INTRODUCTION: The quantification of total knee replacement (TKR) kinematics is an essential part of the design and evaluation process for any implant. *In vivo* kinematics however are not easily predicted from implant design alone. To accurately assess these kinematics, the researcher must address the combined effects of the implant’s design with both the surgical and patient factors; including intra-operative alignment, soft tissue constraint and patient activity. Too often, TKR kinematic measurement is reserved for the *in vivo* arena; presenting the researcher with many experimental data collection challenges and very little access to the implant itself. The large deviations reported in this *in vivo* kinematic data are often attributed to the inability to control or quantify the numerous surgical and patient factors, making it difficult to apply these results towards the advancement of the implant’s design.

With recent advances in laboratory simulator design, the experimental *in vitro* assessment of TKR kinematics has begun to provide the TKR designer with accessible, reproducible and timely information that can readily advance the design of the final implant. These *in vitro* tests must consider the same surgical and patient factors that make *in vivo* testing so indeterminate. The critical difference is that *in vitro* testing can simulate and then quantify these variables; making them powerful tools in the assessment of simulated *in vivo* TKR performance. The current hypothesis asserts that it is now possible to simulate complex *in vivo* kinematics using force-controlled testing for the evaluation of new and current TKR designs. The current study utilizes a multi-axis, multi-station TKR wear simulator to quantify the functional kinematic laxities of eight different TKR designs during a standardized force-controlled walking cycle.

METHODS: The eight TKR designs tested were chosen to include a wide range of suspected anterior/posterior (AP) and internal/external (IE) laxity. They were as follows: Zimmer Mobile Bearing Knee, Mark III (MBK); Howmedica Pinnacle (HOWD); Johnson & Johnson PFC (JPJC); Sulzer Protek Guepar (SPRO), Zimmer Nexgen Cruciate Retaining (NGCR), Zimmer Insall/Burstein II (IB), Zimmer NexGen Legacy posterior stabilized (NGL), and Stanmore SMILES rotating hinged knee (SML). All implants were made of average manufacturer’s sizes (tibial AP 42-47mm, tibia ML 71-77mm, femoral ML 66-72mm). The average femoral flexion axis between 0 and 90º was first determined as a global kinematic reference. All TKRs were installed on the kinematic knee simulator, insuring that the relative alignment of the tibia with respect to the femur mimicked that of the recommended surgical implantation. The zero or kinematic “home position” of the tibia with respect to the femur corresponded to the location of the tibia at 0º flexion with no AP load or torque and a 50N axial coupling load. For all studies, passive linear spring soft tissue constraints of 20N/mm for AP displacements and 0.27Nmm/deg. for rotation were incorporated to approximate the passive linear spring soft tissue constraints of 20N/mm for AP displacements with no AP load or torque and a 50N axial coupling load. For all studies, flexion (de g.) was used as a lubricant.

RESULTS: Figure 2 shows the average AP and IE walking cycle kinematics for all TKR designs tested. The data is separated into two groups of four implants for clarity. The data collected was very reproducible, with standard deviations at any point in the gait cycle averaging less than 0.24 mm and 0.56 degrees for any one implant. Maximum posterior displacements and internal rotations were consistently recorded during the stance phase, at about 60% of the gait cycle. This was just before toe-off, when the axial load was returning to zero, and small AP and IE loads encountered the least frictional resistance. Most implants reversed direction many times during a complete gait cycle, indicating significant sliding of the bearing surface. The MBK exhibited decreased motions, and showed an increased tendency to stick and slip with changes in load direction. External rotation was observed during heel strike, but only for the less conforming tibial designs.

DISCUSSION: Unlike *in vivo* testing, *in vitro* simulation provides a valuable standardized platform from which the evaluation of implant design can progress. With *in vivo* testing, it is often not possible to achieve a consensus on the performance of even a single design’s kinematics, let alone compare differences between designs. The results of this study show that the simulation of *in vivo* activity has advanced to the stage of providing reproducible kinematic performance data for a wide range of implant designs. From this data, significant differences between the mechanical performance of different designs can be determined. These simulations can also incorporate and study the effects of numerous patient and surgical variables that until now were hindrances to the study of implant design *in vivo*. The force-controlled simulation of knee activity allows the design of the implant to be the critical factor in testing results; allowing the joint to displace and rotate to the degree that the design intends. As with any *in vitro* investigation, the results are limited by the ability of the experiment to reproduce the *in vivo* condition. Future studies should concentrate on extrapolating these kinematic results to *in vivo* findings. The results obtained in this study are a first step toward investigating how these measured kinematics differences affect other mechanical factors of the joint; namely joint load distribution, polyethylene wear and load transfer to the surrounding soft tissue structures. These are topics for future research.


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