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Impacts of Robotic Compliance and Bone Bending on Simulated *in vivo* Knee Kinematics

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Abstract Robotic testing offers researchers the opportunity to quantify native tissue loads for the structures of the knee joint during activities of daily living. These loads may then be translated into design requirements for future treatments and procedures to combat the early onset of knee degeneration following an injury. However, high knee loads during testing have the potential to deflect a robotic end effector and cause inaccuracies in the applied kinematics. Furthermore, bone bending could also induce kinematic change. This study aimed to quantify the effects of robotic compliance and bone bending on the accuracy of simulated in vivo kinematics in a KUKA KRC210 serial robotic system. Six (6) human cadaver knees were subjected to cyclic human gait motion while 6 DOF forces and torques were recorded at the joint. A Vicon T-Series camera system was used to independently record the applied kinematics. Periods of highest kinematic deviation occurred during instances of low joint loading, suggesting negligible levels of forced deflection for simulations of moderate levels of activity while results of this small study indicate that high physiologic loading poses low risk of deviation from target kinematics, further testing is necessary to confirm.

Keywords Knee, Robotics, Simulation, Gait, Kinematics

1. Introduction

Current treatment and reconstruction methods of native knee tissues fail to prevent the early onset of osteoarthritis (OA) and knee joint degeneration. [1] Due to the difficulties associated with the identification of unmet functional demands via direct in vivo measurement, many researchers have developed robotic testing platforms to study tissues during simulated kinematics in the laboratory. [2-8] These methodologies allow for the *in vitro* investigation of mechanical behaviors within the tibiofemoral joint during simulated clinical tests [6, 9-11] as well as *in vivo* recorded gait [12] and athletic tasks. [13] It has been demonstrated that these robotic models can reproduce joint motion with excellent intra- and inter-specimen reliability. [13] However, this reliability does not implicate that robotic models of joint articulation are without limitation.

One such robotic limitation lies within the relative joint stiffness of the robotic manipulator and biologic specimen to be examined. [14] While several studies have demonstrated highly accurate methods of collecting kinematic data during activities of daily living (ADLs), [15-21] only limited data exists to support the accurate reproduction of these kinematics on human knee specimens using robotics. [13, 22] Further, though the referenced study comments on the reliability of robotic simulation with identical input, it does not explicitly quantify the influence of confounding factors such as system stiffness. While inputs would be precisely executed in theoretical systems with infinite stiffness, real systems show some degree of compliance, which makes it possible for the high knee loads generated during ADLs to alter the path of a robotic end effector or cause bone bending during testing. [14] This could result in altered kinematics applied to the joint.

Accurate reproduction of in vivo kinematics requires that the system itself be much stiffer than the test object (the knee joint). While this is intuitive, the actual compliance of the robot and bony fixation interfaces in our lab have yet to be quantified. The objective of this study was to better understand the implications of system stiffness and to assess deviation within our own robotic laboratory model. Specifically, this study aimed to 1) apply previously recorded *in vivo* kinematics to human knees *in vitro*, 2) independently measure simulated kinematics and applied loads, 3) quantify the effects of robotic compliance and/or bone bending. This information will allow for the assessment of the kinematic accuracy of applied physiologic motions, and will be invaluable to proceed with this robotic testing platform in the investigation of more aggressive ADLs that produce higher loads across the joint. It was hypothesized that; a) the kinematic errors generated from the present robot

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model would be smaller in magnitude than previously reported kinematic errors associated with *in vivo* motion analysis, and that b) the magnitudes of these kinematic errors would correlate to magnitudes of joint loading for any given time point.

2. Materials and Methods

2.1. Robotic Manipulation

Three specimens were procured and successfully tested from the University of Cincinnati Body Donor program and a tissue bank (Anatomy Gifts Registry, Hanover, MD) with a mean age of 77.7 ± 19.6 year and mass of 71.9 ± 21.1 Kg. All specimens were dissected free of soft tissue, leaving only the menisci and the cruciate and collateral ligaments with surrounding joint capsule intact. For each specimen, the tibia was potted into a custom-built adjustable fixture using PMMA bone cement. A PMMA mold was also applied to the femur as close to the tibiofemoral joint as possible, which allowed the bone to be rigidly bolted into a fixture on the stationary testing platform where the simulations were performed. Detailed accounts of specimen preparation and mounting are documented in previous literature. [7]

A six-camera, infra-red motion tracking system externally recorded specimen motion (Vicon; T-Series). Sixteen (16) reflective markers (10mm diameter) were attached directly to the bony anatomy of each specimen with glue and double-sided tape. Clusters of 4 markers each were attached to 1) the base of the robotic end effector, 2) near the tibiofemoral joint line of the proximal tibia, 3) near the tibiofemoral joint line at the distal femur, and 4) the proximal femur (Figure 1). Markers were placed as close to the joint landmarks as possible to prevent relative motion between them, allowing the cameras to capture the true anatomic motion of the knee. The cameras were placed to surround the robot work volume and marker trajectories were sampled throughout simulation at 120 Hz. Camera calibrations revealed an average residual of 0.674 ± .09 mm.

Anatomical axes were established using the Grood/Suntay method. [23] Using a coordinate measurement machine (CMM), the tibial fixture was rigidly fixed to the robot end effector such that the anatomical axis of the tibial coordinate system was aligned with the primary loading axis of the robotic load cell. The robot was then driven to a position that minimized the loads within the tibiofemoral joint with the femur secured to the testing platform. The CMM registered the anatomic position of the knee along with the relative positions of each of the 16 reflective markers within the robotic coordinate system (Figure 2). This initial measurement was fed into a custom MATLAB program that calculated the corresponding robotic moves needed to align the knee to the starting orientation at mid-stance of gait. Once all angles were verified, translational loads were minimized to establish an initial start point for robotic manipulation. Compression was added to this starting point as needed to achieve a peak force between 2.0-2.5 times body weight [24] when gait motion was applied.

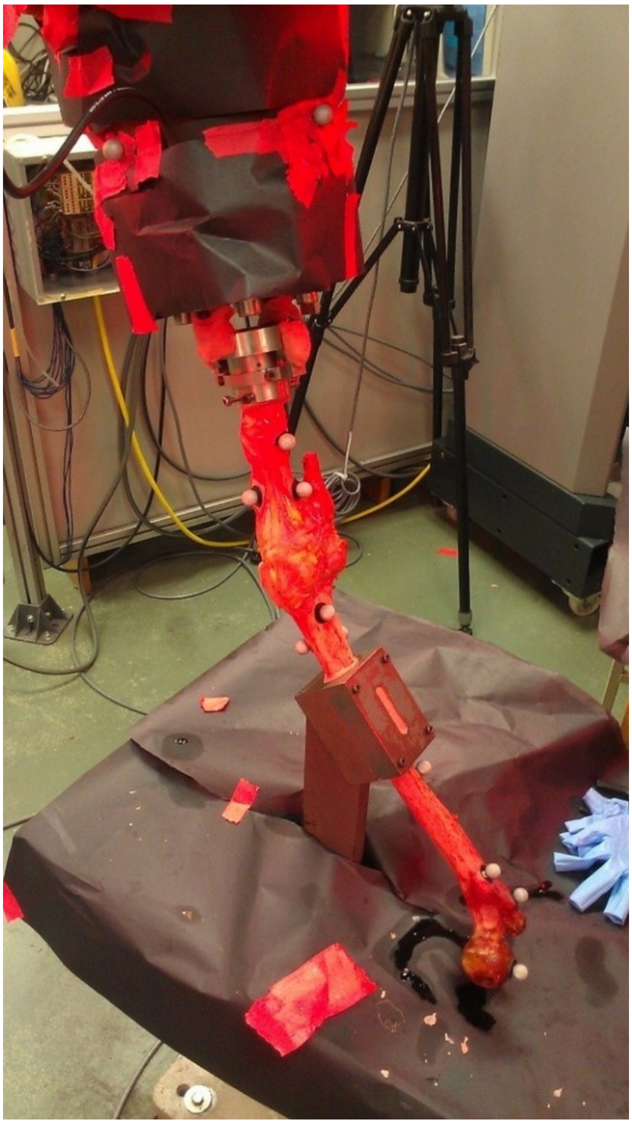

Figure 1. Posterior view of a specimen with reflective markers attached. All other potentially reflective surfaces were covered with black matte paper

A 6 DOF gait motion previously reported in the literature [18] was digitized to be reproduced by a KUKA KRC210, position-accurate, serial robot with a 210 kg rated payload (~2.5 x BW). Peak flexion during mid-stance was selected as the start and end point for each cycle of gait. All translations and rotations occurred about this initial pose. Gait kinematics were applied at an 8-fold reduction of *in vivo* velocity for 10 cycles of preconditioning, [7] followed by another 10 cycles of data collection. A multi-axis load cell (ATI; Theta Transducer) recorded forces and torques transmitted across the joint while the Vicon system recorded the positions of the reflective markers.

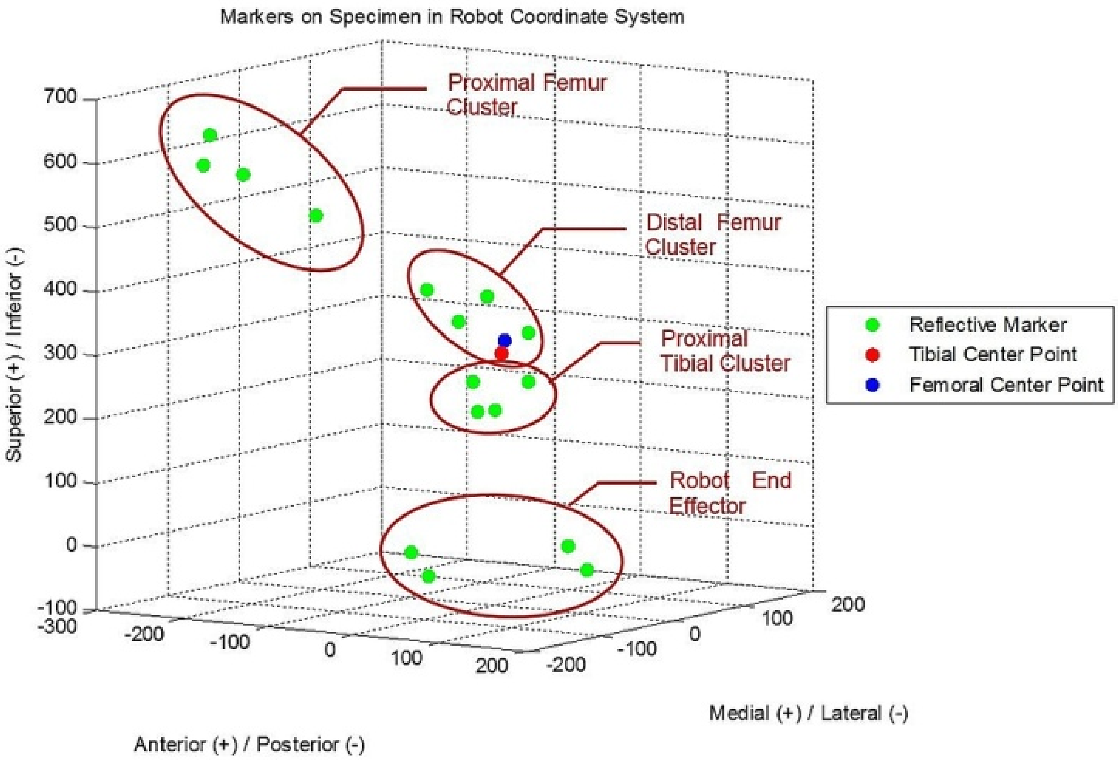

Figure 2. Reflective marker positions as measured in the robotic coordinate system using the CMM. Tibial and femoral center points are examples of bony landmarks used to define the tibial and femoral coordinate systems. Absolute distances from bony landmarks to each of the markers in the closest cluster were used to place landmarks within the marker coordinate system. This depiction is vertically inverted compared to Figure 1

2.2. Data Analysis

Forces and torques were filtered using a Fourier transfer to eliminate high frequency noise and were normalized to body weight before being averaged into a single cycle. Marker data was labeled within the Vicon iQ software workspace and exported to spreadsheets. A custom MATLAB script cleaned the data by unifying trajectories for markers of the same label and eliminating dual instances caused by artifact. A moving average filter applied to each DOF of data extracted noise above 20 Hz. All 10 test cycles were then averaged together to establish overall marker trajectories for a gait cycle. Test cycles missing full or partial spans of data due to "lost" markers were eliminated from the overall calculation. Standard deviations in cycle trajectory averaged approximately 0.3 mm.

Marker data was then used to generate independently recorded anatomic kinematics at the knee joint itself. Because of the setup alignment of the tibia with the robotic end effector, the robotic coordinate system and the tibial aspect of the joint coordinate system (JCS) are identical in unloaded positions while the femoral aspect of the JCS remains stationary throughout testing (Figure 3). Anatomic kinematics reported by the robotic system assume that the tibial and femoral bones are rigid and that offsets from each bone's JCS to the known or stationary fixture positions are constant throughout testing. However, even with short distances from the joint to each of the fixtures, the bones may be able to flex with respect to the fixtures in relatively high load conditions, which could cause deviation from the reported kinematics. Therefore, markers were clustered as close to the tibial and femoral JCS landmarks as possible to minimize relative motion between them, allowing the camera system to independently assess the position of the actual JCS during testing.

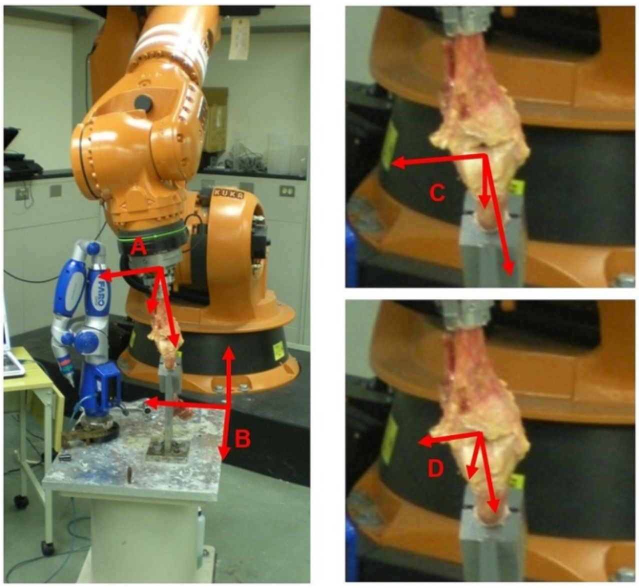

Figure 3. Robotic test setup including each of the coordinate systems referenced in this study: A) Robotic coordinate system, B) Marker coordinate system, C) Femoral coordinate system, D) Tibial coordinate system

Marker trajectories were used to reconstruct actual positions of the tibial and femoral coordinate systems of the knee joint (Figure 3.C. and 3.D.), using CMM data to place joint landmarks within the marker reference frame (Figure 3.B.). All landmark and marker sphere positions were recorded with respect to the robotic coordinate system during specimen set up. Absolute distances of a landmark from each of its 4 closest markers were calculated, allowing placement into the marker coordinate system by solving a system of equations. This placement was iterated for every landmark at every frame of the gait cycle, resulting in a record of independently measured tibial and femoral positions. Coordinate transformations were used to calculate anatomic kinematics as the motion of the tibia about the femur. This data was then compared to the target kinematics.

Kinematic error was compared to the corresponding forces and torques recorded during the motion. For each specimen, correlation coefficients were computed between the error in each DOF and the forces and torques being applied at the corresponding time point. For a sample size of 6, the critical Pearson correlation is 0.811 for an alpha value of 0.05. If, at any point during the simulation, the determinant of the tibial or femoral position matrix fell below 0.994, this data was excluded from the correlation coefficient calculations. As matrix determinants are measures of orthogonality, the further it is from 1.000 (perfectly orthogonal), the more numerical instabilities and inaccuracies are introduced into positional calculations. For reference, determinants of specimen coordinate systems in our current studies must be above 0.997 in unloaded conditions to ensure that target positions are accurately achieved within ± 1 degree.

2.3. Eliminated Data

The three specimens were verified to be orthogonal by the CMM during setup at a zero-load condition. The average femoral orthogonality of the remaining three specimens was 0.998 ± 0.002, and average tibial orthogonality was 0.989 ± 0.015. Instances do occur where we are unable to derive orthogonal bony coordinate systems from a given specimen's recorded marker data. The calculated coordinate systems from such specimens are never orthogonal enough to achieve numerical stability and prevent mathematically accurate calculations kinematics. Accordingly, any tested specimens that fell within this category were excluded from the present investigation.

3. Results

Using only orthogonal data across the gait cycles of each of the 3 numerically stable specimens, the average anterior/posterior kinematic error of the tibia with respect to the femur was calculated at 8.1 ± 13.6 mm posterior; medial/lateral error at 6.9 ± 7.8 mm medial; superior/inferior error at 6.3 ± 9.6 mm superior; internal/external error at 3.0 ± 6.9° external; flexion/extension error at 5.1 ± 5.0° flexed; and ad/abduction error at 6.8 ± 3.9° abducted compared to the input kinematics (Figure 4). Highest deviations from the target kinematics occurred during the swing phase of gait, when knee loads were at their lowest (Figure 5). Correlation coefficients of the same subgroup were largest for medial deviation and medial force (0.68), medial deviation and distractive force (-0.73), and medial deviation and flexion torque (0.73, Table 1).

Table 1. Correlation matrix that expresses the correlation coefficients between respective loads (columns) and positional deviations (rows) across all six DOFs for the joint and robotic manipulator. (A/P = anterior/posterior translation, M/L = medial/lateral translation, C/D = compression/distraction translation, I/E = internal/external rotation, F/E = flexion/extension rotation, A/A = abduction/adduction rotation)

	A/P	M/L	C/D	I/E	F/E	A/A
A/P	0.45	-0.04	0.18	-0.23	-0.16	-0.40
M/L	0.04	0.68	-0.73	-0.49	0.73	-0.30
C/D	0.15	0.52	-0.60	-0.32	0.57	-0.22
I/E	-0.01	0.22	-0.23	-0.19	0.21	0.08
F/E	-0.14	-0.39	0.27	0.46	-0.29	0.42
A/A	-0.88	-0.27	0.29	0.27	-0.29	0.17

However, considering only one specimen with the most orthogonal data, average anterior/posterior kinematic error was calculated at only 2.1 ± 0.6 mm anterior; medial/lateral error at 1.6 ± 1.1 mm medial; superior/inferior error at 0.8 ± 0.4 mm inferior; internal/external error at 0.1 ± 9.6° external; flexion/extension error at 1.0 ± 4.2° flexed; and ad/abduction error at 3.0 ± 1.0° abducted compared to the input kinematics. Correlation coefficients with the corresponding load data were much lower, with a maximum coefficient of only 0.47 for anterior deviation compared with compressive load.

All correlation coefficients were less than the critical value of .811.

4. Discussion

The present investigation examined the effects of system stiffness relative to tibiofemoral joint kinematics during an *in vitro* gait simulation. It was hypothesized that a) the kinematic errors generated from the present robot model would be smaller in magnitude than previously reported kinematic errors associated with *in vivo* motion analysis , and that b) the magnitudes of these kinematic errors would correlate to magnitudes of joint loading for any given time point.

The present data partially supported the hypotheses. Mean errors from skin-based markers throughout a gait cycle were 9.0 mm anterior/posterior, 6.2 mm medial/lateral, 4.4 mm superior/inferior, 2.5° internal/external, 2.6° flexion/ extension, and 3.3° ad/abduction. [16] These magnitudes generally increased when evaluated during a more dynamic task such as athletic cutting. [15, 16] Error values from the most orthogonal subject in the present study were smaller in magnitude than these previously reported bone pin versus skin marker errors throughout the gait cycle. However, the average magnitude of errors across all three subjects were greater than those previously reported from *in vivo* captures.

During the present study, the most orthogonal specimen created a simulation that maintained greater accuracy than the current accepted standard for *in vivo* 3D motion analysis, while less-orthogonal systems did not.

Although verified calibration and calculation techniques were used to complete and analyze this study, [7, 8, 12] only half of the specimens tested generated usable results as defined in the methodology. This was due to errors in orthogonality of mathematically reconstructed bony coordinate systems during gait simulation, leaving true anatomic position impossible to accurately calculate. These errors may have been generated in two ways: 1) The method of calculating and placing anatomical landmarks into the marker coordinate system was not accurate enough to maintain orthogonality. This includes possible marker occlusion due to camera placement and the introduction of error by prodding markers with the CMM tip, itself; 2) While CMM data confirmed orthogonality during limb setup at zero load, bone bending caused landmark positions to shift relative to one another during testing, leading to physical orthogonality disparities.

With residuals of almost 1 mm in magnitude in the Vicon system, the inherent system error may have easily been high enough to promote inconsistencies and calculation error when solving for landmark positions. Unfortunately, these landmark position calculations could not be avoided in the current study, as the working envelope of the CMM prevented the marker and robotic coordinate systems from being dually measured by the same equipment. Accordingly, it is possible that these error sources would prevent our calculations from accurately placing marker positions within the system despite the bony constructs maintaining an anatomic orthogonality.

On the other hand, bone deformation could be the culprit if to relative motion between markers. Small perturbations in the motion path of robotically simulated joint articulations can create significant loading differences. [25, 26] On a series robot, a 1.0 mm shift in the applied compression position altered the corresponding joint compression forces by 359.4 ± 8.5 N during the stance phase of gait. [26] Similarly, on a parallel robot system, perturbations of 0.5 mm altered ligament loading by between 34% and 110%. [25] However, orthogonality errors in the present investigation were not accompanied by these drastic alterations in force or torque at the joint. For this reason, a mathematical or marker/camera placement error is more probable than a physical mal-alignment due to significant bone bending.

Even with computational challenges, a subset of data was successfully presented which compares actual kinematic output with target values during simulations of anatomic motion. Results showed that kinematic deviations were not significantly linked with the amount of load experienced at the knee joint (all correlation coefficients <.811). In fact, most deviation occurred during flexed periods of gait when loads were low – namely mid stance and swing. These errors may be explained by the flexion of the specimen interfering with the line of sight between one or more markers and the more posteriorly positioned cameras. This outcome suggests that end effector deflection truly is negligible compared to other accepted sources of error associated with *in vivo* motion analysis.

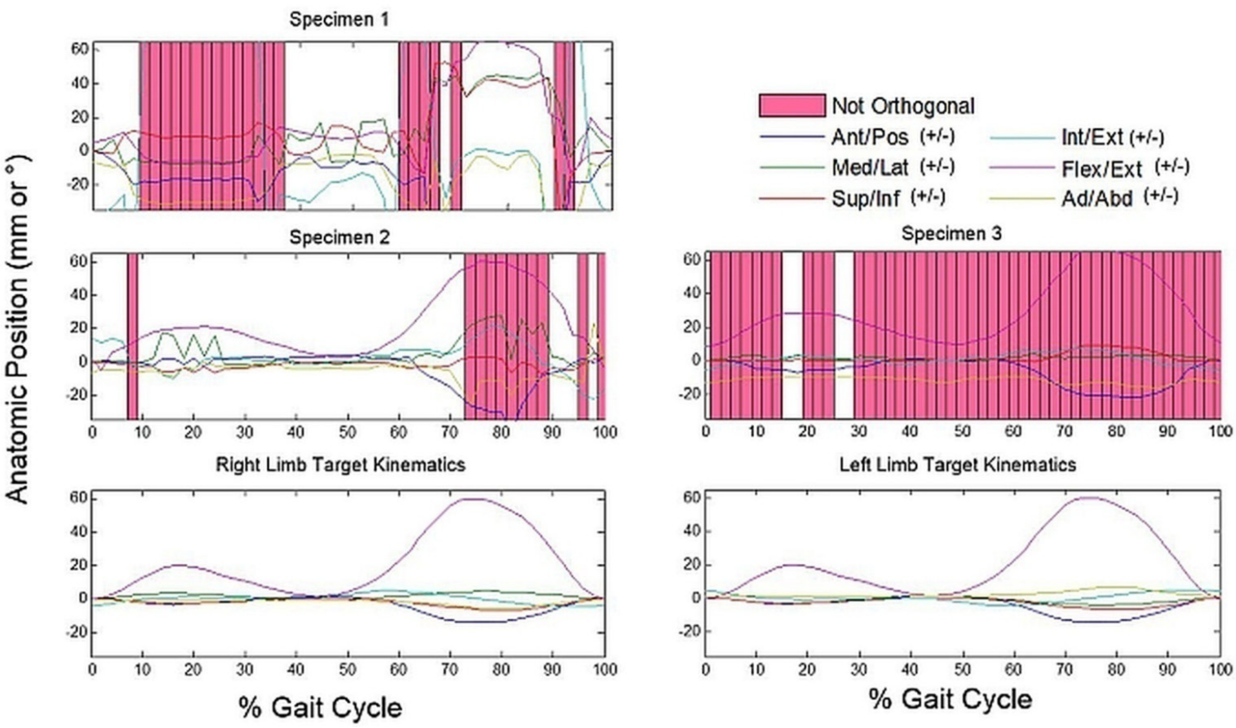

Figure 4. Comparisons of calculated and target anatomic positions of the tibial coordinate system with respect to the femoral coordinate system in each of the 3 specimens resulting in orthogonal bony coordinate systems for at least one time point during gait. Deviations from target kinematics were greatest during periods of increased flexion

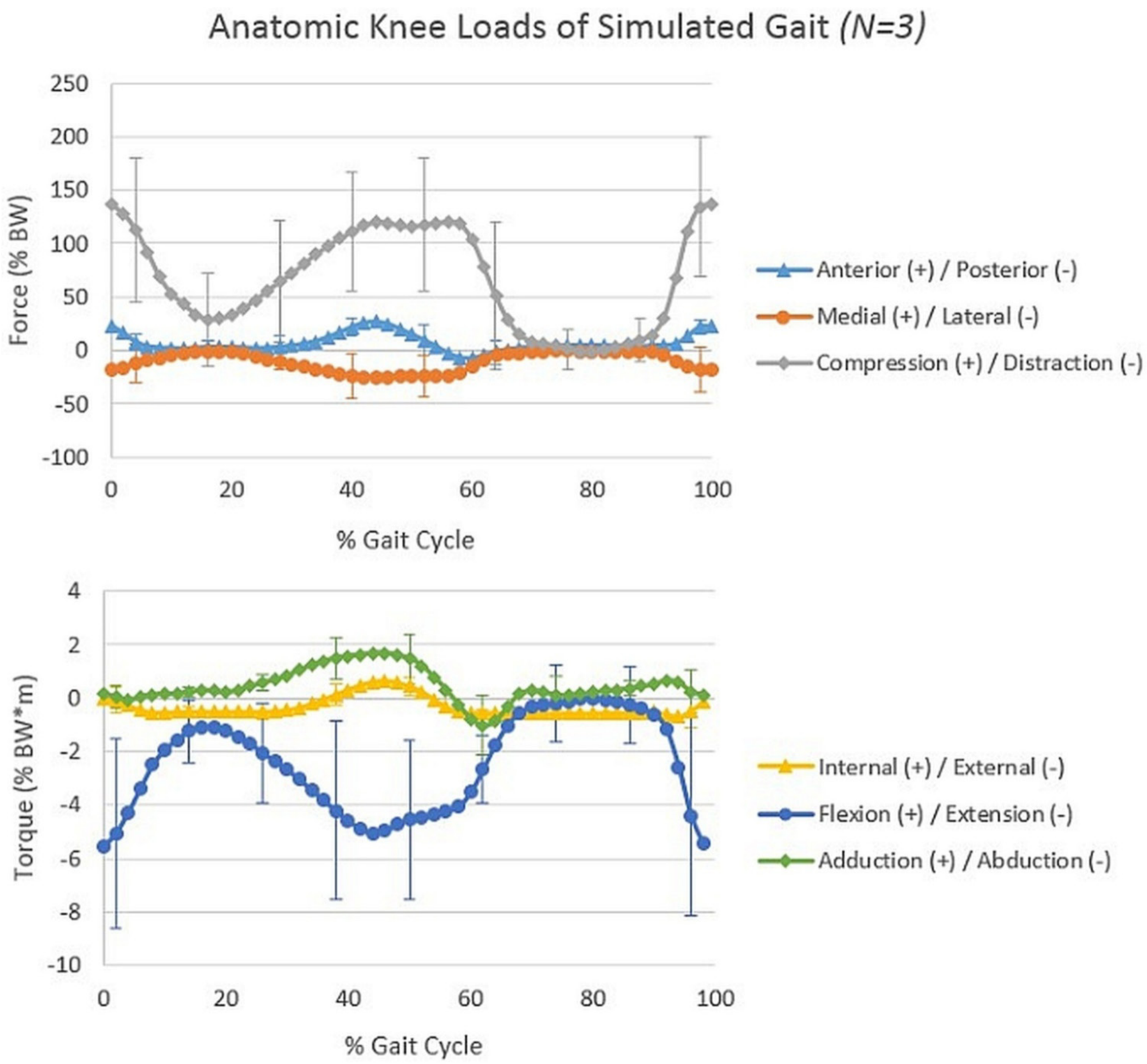

Figure 5. Average dynamic knee forces and torques of the 3 orthogonal specimens, normalized to body weight, throughout a simulated gait cycle with standard deviation error bars. Heel strike corresponds to 0% and 100% gait. Toe-off occurs at 64% of gait. Heel strike and push-off demonstrate the highest loading values during gait

5. Conclusions

The results indicate that kinematics of simulated activities producing high loads at the joint are *not* at an increased risk to be influenced by bone bending and/or robotic compliance. Periods of high-loading did not correspond to decreased orthogonality, nor did they correlate to increased kinematic deviation. However, future studies employing higher resolution equipment are necessary to confirm these findings.

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