Hip, Knee, and Ankle Kinematics of High Range of Motion Activities of Daily Living

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Received 8 December 2004; accepted 25 October 2005

Published online 2 March 2006 in Wiley InterScience (www.interscience.wiley.com). DOI 10.1002/jor.20114

ABSTRACT: Treatment of joint disease that results in limited flexion is often rejected by patients in non-Western cultures whose activities of daily living require a higher range of motion at the hip, knee, or ankle. However, limited information is available about the joint kinematics required for high range of motion activities, such as squatting, kneeling, and sitting cross-legged, making it difficult to design prosthetic implants that will meet the needs of these populations. Therefore, the objective of this work was to generate three-dimensional kinematics at the hip, knee, and ankle joints of Indian subjects while performing activities of daily living. Thirty healthy Indian subjects (average age: 48.2 ± 7.6 years) were asked to perform six trials of the following activities: squatting, kneeling, and sitting cross-legged. Floating axis angles were calculated at the joints using the kinematic data collected by an electromagnetic motion tracking device with receivers located on the subject's foot, shank, thigh, and sacrum. A mean maximum flexion of $157^\circ \pm 6^\circ$ at the knee joint was required for squatting with heels up. Mean maximum hip flexion angles reached up to $95^\circ \pm 27^\circ$ for squatting with heels flat. The high standard deviation associated with this activity underscored the large range in maximum hip flexion angles required by different subjects. Mean ankle range of flexion reached $58^\circ \pm 14^\circ$ for the sitting cross-legged activity. The ranges of motion required to perform the activities studied are greater than that provided by most currently available joint prostheses, demonstrating the need for high range of motion implant design. © 2006 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res 24:770–781, 2006

Keywords: hip; knee; ankle; range of motion; activities of daily living

INTRODUCTION

Traditionally, the primary goal of joint arthroplasty has been to reduce pain associated with a diseased joint. Artificial hip and knee implants are commonly used when no other options are available for patients with severe arthritis at those joints. Mobility following total knee arthroplasty (TKA) or total hip arthroplasty (THA) is often considered adequate if it allows patients to walk, climb stairs, or sit on a chair.[1–4] Furthermore, according to Tew and colleagues,[4] at the time of their study, kneeling or squatting would be unattainable following TKA. This is because the maximum flexion provided by most available knee prostheses is between 110°[5] and 120°.[6] Recently, it is has become more common for patients to be able to reach higher flexion angles following TKA[7–10]; however, the majority of patients are still not able to achieve postures they had been using on a daily basis before their knee problems, which require up to 165° of flexion.[11] Ankle replacement is far less common than knee or hip replacement, with arthrodesis being the recommended method of treatment for patients in whom heavy activity is anticipated. Consequently, the range of motion (ROM) at the ankle following treatment is extremely restricted. Despite the limitations in mobility following arthroplasty, these treatments have still been considered successful for most patients in the West.

In non-Western cultures, however, it is not uncommon for treatment to be rejected due to the resulting limited ROM.[12] In most countries outside of North America or Europe, activities of daily living involve postures that require a much higher range of flexion at the joints. Squatting is an activity used for conducting activities on the floor, toileting, or simply for resting, and can easily be maintained for hours at a time.[13] In many parts of Asia, the cross-legged sitting posture is used for eating on the floor, as well as during leisure activities such as socializing. In addition, kneeling

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is an activity commonly used by Muslims in the Middle East for prayer, as well as by people in Japan for traditional ceremonies. For some patients, if the ability to return to their activities of daily living (ADL) following treatment is not assured, they will often prefer to live with the pain of arthritis.[12] It is therefore important to provide people in non-Western cultures with prosthetic implants that will restore the needed range of joint motion in order to accommodate their ADL.

Activities that require greater joint mobility, such as stretching, kneeling, and gardening, are also considered important to patients in the West.[14] Furthermore, researchers have begun to recognize the benefit of increased ROM following TKA to Western ADL, which results in higher scores as evaluated by the Hospital for Special Surgery knee rating scale 6 months after TKA.[15] Consequently, further research is required in this field to meet the needs of both Western and non-Western populations.

Little quantitative data is available on the kinematics at the joints required in the three axial planes for deep flexion activities.[6] In vivo studies have examined knee motion during kneeling[6,16] and squatting,[17] as well as hip motion during squatting[17]; however, no studies describing data collection on joint ROM while sitting cross-legged could be found. Furthermore, data collected simultaneously from more than one joint has been limited to the squatting activity with only Western subjects having been examined. It has been shown that variations in joint ROM exist across cultural groups.[18] Therefore, it is important to collect data from as many groups as possible from which orthopaedic implants can be designed, so as to meet the needs of the greatest number of people. India is a country of diverse culture and religion with many daily activities commonly performed on the floor; as a result, sitting postures that require deep flexion at the joints are of great importance to this population. Therefore, the objective of this study was to generate kinematic data at the hip, knee, and ankle joints of healthy Indian subjects during non-Western ADL, including squatting, kneeling, and sitting cross-legged.

MATERIALS AND METHODS

Data Collection Protocol

Thirty Indian subjects (10 female, 20 male) above the age of 40 years were recruited for this study. The average subject age was 48.2 years (SD 7.6), average height was 158.6 cm (SD 7.3), and average weight was 57.1 kg (SD 10.2). Consent for the study was obtained from the Medical Ethics Committee of the Sri Ramachandra Medical College and Research Institute in Chennai, India, where data was collected. Subjects had no history of joint disease or complaints of pain that would impede their ability to perform the activities being investigated. Prior to data collection, each subject was asked to fill out a questionnaire about their ethnic background and ability to perform activities that incorporated each posture being studied in order to aid in the analysis of the data. Anthropometric data was recorded for use in a kinetics study, as well as for normalization purposes.

Motion data were recorded using a six degree-of-freedom electromagnetic tracking system (Fastrak®, Polhemus, Colchester, VT) with individual receivers tracking the position and orientation of the subject's foot, shank, thigh, and pelvis. A nonconductive force plate (BP400600NC, AMTI, Watertown, MA) was used to collect ground reaction forces with applications in the calculation of one of the segment coordinate systems, as well as for a separate kinetics study. In order to minimize electromagnetic distortion from the floor, the entire system was raised onto a wooden platform 30 cm in height.

Prior to data collection, three receivers were securely affixed to the lateral surface of the subject's right foot, shank, and thigh using double-sided tape and reinforced with a 10 cm by 10 cm piece of flexible medical adhesive (Mefix®, 3M, Canada). If the medical adhesive began to detach, a larger piece of adhesive was applied over top. Locations of the receivers on each segment were chosen such that movement of the underlying soft tissue would be minimized, while simultaneously ensuring that the receiver would not interfere with the subject's movements. Large muscles and areas of fat, such as the quadriceps and around the greater trochanter, respectively, were avoided. In addition, a receiver location that would not impede high flexion motion was required. The optimal receiver location to meet the above criteria on the thigh and shank segments, therefore, was on the lateral side, one third of the segment distance in the proximal direction from the distal end of the joint, with the cable protruding distally. The foot receiver was placed on the dorsal lateral surface, approximately half a foot length anterior to the malleoli, with the cable protruding posteriorly.

The fourth receiver, equipped with a stylus, was used to digitize palpated bony landmarks on the three lower limb segments. These digitized points were used to define a local coordinate system (LCS) for each lower limb segment, which was then linked to the segment's individual receiver by means of coordinate transformations. The stylus was also used to digitize points on the top surface of the force plate, establishing its position and orientation relative to the Fastrak® transmitter, and to enable the calculation of the foot LCS. Once these force plate–transmitter and bony landmark–receiver relationships had been established, the transmitter and segment receiver could no longer be moved without invalidating the coordinate transformation calculations.

Therefore, none of the foot, shank, or thigh receivers could be used to digitize landmarks, nor could the fourth receiver simultaneously be fixed to the pelvis while digitizing bony landmarks on this segment. Consequently, following the digitization process, a reference position was recorded with the fourth receiver fixed to the sacrum that permitted the calculation of the pelvis LCS. An advantage to using the reference position to define the pelvis LCS was that a new measurement could easily be recorded as necessary throughout the data collection if the sacral receiver slipped, thereby permitting accurate data analysis following this event. Although a new reference position did not give identical results to those of the original reference position, the standard deviation of motion in all planes for the investigated activities resulting from the variation between recordings was found to be less than 1°; this result was considered to be negligible when compared to the overall ROM and standard deviation between activity trials.

Each subject was asked to perform one or more practice trials of each activity until they felt comfortable with the equipment and the given instructions. They were then asked to perform six trials of each of the following activities that they considered to be an ADL (an activity conducted at least once daily or in which the subject felt comfortable for longer than 5 min at a time): squatting with heels down (i.e., foot flat), squatting with heels up (i.e., balanced on flexed toes), kneeling with ankles dorsi-flexed, kneeling with ankles plantar-flexed, and sitting cross-legged. Each trial was performed by having the subject descend from a standing posture into the desired resting position, remaining motionless for approximately 2 s in order to ensure that the subject achieved the full resting position, and then rising to a standing posture.

Data Analysis

Local coordinate systems (LCS) were defined for the foot, shank, thigh, and pelvis segments to enable the calculation of the floating axis angles[19] at the ankle, knee, and hip joints. In general, the definition of the LCS for the thigh, shank, and foot segments followed the convention of Grood and Suntay[19] in which the y- and z-axes of the proximal segment defined the sagittal plane, normal to which was projected the flexion–extension axis (x-axis of the proximal segment). The z-axis (the long axis of the proximal segment) was initially defined, followed by the additional bony landmarks on the proximal segment that defined the frontal plane. The y-axis then followed as the normal to the frontal plane and the x-axis was the cross-product of the y- and z-axes. Rotation was assumed to occur about the z-axis of the distal segment and abduction/adduction about the floating axis (perpendicular to both the proximal x-axis and distal z-axis).

The origin of the thigh coordinate system was at the mid-point of the medial and lateral femoral epicondyles. The thigh z-axis (mechanical axis) was directed from the origin to the hip joint center, where the hip joint center was located 2 cm distal to the midpoint of the line between the anterior superior iliac spine and the symphysis pubis on the frontal plane.[20] The thigh y-axis was defined as normal to the frontal plane. The transepicondylar line, defined as the line from the medial to lateral tibial epicondyles,[21] as well as the z-axis, made up the frontal plane of the thigh. The y-axis was in the posterior to anterior direction. The thigh x-axis then followed as the cross-product of the y- and z-axes.

The shank LCS was placed at the mid-point of the medial and lateral malleoli. The shank z-axis (distal–proximal axis) was directed from the origin to the mid-point of the line from the medial to the lateral tibial condyles. The shank y-axis was the cross-product of the z-axis and the line from the medial to the lateral tibial condyles. The shank x-axis was again defined by the right-hand rule.

The origin of the foot LCS used the same origin as the shank LCS: the mid-point of the medial and lateral malleoli. The z-axis of the foot was the axis about which rotation of this segment was assumed to occur. It was defined such that it was perpendicular to the plantar aspect of the foot, which was taken to be parallel to the top surface of the force plate when the subject was in a standing position. The y-axis was consequently parallel to the surface of the force plate and in the direction of the line from the origin to the second metatarsal head. The x-axis was defined by the right-hand rule, that is, as the cross-product of the y- and z-axes.

The pelvis LCS was determined with the help of the reference position, in which the subject was aligned in a known orientation relative to the Fastrak® transmitter. Using coordinate transformations, the orientation of the pelvis could then be calculated relative to the receiver fixed to the sacrum of the subject. The reference position required the subject to stand with his feet symmetrically aligned about the midline of a template on the force plate whose position relative to the transmitter had previously been determined. With the subject standing upright with knees fully extended and weight evenly distributed on both feet, it could reasonably be assumed that the trunk followed the established orientation of the feet in both the frontal and transverse planes. An inclination of the trunk in the frontal plane (i.e., asymmetrical abduction/adduction of the hips) would have resulted in a center of pressure measurement at either side of the template midline. Furthermore, malalignment in the transverse plane could only have been achieved with a twisting of the trunk, which would have been apparent to the investigator in both the frontal and sagittal plane views. Sagittal plane alignment was achieved by ensuring that the subject's lateral malleolus, lateral tibial condyle, greater trochanter, and acromion process of the shoulder formed a vertical line (i.e., a line perpendicular to the surface of the force plate) in that plane.

Once the orientations of the LCS for each segment were known, relative angles could be calculated using the floating axis method developed by Grood and Suntay.[19] Position and orientation data of each of the four receivers, from which segment LCS and joint angles could be calculated, were acquired at a sample rate of 30 Hz. Each trial was split into two phases: getting into position with gravity (standing to sitting posture) and getting out of position against gravity (sitting to standing posture), with each phase normalized to a 100% cycle. By dividing the data into two phases, the varying length of the rest period (approximately 2 s) at mid-trial was eliminated. The start of each phase of the trial was triggered by movement at any of the three joints, while the end of each trial was defined as the termination of motion at all three joints. The mean curve of the six individual trials was found for each subject and group means and standard deviation curves were calculated from the subject means.

The maximum, minimum, and ROM values were extracted from each curve before the trial was split into the two phases (getting into and out of position). Because these values were not time dependant, resting period data at the start and end of each phase of the trial could be included to ensure that the extreme ranges of motion were not omitted. The means and standard deviations of all subject values were then calculated from each individual subject mean to produce overall group statistics.

Data was not collected or excluded for a subject if the particular activity was not considered an ADL, if a receiver slipped making the data invalid, or due to equipment failure.

RESULTS

Figures 1 through 5 show the joint angles of the ankle, knee, and hip joints for the activities studied. Mean curves are presented with one standard deviation in each of the three axial planes. Mean kinematic angles of significance are listed with standard deviations in Tables 1 through 3. The number of subjects from whom data was collected for each activity was also included with the figures and in the tables.

All five activities show a prevailing motion in the sagittal plane (flexion–extension) at all three joints: ankle, knee, and hip. However, flexion angles did not always reach a maximum while in the final resting posture. As demonstrated in Figures 3–5 for the two kneeling and sitting cross-legged activities, maximum flexion angles at all three joints occurred closer to 50% of the cycle into position and out of position phases of the trial (where resting posture occurs at 100% of the into position cycle and 0% of the out of position cycle).

Mean maximum flexion angles at the knee joint reached values greater than 150° for both types of squatting, as well as for the kneeling with

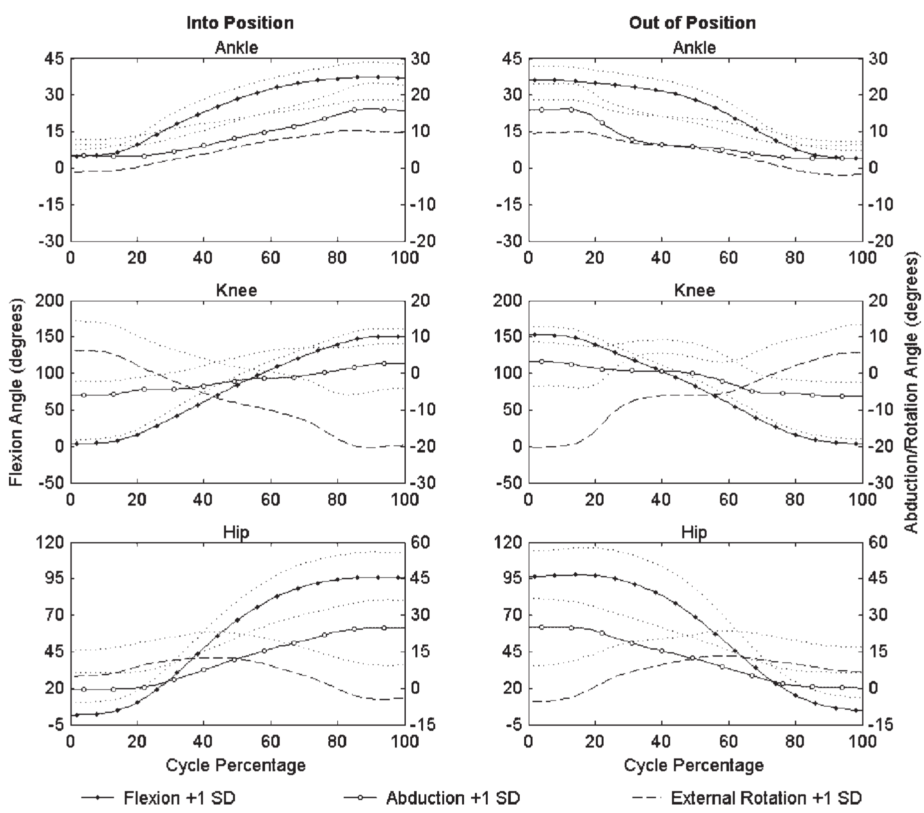

Figure 1. Squatting heels down group mean angles and standard deviations at the three joints in the three axial planes. Ankle and knee: $n = 26$; hip: $n = 25$.

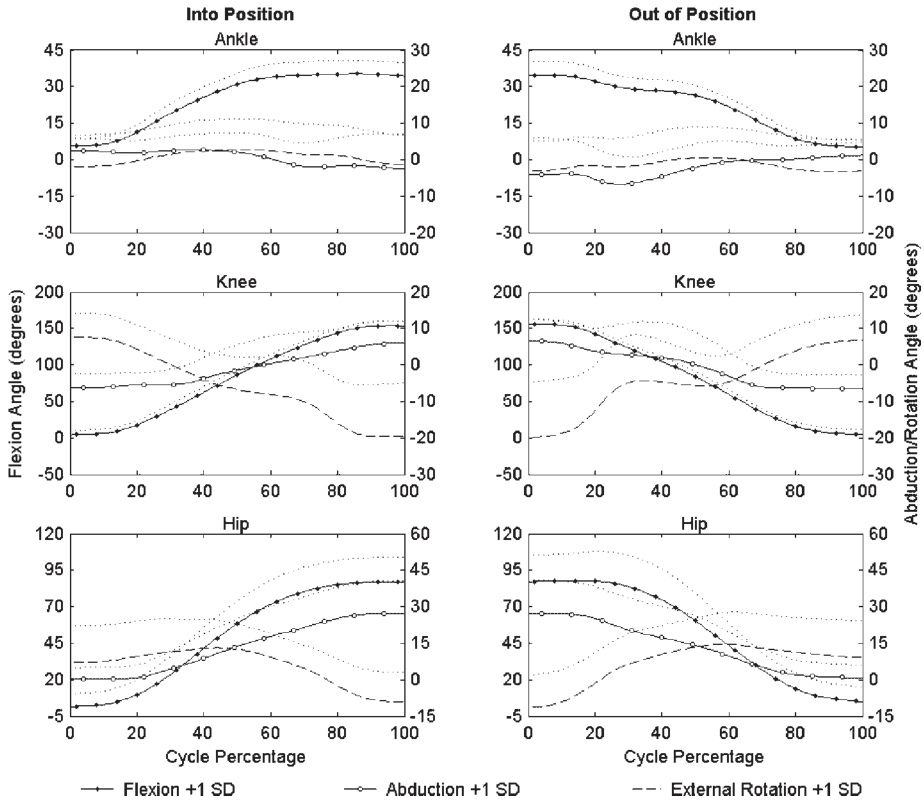

Figure 2. Squatting heels up group mean angles and standard deviations at the three joints in the three axial planes. Ankle and knee: $n = 26$; hip: $n = 26$.

ankles dorsi-flexed activity. Mean maximum ankle range of flexion requirements varied between $34^\circ \pm 6^\circ$ for the squatting with heels up activity to $58^\circ \pm 14^\circ$ for the sitting cross-legged activity. The greatest ROM at the hip joint in the flexion–extension plane was for the squatting with heels down activity, for which the mean ROM was $95^\circ \pm 26^\circ$.

Internal rotation at the knee joint accompanied high flexion during all activities. The greatest mean maximum knee internal rotation was required for the sitting cross-legged activity in which the shank rotated inward an average of $33^\circ \pm 12^\circ$ with respect to the thigh segment. This maximum rotation occurred during deep flexion of the joint.

The greatest ROM in the frontal plane (eversion–inversion for the ankle and abduction–adduction for the knee and hip) occurred during the sitting cross-legged activity, as shown in Tables 1–3, respectively. Angles in this plane, however, had relatively large standard deviations, so the difference between the range required for sitting cross-legged and the other activities investigated was not significant.

DISCUSSION

Few studies reporting kinematics at all three lower limb joints during squatting, kneeling, or sitting cross-legged activities have been published with which to compare our data. Our findings agree well with the literature available[11,16,17]; in those cases in which our data varies from results of other studies, the discrepancies can generally be attributed to variations in the method by which the activity was performed or differences in the physical characteristics of the population groups studied. These distinctions underline the importance of collecting data from different populations and to allow every subject to perform each activity in the manner most comfortable for him.

Subjects were not instructed on the method by which they were to get into and out of each activity position because it was found that this resulted in unnatural motion. Methods by which an activity was performed occasionally varied between individuals. As a result, high standard deviations accompanied motion angles at the three joints for those activities that were less repetitive, in particular, kneeling and sitting cross-legged.

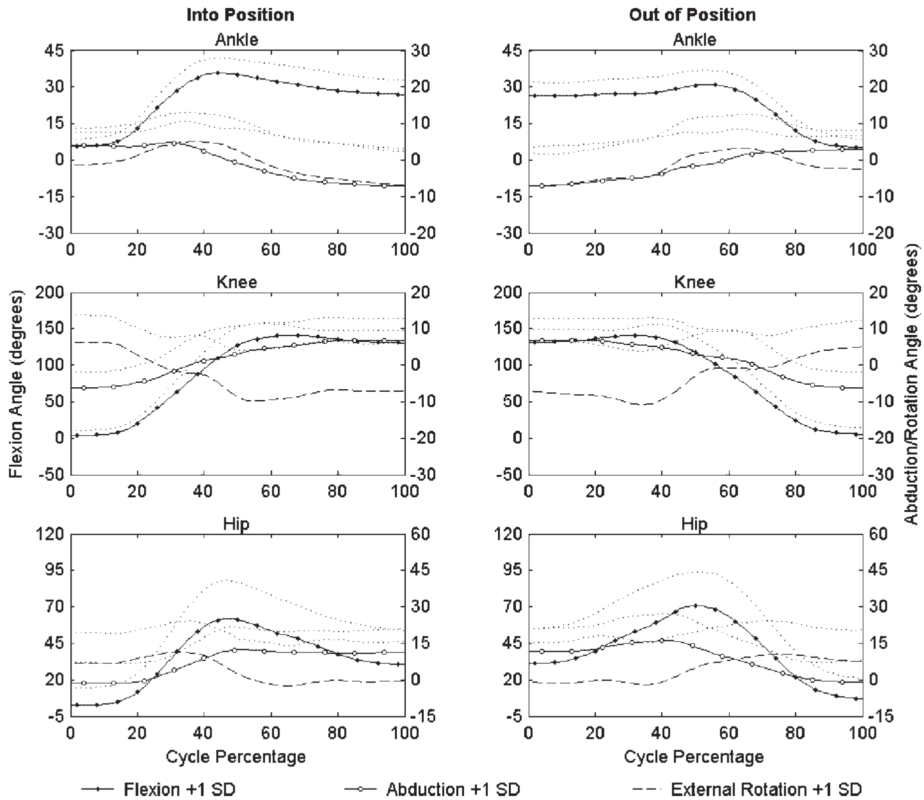

Figure 3. Kneeling dorsi-flexed group mean angles and standard deviations at the three joints in the three axial planes. Ankle and knee: $n = 26$; hip: $n = 24$.

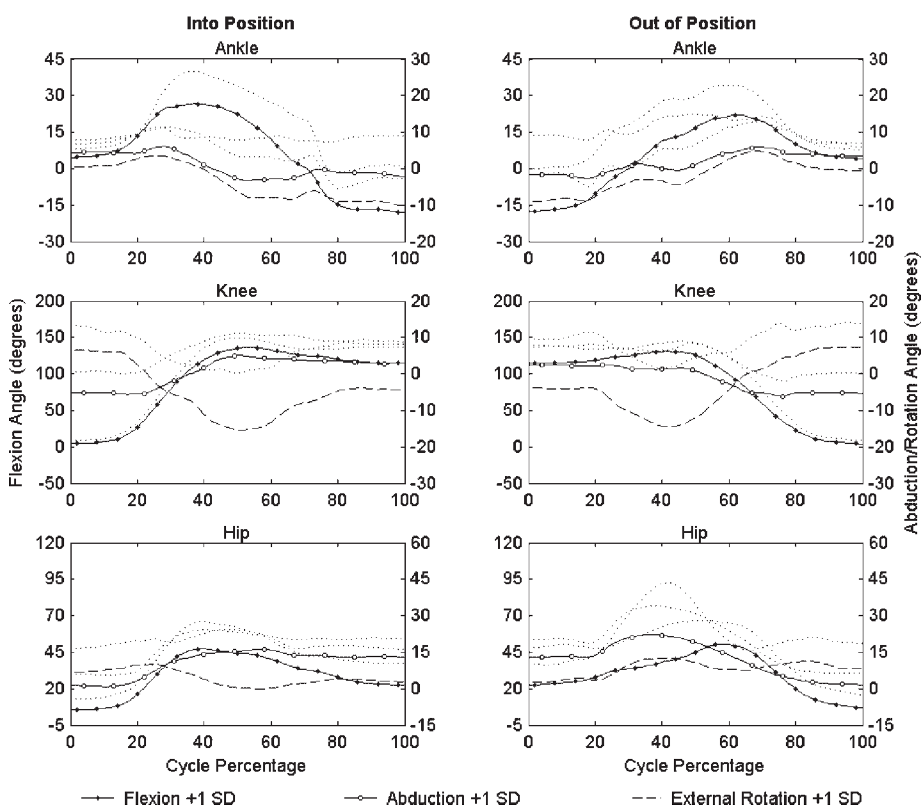

Figure 4. Kneeling plantar-flexed group mean angles and standard deviations at the three joints in the three axial planes. Ankle and knee: $n = 6$; hip: $n = 6$.

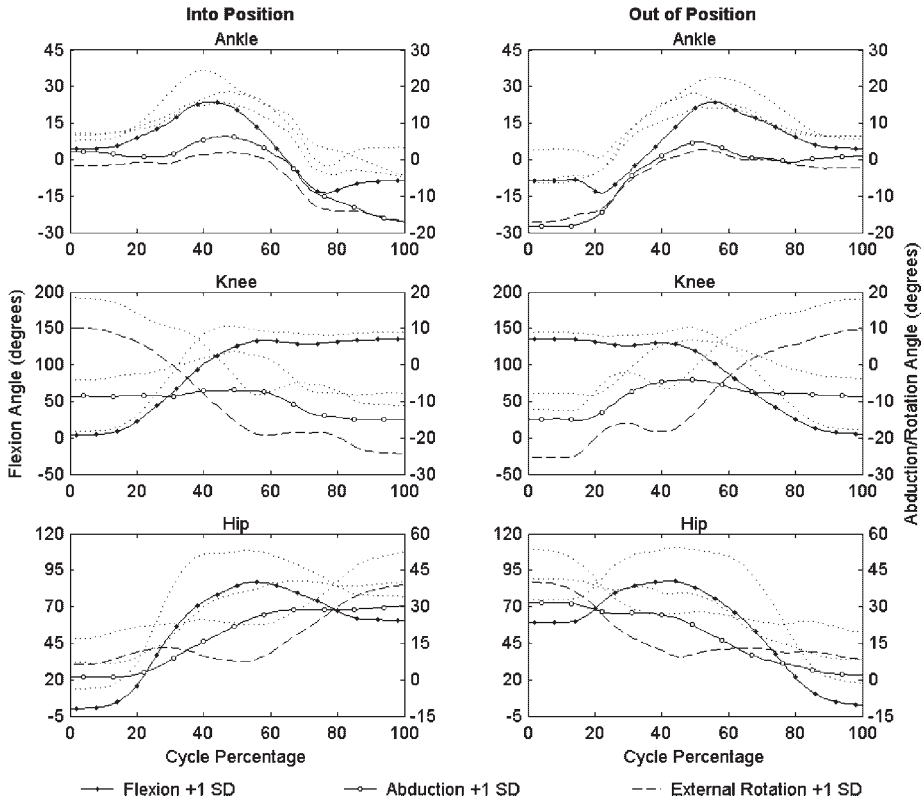

Figure 5. Sitting cross-legged group mean angles and standard deviations at the three joints in the three axial planes. Ankle and knee: $n = 26$; hip: $n = 23$.

Not only was kneeling with ankles plantar-flexed less repetitive than squatting for these subjects, it was also not considered an ADL by many subjects. For those who considered this an ADL, it was observed that the sitting style was unlike that assumed by Muslims for prayer. In fact, only 1 of the 30 subjects was Muslim, which reflects the predominantly Hindu and Christian culture in

Table 1. Group Mean Ankle Kinematics for the Five Activities in the Three Axial Planes

	n	Maximum Angle (°)	SD (°)	Minimum Angle (°)	SD (°)	ROM (°)	SD (°)
Dorsi-flexion							
Squatting heels down	26	38.5	5.9	3.1	2.8	35.4	5.5
Squatting heels up	26	37.7	5.1	3.6	2.5	34.1	5.6
Kneeling dorsi-flexed	26	39.7	5.1	3.3	3.0	36.4	5.8
Kneeling plantar-flexed	6	32.7	6.9	−23.8	16.0	56.5	18.7
Sitting cross-legged	26	32.3	7.1	−26.1	10.7	58.4	14.1
Eversion							
Squatting heels down	26	17.7	6.5	0.0	4.2	6.1	6.1
Squatting heels up	26	6.6	4.0	−10.1	8.0	6.5	6.5
Kneeling dorsi-flexed	26	9.4	5.1	−12.5	9.4	7.6	7.6
Kneeling plantar-flexed	6	12.1	4.1	−12.4	10.5	7.2	7.2
Sitting cross-legged	26	17.2	7.8	−24.7	10.6	11.9	11.9
External rotation							
Squatting heels down	26	11.7	7.9	−2.8	8.9	14.5	4.3
Squatting heels up	26	5.6	7.7	−8.2	7.4	13.8	5.1
Kneeling dorsi-flexed	26	7.8	8.3	−12.6	9.2	20.4	5.4
Kneeling plantar-flexed	6	7.7	7.2	−19.2	8.4	26.8	7.9
Sitting cross-legged	26	10.1	8.9	−27.5	9.7	37.7	10.9

Table 2. Group Mean Knee Kinematics for the Five Activities in the Three Axial Planes

	n	Maximum Angle (°)	SD (°)	Minimum Angle (°)	SD (°)	ROM (°)	SD (°)
Flexion							
Squatting heels down	26	153.7	10.4	1.6	5.6	152.0	11.0
Squatting heels up	26	156.9	5.7	1.8	5.6	155.1	7.2
Kneeling dorsi-flexed	26	154.9	8.6	1.5	6.4	153.4	9.6
Kneeling plantar-flexed	6	144.4	13.2	1.5	3.6	142.9	13.5
Sitting cross-legged	26	150.0	8.1	1.4	6.0	148.5	8.4
Abduction							
Squatting heels down	26	6.0	5.2	−9.6	5.3	15.6	5.8
Squatting heels up	26	8.3	5.2	−10.3	6.5	18.7	6.4
Kneeling dorsi-flexed	26	10.2	5.8	−9.6	6.7	19.8	5.0
Kneeling plantar-flexed	6	7.2	5.5	−9.5	5.0	16.6	2.9
Sitting cross-legged	26	1.0	7.9	−21.0	8.3	21.9	5.4
External rotation							
Squatting heels down	26	9.2	7.5	−23.6	14.0	32.8	14.3
Squatting heels up	26	10.8	6.5	−22.5	14.3	33.3	14.8
Kneeling dorsi-flexed	26	11.1	7.1	−18.4	14.7	29.5	15.0
Kneeling plantar-flexed	6	12.4	6.0	−19.7	15.3	32.0	14.6
Sitting cross-legged	26	14.5	9.6	−32.7	12.3	47.2	16.3

that region of India. During both types of kneeling, instead of sitting with their buttocks resting on their heels, most subjects remained with their thigh and trunk upright, resulting in low hip flexion angles. Furthermore, knee flexion angles in the final plantar-flexed kneeling position were less than one would expect in the Muslim prayer position, such as the values obtained by Hefzy and colleagues[6] for kneeling by Saudi subjects, because deep flexion was not attained. The discrepancy between the values obtained by Hefzy and colleagues ($157.3° \pm 4.9°$) and those in our study ($143° \pm 14°$) can be attributed to the method in which this activity was performed by the Chennai subjects. Of the six subjects who felt comfortable kneeling with ankles plantar-flexed, only one sat back with his weight on his heels in the resting position, while the others remained with their knees flexed to about 90° and the thigh and torso in an upright position. This variation in the method in which this activity was performed also partially accounts for the higher standard deviation in knee

Table 3. Group Mean Hip Kinematics for the Five Activities in the Three Axial Planes

	n	Maximum Angle (°)	SD (°)	Minimum Angle (°)	SD (°)	ROM (°)	SD (°)
Flexion							
Squatting heels down	25	95.4	26.6	0.0	7.9	95.4	26.2
Squatting heels up	26	90.6	19.3	−0.7	8.6	91.3	17.1
Kneeling dorsi-flexed	24	73.9	33.3	0.0	10.4	73.9	29.4
Kneeling plantar-flexed	6	62.0	16.2	3.2	8.6	58.8	9.7
Sitting cross-legged	23	83.5	35.7	−1.9	12.6	85.4	34.2
Abduction							
Squatting heels down	25	26.1	11.6	−2.1	7.2	28.2	13.9
Squatting heels up	26	29.5	13.1	−2.2	5.8	31.7	11.2
Kneeling dorsi-flexed	24	21.3	12.2	−4.0	8.3	25.3	15.3
Kneeling plantar-flexed	6	26.7	9.3	−0.9	5.3	27.6	12.5
Sitting cross-legged	23	34.1	14.7	−2.4	6.6	36.5	15.0
External rotation							
Squatting heels down	25	16.5	10.5	−9.2	11.6	25.7	11.8
Squatting heels up	26	18.8	11.4	−14.9	12.7	33.7	12.7
Kneeling dorsi-flexed	24	16.0	12.9	−12.1	16.1	28.1	12.8
Kneeling plantar-flexed	6	25.1	23.0	−8.9	15.1	34.0	14.9
Sitting cross-legged	23	37.1	17.7	−3.2	12.5	40.3	18.4

flexion angle compared to the Saudi subjects. The mean maximum knee flexion angle for the kneeling dorsi-flexed activity, on the other hand, was $155^\circ \pm 9^\circ$ for the Chennai subjects, much closer to the mean value obtained by Hefzy and colleagues.[6] For the kneeling with ankles dorsi-flexed activity, a greater proportion of the Chennai subjects sat with their buttocks resting on their heels in the into position posture (1:2 during kneeling dorsi-flexed versus 1:6 during kneeling plantar-flexed).

A high standard deviation associated with the squatting range of hip flexion is again evidence that the manner in which this activity was performed varied from subject to subject. Whereas some subjects attained peak hip flexions of less than 90°, which was lower than mean values reported in the literature,[11,17,22] several subjects reached maximum hip angles of greater than 110° and one subject required over 140° of flexion at the hip to perform the squatting activities. The reason for lower hip flexion angles may be attributed to the curvature of the spine during high ROM activities for several of the Chennai subjects. Instead of obtaining all necessary flexion at the hip joint center, it appears that some flexion comes from the lumbar region of the spine. Figure 6 shows the sagittal view of the thigh, shank, and foot segments, in addition to the pelvis local coordinate system, at the start and end of the into position phase of the squatting heels down trial. Photographs of the subject in these two positions taken during the data collection (from a different angle) are shown in Figure 7. The counterclockwise rotation of the pelvis LCS from the initial standing posture to the final squatting position is clearly shown in Figure 6. As a result, once in the squatting position, there was a smaller angle between the pelvis and thigh LCS than anticipated.

While the maximum hip flexion required for the squatting activity varied between subjects, knee flexion angles were not only more consistent, but also quite high, demonstrating that subjects went into deep flexion to achieve this posture. The mean ROM in the flexion–extension plane for the two types of squatting (with heels down and with heels up) were $152^\circ \pm 11^\circ$ and $155^\circ \pm 7^\circ$, respectively. These results demonstrate a much larger range required for squatting than that of $101^\circ \pm 21^\circ$ reported by Flanagan and coworkers[17] for a group of Western subjects. The greater ROM required by the Indian subjects may be attributed to various differences in the subject groups. Not only were the Indian subjects younger than the Western subjects (with a mean age of 48.1 ± 7.6 years vs. 74.5 ± 4.39 years for the Western subjects), but they were also smaller in terms of average body mass index (22.7 kg/m^2 vs. 25.3 kg/m^2 for the Western subjects). A smaller body mass index indicates less soft tissue to restrict motion in deep flexion. Most importantly, the Indian subjects were accustomed to performing this activity on a daily basis, whereas the Western subjects likely were not.

The negative external rotation angles at the knee joint (i.e., internal rotation of the shank with

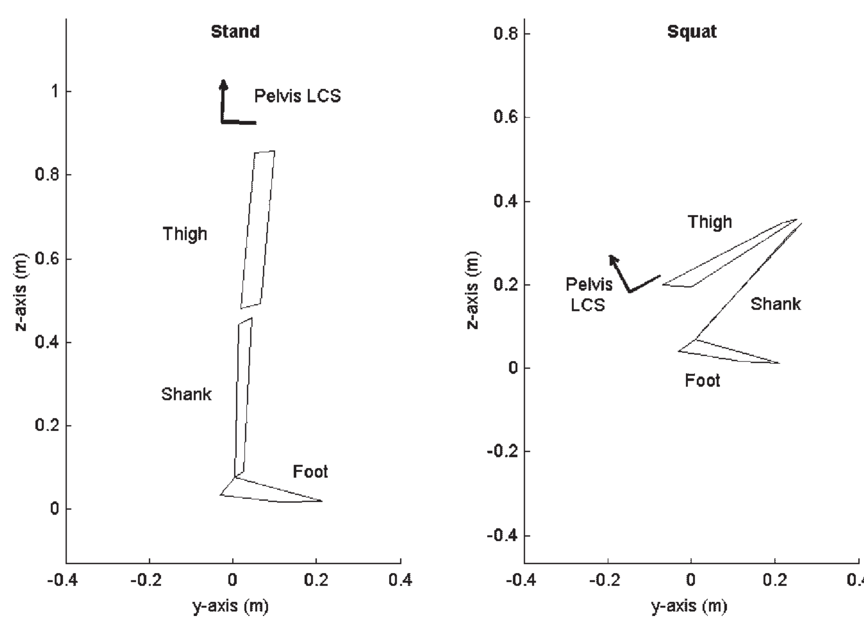

Figure 6. Sagittal plane rotation of pelvis segment results in smaller hip flexion angles (start and end of squatting heels down activity into position phase).

Figure 7. Subject at start and end of squatting heels down activity into position phase.

respect to the thigh) during deep knee flexion that was found in all activity mean curves are consistent with the data produced by Hefzy and colleagues,[6] Li and coworkers,[23] and Nakagawa and colleagues.[24] In the study conducted by Li and coworkers,[23] the tibia rotated internally throughout the entire range of passive flexion. With the addition of quadriceps muscle loads applied to the joint, further internal knee rotation was observed as the flexion angle increased from 120° to 150°.[23] A comparable observation was made by Hefzy and colleagues[6] using biplanar radiographs taken from healthy subjects at three different stages of kneeling. Their data suggested a greater amount of femoral roll-back on the lateral side than on the medial side during deep flexion.[6] Similarly, an internal tibial rotation of about 28° at the higher end of the knee flexion range was observed by Nakagawa and coworkers.[24] This fact also explains the internal knee rotation observed for all five deep flexion activities investigated in this study.

By comparing the data in the current study to data reported on the ROM achieved following TKA, the need for an improvement in joint arthroplasty is demonstrated. Two studies that monitored the outcome of TKA using knee replacements specially designed for high flexion reported mean ROM just over 120°.[9,25] Despite superior designs to previous implants, these prostheses would not accommodate the majority of subjects in the current study who regularly performed squatting, kneeling, and sitting cross-legged activities on a daily basis, as their mean knee ROM in the flexion–extension plane was consistently over 140°. Only one study reported the ability of their patients to achieve over 145° of flexion in 50% of the knees requiring replacement surgery.[7] Of those, only 10 of 18 knees were able to achieve *seiza* style sitting following TKA, a customary posture involving very deep flexion in the Japanese culture, demonstrating that certain activities of daily living require either higher degrees of flexion or greater motion in the other axial planes for which the current prosthesis was not adequately designed.

Total ankle arthroplasty (TAA) is still far less widespread than knee and hip arthroplasty. It is difficult to implant stable, painfree implants, especially for users who heavily load the joint. Therefore, arthrodesis is often recommended over ankle replacement, which naturally limits the ROM at the joint. The ROM required for the five activities investigated in this study ranged from $34^\circ \pm 6^\circ$ for the squatting with heels up activity to $58^\circ \pm 14^\circ$ for the sitting cross-legged activity, with mean maximum dorsi-flexion angles reaching over 32° for all activities. Even in successful ankle arthroplasty, the patient's ability to flex and extend the joint is severely limited and would not be able to accommodate this ROM. In one recent study following TAA, the mean ROM at the ankle was only 36° and of those patients investigated, only 39% were able to dorsi-flex the ankle beyond neutral![26] Similarly, two other studies reported mean ROM of 27° and 28° following TAA,[27,28] one of which again reported only 59% of patients with the ability to dorsi-flex the ankle greater than or equal to 10°.[28] Although these studies considered this outcome adequate for normal walking in which 30° of ankle motion is considered sufficient, the ROM provided by these TAA surgeries would not allow most of these patients to perform any of the activities investigated in the current study.

The ideal implant should be designed to allow the patient to achieve the kinematics of a healthy joint. The data presented in this study, together with an understanding of the kinetics at the joint, could be used for a finite element analysis of the implant components or to test the implant in vitro, thereby improving the design of total joint arthroplasty. Designing prostheses to accommodate higher ranges of motion is important, not only to meet the mean ranges of motion reported in the current study, but also to accommodate the needs of those people who fall into the higher ROM percentile, as reflected by the higher standard deviations, for example, at the hip joint. Challenges in positioning the implant optimally during surgery also reinforce the need to design implants that surpass the mean kinematic values observed. If a surgeon is not able to place the implant in the ideal location to use the ROM to its greatest potential, an implant designed to reach ranges of motion 5° to 10° greater than that required by the patient should be used. Furthermore, the activities studied here have primarily measured active ROM, whereas passive motion would generate even greater joint angles. Therefore, the data presented in this study should be considered the minimum ROM requirements in prosthesis design in order to permit a return to these activities of daily living.

Errors associated with measurement of human movement are inevitable. The three-dimensional model assumes that the Fastrak® receivers used to locate the lower limb segments were rigidly attached to the body. However, because the receivers were merely fixed to the skin, an error that was associated with the movement of the soft tissue surrounding the more rigid underlying bone was accrued. It has been shown that the thigh is more affected by soft tissue artefact than the shank[29,30] and that the proximal areas of the segment accrued greater error than the distal and lateral areas.[30,31] The placement of the receivers in the current study was on the lateral side at one-third of the segment length from the distal end, where it would be less affected by soft tissue movement.[29–32]

Several studies have also advised caution in interpreting the data in the transverse and frontal planes at the knee joint, as these are prone to greater inaccuracies due to soft tissue artefact than in the flexion–extension plane.[29–32] In the current study, rotation and abduction–adduction angles must also be considered subject to greater error than angles in the sagittal plane. However, most subjects had little excess soft tissue as indicated by their low mean body mass index (22.7 kg/m^2). Furthermore, the activities studied were low impact, suggesting negligible error due to inertial effects.[31] We therefore consider the error induced by soft tissue movement to have had a minimal effect on our results. This is supported by the results reported in other studies using imaging techniques to examine deep flexion motion,[6,24] in which the rotation angles at the knee were similar to those of the current study.

It is possible that the motion observed in the frontal and transverse planes was caused by kinematic crosstalk, an occurrence that arises when the segment's local coordinate system is not aligned with the axes about which rotations are assumed to occur.[33] Since the ROM in the non-sagittal planes remained within reasonable limits, even at extreme flexion angles, and was similar to results observed by other researchers, it is not believed that error due to kinematic crosstalk could have had a significant effect on the results.

CONCLUSION

Three-dimensional kinematics were generated at the hip, knee, and ankle joints for Indian activities of daily living. The variations of squatting, kneeling, and sitting cross-legged that were studied demonstrated high degrees of flexion at all three joints. In fact, the ranges of motion required by most subjects to complete the activities were greater than what can currently be accommodated by existing prostheses. In order to meet the needs of non-Western populations, prosthetic implants must be developed to reach flexion angles great enough to account for the variations in methods in which the activities are performed by different people, as well as possible imprecise surgical positioning. Furthermore, high flexion designs must account for concomitant motion in the frontal and transverse planes. Patients in North America and Europe who live a more active lifestyle will, no doubt, also benefit from high ROM prostheses.

ACKNOWLEDGMENTS

This work was financially supported by NSERC, Zimmer, Inc., and the Human Mobility Research Centre. The authors acknowledge Dr. Sabin Visnarath, Vani Malar, Amanda Knutson, Onno Oosten, and Alex Pappas for their contributions.

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