ASSESSING THE VARIABILITY OF HIP LOADS IN TOTAL HIP REPLACEMENT PATIENTS

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INTRODUCTION

Preclinical wear testing of implants is currently performed using a stylized waveform which resembles loads during gait but does not consider the load variability experienced in other activities of daily living (ADLs) [1]. Additionally, testing standards consider total hip replacement (THR) patients as a homogenous group, however, previous large cohort studies have identified large variability in their kinematics [2]. The wide-spread failure of the ASR hip implant highlighted the potential importance of patient characteristics [3]. Differences in implant loadings could partially explain the large variability in implant survival rates across THR patients, which current testing standards fail to predict. While data from instrumented implants [1] only represent a small subset of the overall population, the use of motion capture analysis and musculoskeletal modelling can overcome this and enable the analysis of large cohorts of patients. This study aimed to identify differences in hip contact forces (HCF) in a large cohort of THR patients stratified by patient characteristics and while performing different ADLs.

METHODS

132 THR patients, >12 months post-surgery, underwent 3D kinematic (Vicon, UK) and kinetic (AMTI, USA) analysis whilst performing different ADLs, among which were level gait, fast gait, stair ascent and descent, sit down and stand up from a chair, squats and lunges. 2148 trials were processed and analysed through multibody modeling (AnyBody Technology, Denmark). A recent generic lower-limb musculoskeletal model [4] was scaled to match the anthropometrics of each patient, based on marker data collected during a static trial. The model, previously validated against HCF from instrumented implants during gait, presents refined muscle geometries, which were verified against cadaveric MRI scans, and realistic muscle lever arms over a large range of hip angles, suggesting its suitability for modelling other ADLs. Average HCFs for each patient were computed during each ADL. Mean predicted HCFs were qualitatively compared to measurements from instrumented implants and across activities. Force contact pathways at the bearing surface were calculated as the intersection points of the force vector with an acetabular 32-mm cup fixed on the pelvis with a standardized 40° inclination and 15° anteversion. Mean contact pathways were calculated across all patients for each activity.
To explore the effect of patient-specific characteristics on the variability of the data patients were additionally stratified into three BMI groups, five age groups, and three functional groups as determined by their self-selected gait speed. By means of statistical parametric mapping (SPM), statistical analyses of the 1-dimensional time series were performed to separately evaluate the influence of age, BMI and functionality on HCF during gait [5].

RESULTS AND DISCUSSION

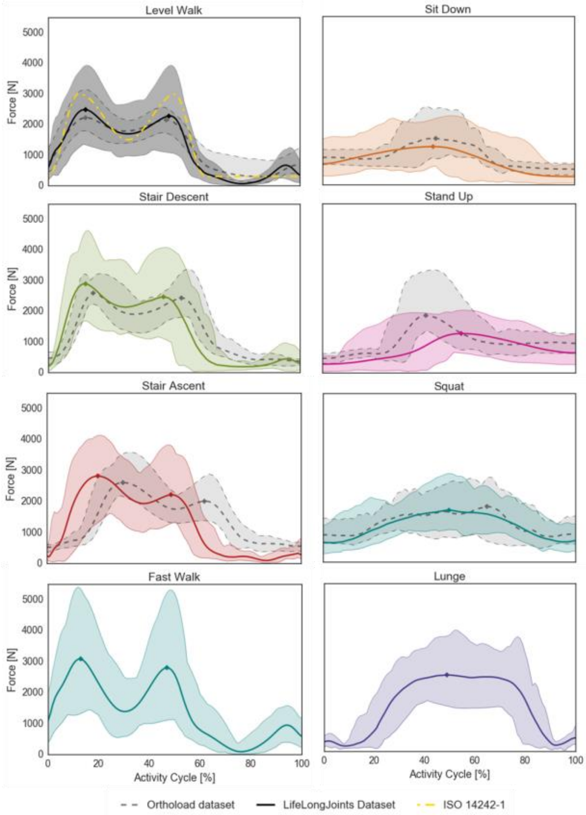

Fig 1: Predicted mean HCF, with min/max range of variation, across 132 patients for different ADLs, compared to data from instrumented implants [1] and current preclinical testing standards.

Predicted mean HCF showed similar trends and values with data from instrumented implants for all compared activities, further supporting the validity of the models' predictions (Fig 1). Additionally, the predicted ranges of variation were generally wider, as might be expected from a larger dataset.

The analysis of the resultant HCF revealed qualitative differences between the waveform profiles of the different ADLs. Fast walking is characterized by higher peak HCF values, compared to walking and a lower force during mid-stance, whilst stair ascent and descent present higher or comparable HCF throughout their loaded phase compared to level walking. Stair descent exhibited a prolonged posterior load, which was not present in other locomotive tasks. The larger kinematic variability of the non-locomotive tasks translated in more evident waveform differences in HCF, with lunges presenting the highest loads.

The analysis of the contact paths (Fig 2) showed different topological distributions of the load on the bearing surface depending on the type of activity. While walking presented a typical figure-8 pattern with peak loads in the anterior-superior quarter of the cup, stair ascent presented a contact path more circular in shape with peak forces occurring on the cup superior-posterior portion. The non-locomotive activities such as stand-up, sit down and squat task had a linear shape, spanning across the cup superior-posterior quarter.

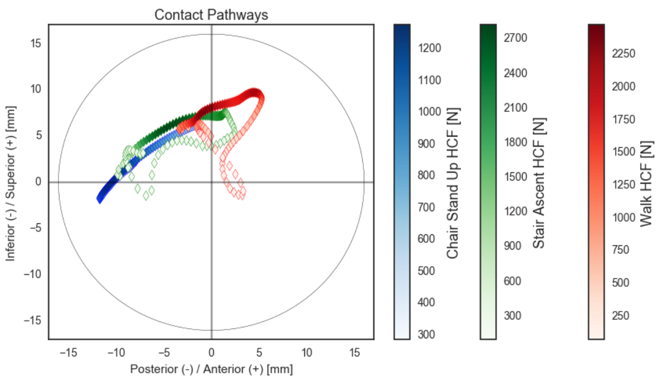

Fig 2: Mean resultant hip contact force pathways for gait, stair ascent and standing up from a chair.

Lastly, the SPM analysis of the patients' characteristics highlighted a statistically significant positive linear correlation between BMI and HCF, indicating that obese patients are more likely to experience higher HCF during most of the stance phase, while a statistically significant negative correlation with age was found only during the late swing-phase. Patients with higher functional ability (self-selected walking speed >1SD above whole sample mean) exhibited significantly increased peak contact forces, while patients with lower functional ability (>1SD below whole sample mean) demonstrated lower HCF overall and a pathological flattening of the typical double hump force profile (Fig 3) [5].

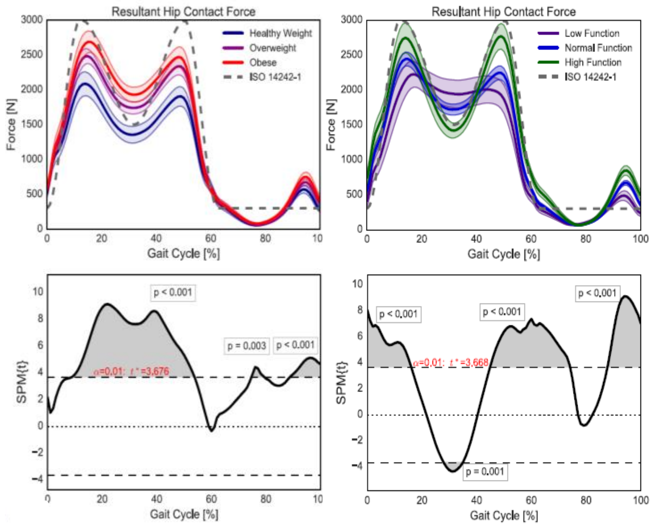

Fig 3: Mean HCF with relative 95% CI during a gait cycle, stratified by BMI and functioning level. SPM curve analysis showed significant difference in the supra-threshold clusters [5].

CONCLUSIONS

This study has been the first to explore HCF during different ADLs in a large cohort of THR patients. We identified differences in the overall loading patterns at the hip during different ADLs. Differences observed in contact path directionality and in the location of peak loads could be critical in determining realistic wear rates. Additionally, HCFs experienced at the bearing surface are highly dependent on patient characteristics. BMI and functional ability were determined to have the biggest influence on contact forces. This variance is not taken into account in current preclinical testing standards, such as ISO 14242-1, which could partially explain the heterogeneity in implant survival rates. Our results suggest that including ADLs and differing patient characteristics in preclinical testing would provide more accurate prediction of implant performance.

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ACKNOWLEDGEMENTS

Funding from the European Union's Seventh Framework Program (FP7/2007-2013), grant agreement GA-310477 (Life-Long Joints).