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Hip contact force pathways in total hip replacement differ between patients and activities of daily living

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Abstract

One of the main causes of implant failure and revision surgery in total hip replacement (THR) is aseptic loosening often caused by the accumulation of wear debris arising between the contact surfaces of the acetabular cup and femoral head during activities of daily living (ADL's). However, limited information is available regarding the contact force pathways between these two surfaces during specific ADL's. In this study, through musculoskeletal modelling, we aimed to estimate the orientation of the hip contact force pathway on the acetabular cup. One hundred and thirty-two THR patients underwent motion capture analysis whilst undertaking locomotor and non-locomotor ADL's. Musculoskeletal simulations were performed to calculate contact force pathways using inverse dynamics analysis. We then qualitatively compared differences in the contact force pathways between patients and between ADL's. Walking resulted in a typical figure-of-eight pattern, with the peak contact forces occurring in the superior-anterior area of the cup. The non-locomotive activities such as stand up, sit down and squat had a more linear shape, spanning across the superior-posterior quarter of the cup. Our results showed a large inter-patient variability in the shape and location of the contact force pathway.

There is a distinct difference in the location and shape of the pathway between locomotor and non-locomotor activities and this could result in different wear accumulations. These results could enhance our understanding why revision rates vary across the population and could inform the development of personalised implant design.

Keywords: Total hip replacement, gait, musculoskeletal modelling, contact force pathways, acetabular cup, femoral head

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1. Introduction

Total hip replacement (THR) is a highly successful orthopaedic procedure with a long history of implant development and refinement with well-established surgical techniques (Knight et al., 2011 ; Learmonth et al., 2007 ; Pivec et al., 2012). However, when an implant fails revision surgery can be required. Revision surgeries are usually associated with worse functional outcomes than primary THR and a higher risk of complications, such as hip dislocation and infection, which translate into a higher financial burden for healthcare providers (Mahomed et al., 2003)

The main cause of implant failure and revision surgery is aseptic loosening (Bozic et al., 2009 ; Fisher et al., 2012 ; Sadoghi et al., 2013 ; Wroblewski et al., 2007 ; Zietz et al., 2015) often due to the accumulation of wear debris at the implantation site (Ingham and Fisher, 2000, 2005 ; Schmalzried et al., 1992). Edge-loading has been associated with high wear rates of polyethylene liners (Fisher, 2012 ; Patil et al., 2003) leading to increased contact stresses and plastic deformation of the polyethylene liner, which would eventually lead to wear of the component (Hua et al., 2014). In metal-on-metal (MoM) bearings wear, as a result of edge loading, can lead to metal ion release leading to failures due to biological reactions such as metallosis (Hart et al., 2013 ; Leslie et al., 2009). Whilst the mechanism of wear accumulation can be different between material component type, the development of wear in all THR implants presents a strong correlation with individual patient characteristics (Schmalzried and Huk, 2004) and particularly with their activity level, leading to the general understanding that implant wear develops as a consequence of use, rather than time (Schmalzried et al., 2000). Research suggests that individual motion patterns, in combination with implant positioning, play a significant role in joint contact forces (Foucher et al., 2009) and wear rates (Ardestani et al., 2017 ; Mellon et al., 2013) . From a mechanical perspective, the development of wear has been described as a function of the local stresses on the bearing surface, the relative sliding distance between femoral head and cup, the roughness of the surfaces, and material-specific mechanical properties (Di Puccio and Mattei, 2015 ; Gao et al., 2018 ; Li, 2017). In addition to individual motion patterns, inaccurate implant positioning in THR can cause edge-loading, as well as micro-separation of the femoral head relative to the cup, which are known to lead to higher wear rates and earlier failure of the implant (Fisher, 2012 ; Zietz et al., 2015) .Edge-loading results in an increase in the mechanical stresses at the rim of the bearing surface, either of acetabular cup or the liner (Zietz et al., 2015). Using musculoskeletal modelling, Mellon et al. (Mellon et al., 2013) analysed the distance between the contact force and the rim of the bearing surface as an indication of edge-loading. Understanding the individual motion patterns and the location of peak contact force between the bearing surfaces is important to understand how these variables may influence wear.

To predict and improve the wear performance of THR implants it is necessary to understand the realistic implant conditions, *in vivo*, including both kinematics and loading. (Fisher, 2012 ; Medley, 2016). Hip joint kinematics can be determined through conventional gait analysis and once known, it is possible to compute the movement "pathways" (also referred to as "paths" or "loci") that selected points of the femoral head surface would mark on the cup surface during a given activity. The length of these paths can then be calculated as the sliding distance of the femoral head relative to the acetabular cup (Bennett et al., 2008a ; Bennett et al., 2008b ; Bennett et al., 2002 ; Budenberg et al., 2012 ; Davey et al., 2005). Direction of these motion paths between the acetabular cup and femoral head have been shown to be critical in determining the wear behaviour of polyethylene material. Linear pathways exhibit surface strain hardening and non-linear pathways (multidirectional motions) cause strain softening and subsequently increase wear (Davey et al., 2005 ; Turell et al., 2003). Previous analysis of contact pathways at the implant bearing surface showed that patient-specific sliding distances are positively correlated with *in-vivo* wear rates, with higher wear rates also associated with more multi-directional motions that yield sliding pathways with lower aspect ratios (Bennett et al., 2002). Clear differences have also emerged between the contact pathways produced in current preclinical testing standards and those observed in patients during different ADL's (Fabry et al., 2013). Weber et al. (Weber et al., 2012) using musculoskeletal modelling, used hip kinematics and joint loading information to compute the contact pathways of the joint force vector on the femoral head surface. This method provides a synthetic spatial representation of relative motion and, crucially, joint loading, and can be used to effectively investigate different patient characteristics and ADLs. Previous studies using this method have analysed a small number of patients (number of patients=2) (De Pieri et al., 2021a) which does not represent the broad population and have only analysed force pathways during walking. This study aims to quantify differences in contact force pathways during different activities of daily living in a large cohort of patients.

2. Methods

The underlying patient data (patient demographics, c3d file data and musculoskeletal models) presented below has been previously presented in two journal articles (Lunn et al., 2019 ; Lunn et al., 2020). In this paper we have applied a previously used contact force path modelling approach (De Pieri et al., 2020) to this data to answer this paper's research question and aims.

2.1. Participants

A total of 132 THR patients were recruited through a clinical database of surgical cases. The inclusion criteria were; between 1-5 years post-THR, 18 years and older, no other orthopaedic or neurological issues which may compromise gait. Ethical approval was obtained through the UK national NHS ethics (IRAS) system and all participants provided written informed consent. Information on patient demographics can be found in Table 1.

Table 1. Patient characteristics presented for each activity of daily living.

	No. Patients	Mean Body Mass (kg)	Mean Height (cm)	Mean BMI	Male/ Female	Mean Age (years)	Years Since Op

Walk	132	78.10(12.79)	166.28 (8.40)	28.20 (3.85)	66/66	71.62(7.61)	2.80(1.42)
Fast	117	78.59(12.81)	167.36 (8.08)	27.99 (3.71)	62/55	70.56(7.31)	2.84(1.43)
Ascent	49	80.13(13.81)	167.55 (9.37)	28.50(4.03)	28/21	69.90(7.70)	3.00 (1.47)
Descent	47	79.87(14.12)	168.01 (9.34)	28.22(3.92)	28/19	70.00(7.87)	3.09 (1.46)
Sit	131	78.08(12.83)	166.25 (8.42)	28.20(3.86)	65/66	71.57(7.61)	2.82 (1.42)
Stand	131	78.08(12.83)	166.25 (8.42)	28.20(3.86)	65/66	71.57(7.61)	2.82 (1.42)
Squat	34	78.45(11.80)	169.74 (6.23)	27.20(3.60)	23/11	67.24(6.28)	3.18 (1.59)
Lunge	35	75.89(11.64)	167.23 (6.41)	27.09(3.53)	22/13	70.29(6.85)	2.57(1.58)

2.2. Data Collection

Participants underwent 3D kinematic and kinetic analysis whilst performing different ADLs, including walking at a self-selected speed, fast walk, stair ascent and descent, sit down and stand up from a chair, squats, and lunges. All patients participated in the walking activity, while the more demanding activities were undertaken based on each patient's ability and confidence. A static trial was captured prior to the collection of the dynamic trials patients, more detail regarding the exact definition of the collected ADL's can be found elsewhere (Lunn et al., 2019). Data was recorded using a 10 camera Vicon system (Vicon MX; Oxford Metrics, UK) sampling at 100 Hz, integrated with two force plates (AMTI, Watertown, MA) capturing at 1000 Hz. Reflective markers were placed on the patients according to the CAST marker set (Cappozzo et al., 1995) to track lower limb segments in six degrees of freedom.

2.3. Data Processing

All markers were labelled and gap-filled using the spline fill function in Vicon Nexus 2.5 (Vicon MX, Oxford Metrics, UK), before the labelled marker coordinates and kinetic data were exported to Visual 3D modelling software (C-motion, USA). Kinematic data were filtered using a low-pass (6 Hz) Butterworth filter. Ground reaction force (GRF) data were filtered using a lowpass Butterworth filter

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(25 Hz). Processed C3D files were then exported for further processing using musculoskeletal modelling.

2.4. Musculoskeletal Modelling

Musculoskeletal simulations were performed using a commercially available software (AnyBody Modelling System, version 7.1, Aalborg, Denmark) (Damsgaard et al., 2006). A detailed musculoskeletal model of the lower limb (De Pieri et al., 2018) based on a cadaveric dataset (Carbone et al., 2015) was scaled to match the anthropometrics of each patient based on marker data collected during a static trial (Lund et al., 2015). Marker trajectories and GRF data from each motion trial served as input to an inverse dynamics analysis, based on a third-order-polynomial muscle recruitment criterion, to calculate muscle forces and hip contact forces (HCF).

2.5. Contact Force Pathways

To estimate the orientation of the HCF on the acetabulum, the HCF vector components were decomposed in a reference frame. The reference frame origin was in the Hip Joint Centre (HJC) and was aligned with the cup opening plane. Due to the lack of x-ray information available in this study a standardized cup position of 40° of inclination and 15° of anteversion was used for all subjects (Lewinnek et al., 1978). Using the musculoskeletal model template, the reference frame was defined by two axes parallel to the cup opening plane, with the superoinferior (SI) axis passing through the mid-acetabular notch creating two equivalent halves of the acetabulum, the anteroposterior (AP) axis perpendicular to the SI axis, and the third axis passing through the centre of the hemisphere and perpendicular to the cup opening plane (De Pieri et al., 2020). Positive x components indicate anteriorly oriented forces, while positive z components indicate superolateral oriented forces.

The HCF vector was then intersected with a hemisphere representing the acetabulum, and the contact force pathways were plotted (De Pieri et al., 2020 ; Weber et al., 2012). Contact force pathways on the bearing surface were calculated for each individual patient during different ADLs as the intersection points of the force vector with a hemisphere representing a 32-mm-diameter acetabular cup fixed on the pelvis.

Average contact force pathways were calculated for each patient and for the mean HCF vector across the whole cohort.

3. Results

Interpretation of the results and graphs presented below would be aided by observing the hip kinematic results, graphs of which can be found in Lunn et al (2020).

3.1. Contact force pathways- Activities of Daily Living

Self-selected walking and fast walking resulted in a typical figure-of-eight pattern, with the peak contact forces occurring in the superior-anterior area of the cup, and the pathways were similar during fast walking. Stair ascent resulted in a longer force pathway, more circular in shape and with peak forces occurring on the superior-posterior portion of the bearing surface. A more clustered contact force pathway, localized in the middle of the superior half of the cup, characterized stair descent (Figure 1). The non-locomotive activities such as stand up, sit down and squat had a more linear shape, spanning across the superior-posterior quarter of the cup and extending into the inferior-posterior quarter (Figure 2).

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Insert Figure 1 here

Insert Figure 2 here

There was inter-patient variability across the cohort, both in terms of pathway shape and location. Figure 3 and 4 indicate the contact locations on the acetabular cup bearing surface of the hip contact forces of individual patients during different activities. In particular, stair descent presented a heterogeneous loading distribution across the cohort, with individual patients' contact pathways presenting different shapes. This heterogeneous contact pathway between patients could be the result of the varying coping strategies exhibited by patients during stair descent (King et al., 2018). The activities characterized by a large sagittal flexion motion, such as squatting, sitting down, and standing up from a chair, presented linearly shaped contact pathways across the cohort; however, some of the individual patients' pathways revealed peak loads occurring in the proximity of the edge of the acetabular cup.

Insert Figure 3 here

Insert Figure 4 here

4. Discussion

This is the first study to explore contact force pathways during different ADLs in a large cohort of THR patients. Our results indicate that various ADLs produce significantly different loading patterns on the acetabular cup, both in magnitude and direction. Peak loads were observed in different locations of the bearing surface, with potentially critical implications for the understanding of wear in hip implants, potential risk for edge loading and subluxation.

The results presented in our study are comparable to previous kinematic analysis of contact patterns during gait, where the same figure-of-eight contact pathways was observed (Budenberg *et al.*, 2012). The distinctive figure-of-eight pathways found in locomotor activities have multi-directional pathways and could lead to more wear due to, previously observed, softening of polyethylene materials during multi-directional articulations (Davey et al., 2005). However, the contact pathways experienced during other ADL's could play a critical role, through other modalities, in determining high wear rates and the subsequent implant mal-functioning and failure. Compared to locomotor activities, non-locomotor activities generated higher forces closer to the edge of the acetabular cup, suggesting an increased risk of edge loading (Mellon et al., 2013), a situation shown previously to disrupt the lubrication of the two components, potentially causing increased wear (Liu et al., 2006).

Previous studies suggested that cup placement and patient-specific kinematics are both important factors determining the wear of an implant and the risk of edge loading and that kinematic and kinetic variability across patients could lead to critical and wear-prone configurations of loading (Ardestani et al., 2017 ; Foucher et al., 2009 ; Mellon et al., 2013). The force vector during locomotor activities exhibited a reduced aspect ratio, as evidenced by the larger contact pathways during walking and fast walking. This is possibly due to the large excursion of the hip during these activities which have been previously found (Lunn et al., 2019). This variance in different activities is not taken into account in current preclinical testing standards, such as ISO 14242-1. However, if wear is a function of use rather than time (Ardestani et al., 2017), then the quantity of various activities undertaken by a patient would directly influence the implant's wear rate. Therefore, varying levels of patient activities could partially explain the heterogeneity in clinically-observed implant survival rates (Bayliss et al., 2017 ; Graves et al., 2004 ; Havelin et al., 2000). A further analysis of retrieved components showed that these changes in material properties are correlated with patient-specific kinematics (Davey et al., 2005). The predicted contact pathways for gait also showed a similar topological distribution to wear patterns from well-functioning, autopsy-retrieved components (Pourzal et al., 2016). Indicating that the contact force pathways have the potential to help predict wear in THR patients and in particular could be used to identify high risk activities for wear and thus early failure. In vivo, wear is a complex tribological phenomenon in which the lubrication of the joint and polyethylene molecule behaviour (Davey et al., 2005 ; Turell et al., 2003) plays a major role, therefore the predicted force pathways cannot alone provide an indication of the potential wear mechanisms nor the exact scar formation. Nevertheless, it represents useful information on the topological distribution of contact stresses, which would apply for both dry and lubricated contact modalities(Mattei et al., 2021) .

A limitation of the current study is the reported results only focused on contact pathways in a standardized cup configuration, in terms of inclination and anteversion angles, following the historical indications of Lewinnek et al.(Lewinnek et al., 1978). Unfortunately, this information was not available to the authors, and it was not possible to incorporate these variables into the analysis. Further analysis of various cup placements using a parametric approach could help identify a safe range of cup placements with respect to wear for different ADL's. This approach could be important to identify which patients are more at risk of edge loading during specific activities, and maybe identify potential compensatory mechanisms which allow the patient to perform the same task while avoiding edge loading and/or feeling of instability (Mellon et al., 2013). Additionally, future research may employ a personalized method to determine specific movement traits of edge loaders and ascertain whether alterations in cup orientation could diminish the risk of edge loading. Employing these techniques before surgery could help identify patients who might be susceptible to edge loading with standardized cup placement. Consequently, during pre-surgical planning, surgeons may consider non-standard cup placement for these patients at risk to mitigate the possibility of edge loading. Similarly, the placement of the femoral component could play a role. Femoral torsion was shown to alter HCFs in the native joint (De Pieri et al., 2021b). Nonetheless, current orthopedic surgical approaches do not take the patient kinematics during ADL's into consideration and typically use static x-rays to determine the cup and femoral component placement. Therefore, the results presented in this study, occluding the patient specific cup placement, provide important information regarding likely differences in contact pathways between ADL's and the patients performing them.

These preliminary findings offer a qualitative indication that hip load distribution varies across different ADLs. Although not quantified in the results, figures 3 and 4 highlight differences in contact force pathways between individuals. This is in agreement with previous literature where patient characteristics affected hip joint contact force(Lunn et al., 2020). Future work on the topic should investigate the identified differences by means of quantifiable metrics, such as sliding distance and aspect ratio of the contact pathways (Bennett et al., 2002 ; Calonius and Saikko, 2002) or the distance of the contact path from the acetabular rim (Mellon et al., 2015) as an indicator of edge-loading risk in large patient cohorts. Despite the increasing evidence of different factors which can influence wear, preclinical implant testing is currently performed using a stylized waveform that represents idealized loads during gait (ISO 14242-4, 2018) and it does not consider the load variability experienced in other activities of daily living nor does it represent the variation we observe between patients (Lunn et al., 2020), thus neglecting the effects that different activities and patient populations might have on wear (Bergmann et al., 2010 ; Fabry et al., 2013). A number of studies have shown that the use of more realistic input loading conditions would result in higher and more clinically relevant wear rates, both in physical simulators (Bowsher et al., 2006 ; Bowsher and Shelton, 2001 ; Hadley et al., 2018 ; Hadley et al., 2013 ; Williams et al., 2006) and computational wear models (Gao et al., 2011 ; Gao et al., 2009 ; Meng et al., 2013) . Wear testing in physical simulators is costly and time-consuming (Mattei et al., 2011), limiting the ability of the industry to extensively investigate a broad spectrum of implant loading scenarios. Implant wear is directly dependent on the local contact stresses and the relative sliding distance between the head and the cup surfaces. More detailed computational analysis of the relative kinematics and loadings at an implant scale in future should help give some indications of the tribological behaviour of the implant.

5. Conclusion

In conclusion, this study demonstrates the variation in contact force pathways in THR patients during activities of daily living. We observed a large difference in the location and shape of the pathway between locomotor and non-locomotor activities. Specifically, locomotor activities are located superiorly and exhibit higher aspect ratios. As suggested by previous literature, if these contact force surface pathways contribute to wear, then the type and frequency of activities performed by the patient will affect the wear rate of the implant. Our method, through computational modelling, provides an opportunity to identify daily activities that pose the greatest risk for edge loading. Moreover, the variability in loading and wear rate observed between activities and patients in our study is not accounted for in current implant testing methods.

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