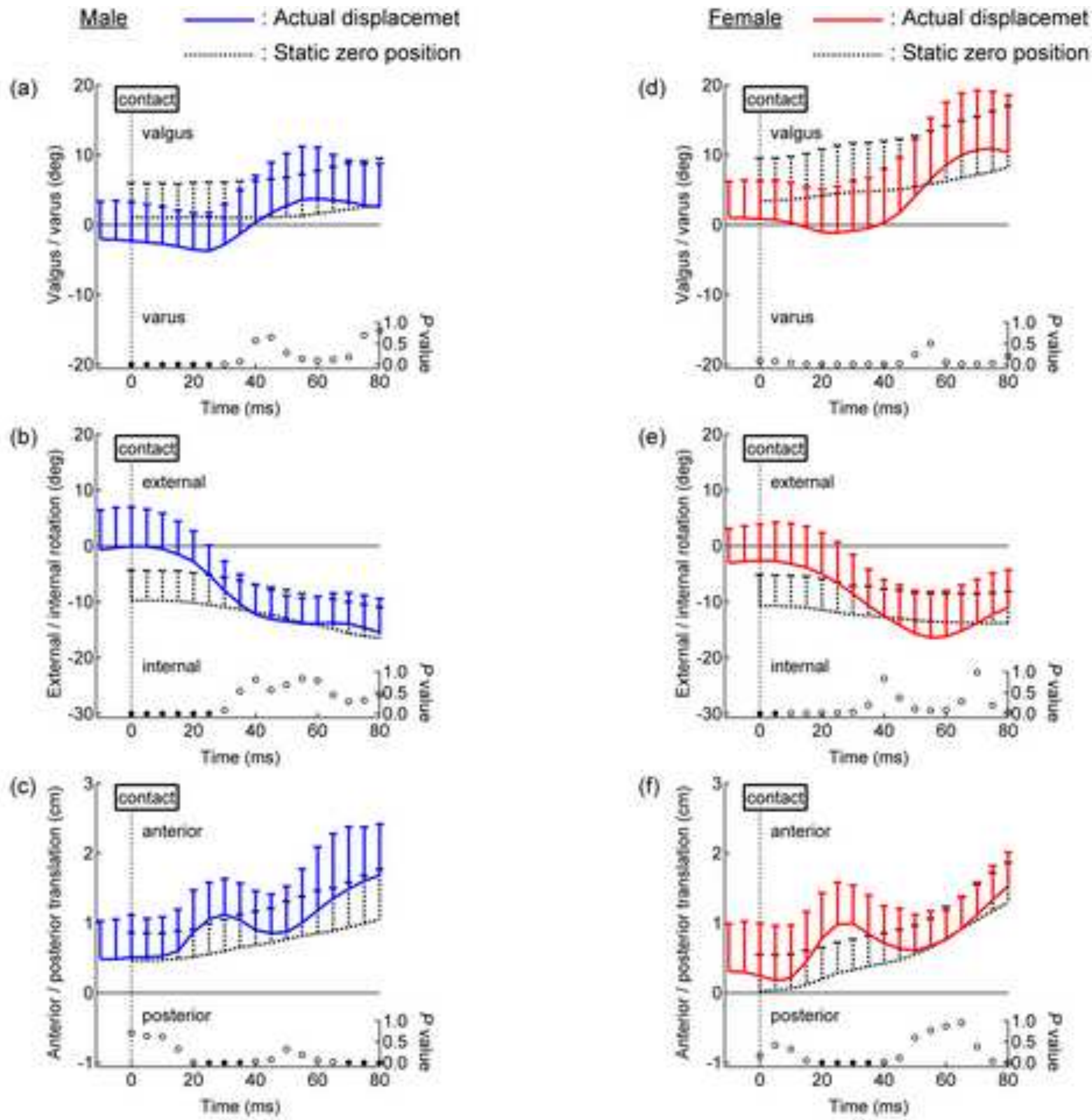

Figure 4