Micromotion of Simulated Daily Activities using Implant-Specific Kinematics from in vivo Measurements

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INTRODUCTION: Unicompartmental Knee Arthroplasty (UKA) is an alternative to total knee arthroplasty (TKA) for patients with single-compartment osteoarthritis (OA) [1]. Medial OA is the most common clinical problem, with an incidence about ten times higher than the lateral compartment [2]. Aseptic loosening is one of the most common reasons for UKA failure [3]. The initial fixation of cementless UKA tibial trays is critical as micromotions exceeding 150 µm prevent bone formation [4]. Previous experimental simulations focused on tray fixation have not recreated the physiological loading environment of UKA patients during activities of daily living. Steklov et al. tested UKA using constant loads in the medial and anterior directions [5], while Yildrim et al. simultaneously tested medial and lateral UKA to balance the joint loads during loading conditions published for TKA in the Orthoload database [6]. This *in vivo* data was obtained with ultra-congruent TKA inserts that cause large articular forces due to their congruent design [7], instead of data measured with flatter UKA implants. In a recent study, Zumbrunn et al. published tibiofemoral (TF) kinematics measured with dynamic fluoroscopy for a modern UKA design in a large cohort of subjects during multiple activities of daily living, including gait (GT), deep knee bending (DKB) and stair descent (SD) [8]. The purpose of this study was to generate physiological experimental simulations using UKA-specific kinematics, to eventually evaluate tray-bone micromotion.

METHODS: Previously published *in vivo* fluoroscopic knee kinematics including knee flexion-extension (F-E), internal-external (I-E) rotation, medio-lateral (M-L), and anterior-posterior (A-P) translations from 16 subjects were analyzed to determine target implant kinematics [8]. The patients were implanted with SIGMA® HP UKA (DePuy Synthes, Inc) for medial OA. Each subject performed multiple cycles of GT, DKB and SD. The kinematics for each subject's second cycle was isolated and averaged across subjects. A VIVO simulator (AMTI) was modified to mount a fixtured mid-sized UKA (Size 3 femur, Size 2 tray, Fig. 1). The average *in vivo* kinematics were used to generate implant-specific boundary conditions, including synchronized displacement profiles (A-P, M-L translations, F-E, A-A, I-E rotations), and loading profiles (S-I compression). TF compressive loading was kept constant at 500N to keep the implants in contact through flexion. Knee kinematics were measured with an optical tracking system (OPTOTRAK, NDI, Waterloo, ON). The flexion profiles during DKB and SD were scaled to accommodate the flexion limits of the simulator (Fig. 3). Root mean square errors (RMSE) and Pearson's coefficients were evaluated between the AP low point translations measured in the experiment and those measured *in vivo* (Fig. 2).

RESULTS: RMSE between the mean *in vivo* low point kinematics and the simulation ranged from 0.5mm (SD) to 3.8mm (swing phase of GT, Fig. 2). The experimental set-up most accurately recreated the *in vivo* knee kinematics during the stance phases of GT (RMSE = 2.5mm) and SD (RMSE = 0.5mm). Pearson's analysis revealed that the coefficients ranged from ρ = -0.61 during the extension phase of DKB (poor correlation) to 0.98 during the stance phase of GT (strong correlation). All the measured coefficients were reported in Fig.2D.

DISCUSSION: In this study, experimental evaluation of UKA kinematics was verified against *in vivo* fluoroscopic kinematics. One limitation of this preliminary simulation was the use of a single implant size for both the femur and tibia, rather than matching each patient's implant size. Variations between the kinematic profiles are to be expected and are likely caused by differences in implant alignment (i.e., tibial slope) and compliance in the implant fixtures. Furthermore, the differences observed in the low point kinematics between the experimental set-up and fluoroscopy may be caused by registration error of the tray's positioning in A-P directions. Another limitation is the use of a constant S-I load. Future simulations should consider the use of TKA S-I loading profiles that have been modified with respect to condylar balance. Nevertheless, our findings suggest that simulated daily activities can be recreated from *in vivo* kinematics, and they can be used to perform future implant micro-motion analyses, either in artificial or cadaveric bone.

SIGNIFICANCE/CLINICAL RELEVANCE: This study demonstrated a combined *in vivo* measurement and experimental methodology to generate physiological loading conditions for pre-clinical evaluation of UKA.

REFERENCES: 1. Jennings JM et al., 2019, 2. Fiocchi et al., 2017, 3. Tai et al., 2021, 4. Jasty et al., 1997, 5. Steklov et al. 2010, 6. Yildrim et al., 2016, 7. Bergmann et al., 2014, 8. Zumbrunn et al., 2019.

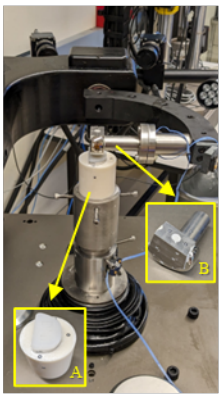

Fig.1 – Experimental Set-up. Size 3 femoral medial implant (A) size 2 tibial medial tray (B).

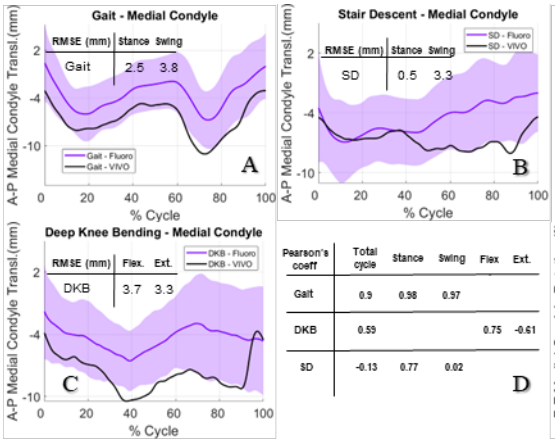

Fig.2 – Comparison between target *in vivo* (purple) and experimental (black) low point A-P translation for the three activities (A,B,C). Pearson's coefficients for total cycle and different phases of the three activities (D).

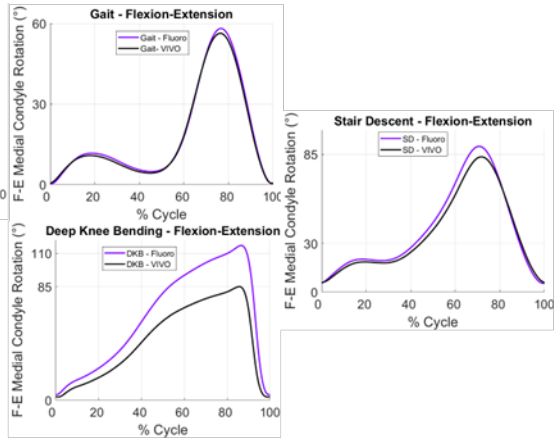

Fig.3 – Comparison between target *in vivo* (purple) and experimental (black) F-E rotation for the three activities.