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Article

Characteristics of hip joint reaction forces during a range of activities

Robin Layton [a], Neil Messenger [b], Todd Stewart [a,*]

[a] Institute of Medical and Biological Engineering, Faculty of Engineering and Physical Sciences, The University of Leeds, Leeds LS2 9JT, United Kingdom
[b] Sport and Exercise Science, Faculty of Biological Sciences, The University of Leeds, Leeds LS2 9JT, United Kingdom

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ABSTRACT

The paper reports the characteristics of joint forces for 9 activities in 18 normal healthy subjects. Activities included Walk, Walk Turn, Stand to Sit, Sit to Stand, Squat, Stand Reach, Kneel Reach, Lunge, and Golf Swing. Within the cohort ~30% variability occurred in the manner in which each activity was completed. Within the activities the average maximum load characteristics varied in magnitude (0.5–6.4 ρBWT) and also in duration (0.96–5.89 s.) when compared to walking (3.1 ρBWT,1.1 s.). The corresponding impulse ranged from 1.6 during the Walk to 6.7 ρ.BWT.s for the Golf Swing . As high loads with low sliding velocities have been shown in the literature to be damaging to the tribology of compliant contact surfaces the findings are postulated by the authors to be specifically important for the pre-clinical testing of cartilage substitutional materials. Note: Force was normalized to body weight (ρBWT) throughout the study.

1. Introduction

Forces generated in the hip are generated primarily from the contributions of muscle forces acting to stabilise the ball-in socket geometry that has no inherent rotational resistance to movement but is broadly supported laterally by the bony structure. It is widely known that despite being a simple ball-in-socket the relative motions of the joint surfaces are very complex and it is the detail of this movement that has been of interest in the past in relation to the durability of joint replacements [1–3]. Polyethylene, for instance is sensitive to local surface movements and shear that can cause molecular re-alignment and increases in wear rates associated with the cross-shear ratio [4–6].

The understanding of shear and even the use of the term is confusing in the literature as it generally refers to the transverse movement between surfaces. Thus all movement in a hip prostheses would be considered as shear as it undergoes multidirectional sliding and it is this sliding or interaction of the surfaces that leads to the generation of wear debris. However, in the field of contact mechanics shear refers more commonly to the strain (deformation) generated within the sub-surface of the material resulting from the application of a surface contact pressure [7]. In a hip replacement the force is applied to the joint surfaces through the requirement for load support statically and joint movement dynamically that in turn causes frictional forces. If we consider static loading alone, underneath the force the surfaces deform by the applied contact pressure (Fig. 1a), and it is this deformation that leads to sub-surface shear within in the material itself (Fig. 1b). This sub-surface shear is generally what leads to structural failure, like the delamination seen in some early knee prostheses that underwent structural fatigue [8]. The authors postulate that this sub-surface shear and thus the magnitude and variability in the hip joint reaction force may be important in the success of cartilage repair [9,10].

In hip replacements the close fitting surfaces produce large contact areas with low contact stresses that are well within the elastic limit of the polyethylene material thus the only long-term concern is wear. History has shown that wear of polyethylene for joint replacements can be simulated in a testing machine that reproduces a smoothed gait motion cycle [11,12] . This testing has revolutionised the development of joint replacements worldwide and the benefit to patients has been shown in increased longevity [13]. However, the actual conditions that occur in our joints are not replicated by this testing as the relative acceleration of the surfaces in a real joint can be several magnitudes greater than applied by the approved testing standards [14]. This raises concern, not for polyethylene bearings where simulation has been very effective, but for other materials/surfaces that may be more sensitive to local tribological conditions.

In healthy joints the cartilage is a tough structure with multi layers of collagen/chondrocytes acting to create a biphasic property that allows the forces applied to it to be supported by the interstitial fluid [15,16]. During sliding there is a complex biological tribology that produces very low friction at the surface thus the surface shear is very small. However,

* Corresponding author.
E-mail address: T.D.Stewart@leeds.ac.uk (T. Stewart).

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the cartilage deforms and this causes high strain/shear internally. With a healthy fluid support cartilage is able to effectively distribute the localised forces applied to it down to the bony structure beneath. When cartilage is damaged locally the potential fluid support can be altered. In this case both the surface(friction) and internal shear (deformation) start to become very important. Cartilage plugs have been in use for many years to repair local defects, however, their clinical performance in relation to the crude methods of debridement /drilling/microfracture and newer methods of gene therapies are debatable [17,18]. As research into larger cartilage substitutional therapies is beginning to gain popularity it is thus very important to introduce clinically relevant testing protocols. This is even more important when considering the cohort of patients receiving these therapies as they are young and active and will be completing more activities than simple walking.

In well lubricated compliant surface contacts, with a similar stiffness to articular cartilage, research has shown that constant loads with reduced movement can cause lubricant starvation between the surfaces [19–21]. This occurs when the deformation in the material and the magnitude and duration of the load are great enough to limit the fluid ingress into the contact causing increases in friction. Similarly, the influence of stationary loading time has been shown to cause friction to increase *in vitro* in cartilage-cartilage contacts, providing additional evidence to the importance of the duration of load and not just the magnitude. The aim of this study was to analyse the characteristics of the hip joint reaction forces generated in the hip during a range of activities in healthy active persons specifically considering the magnitude, duration, impulse and the timing of the application of the load and motion.

2. Materials and methods

2.1. Motion capture

Eighteen subjects were recruited from staff and students at the University of Leeds (Table 1). Ethical approval was granted by The University of Leeds Ethics Committee (MEEC 16–021) and subjects completed informed consent forms/ screening questionnaires. All subjects were healthy and free from any injury, illness or pathology that could impact their natural gait. Methods with respect to movement and gait analysis were published in a previous publication by the authors [14].

Table 1
Demographics for the eighteen healthy subjects who completed thirteen common daily activities within a movement analysis laboratory [22].

Subject demographics	
N	18
Sex	10 Male 8 Female
Age Range	20 to 70
Age (Mean ±SD)	44 ± 19
Mass Range (kg)	50.2 to 106.1
Mass (kg) (Mean ±SD)	76.3 ± 13.1
Height Range (m)	1.5 to 1.8
Height (m) (Mean ±SD)	1.7 ± 0.1
BMI (kg/m^2) Range	19 to 35
BMI (kg/m^2) (Mean ±SD)	26 ± 4

Movement analysis involved the use of a robust proprietory commercial system designed for high speed sports analysis and included a thirteen camera Qualysis Oqus system (Qualysis Medical AB, Goeteborg, Sweden) operated at a frequency of 400 Hz and two 600 mm x 400 mm AMTI force plates (model BP400600, AMTI, Advanced Mechanical Technology Inc. Watertown, MA, USA) at a frequency of 1200 Hz. Gait analysis details included the usage a full-body analysis with 54, 15.9 mm diameter retro reflective markers (B&L Engineering, CA, USA) (Fig. 2). Within this full-body analysis the lower limb was modelled using twenty-eight markers attached to anatomical landmarks, including four semi-rigid thermoplastic shells, fitted with a total of sixteen tracking markers, attached to the thigh and shank. Data was filtered at 10 Hz and body segments were modelled using Visual 3D (Visual3D standard, v5.01.18, C-Motion, Germantown, MD, USA) [23]. Modelling the pelvis was completed using 2 aterior and 2 posterior superior iliac spinne markers to create a Visual 3D composite Pelvis. Trials where there was marker-dropout were not considered, hence the number of trials varied with both the subject and activity; the successful number of subjects ranged from 7 to 17 (Table 2). The hip joint centre was defined virtually uising the methods of Bell et al. [24]. Acetabular inclination and version were adopted from the methods of Jolles and Zanger [25].

Prior to dynamic trials, each subject completed a static trial in order to identify the positions of anatomical markers. This was followed by five trials for each of 9 activities Walk, Walk Turn, Stand to Sit, Sit to

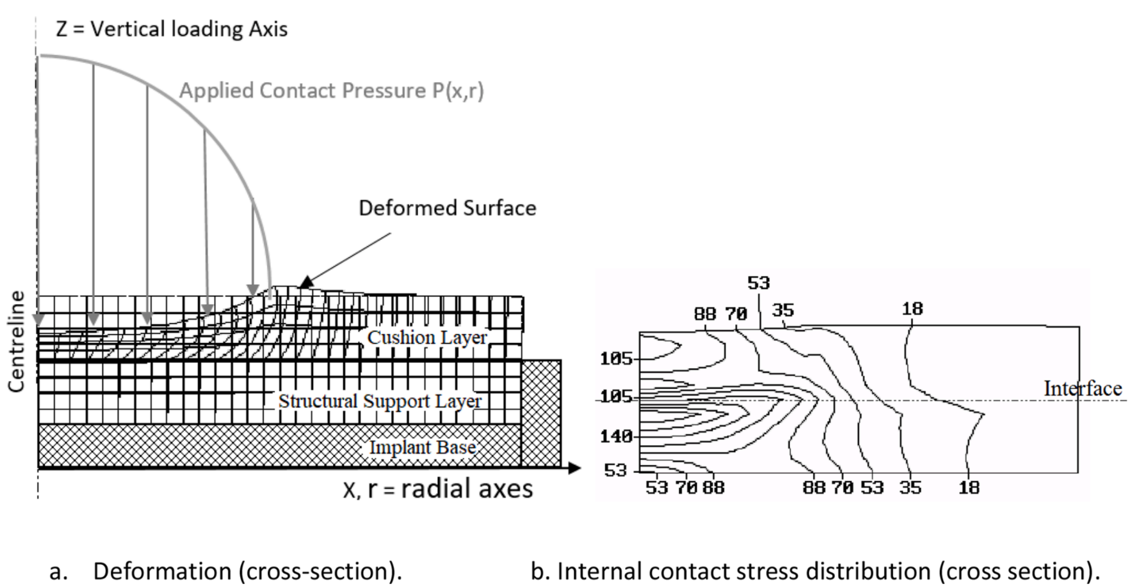


Fig. 1. Cross-sectional deformation (a) under normal loading in a 2 dimensional axisymmetric contact model of a compliant bi-layered structure where “z” is the central load axes, “x”, “r ” are the radial axes, and “P(x,r)” is the surface distribution of the axisymmetric contact pressure. Note that in this example the maximum shear stress contours (b), are not located at the surface but at the sub-surface interface where fixation can be crucially important [10].

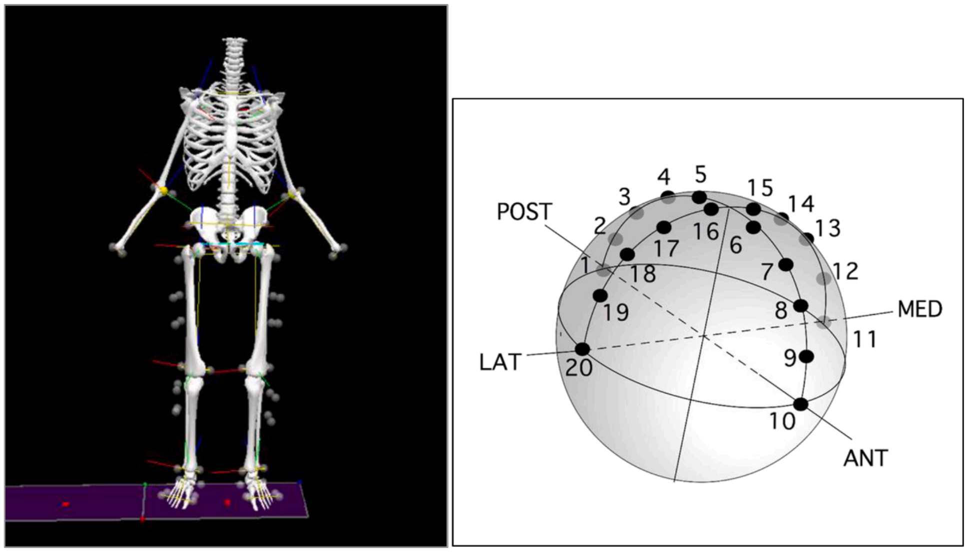


Fig. 2. Typical locations of markers used for full-body analysis (left) and Virtual Hip Joint (right) for localised analysis of motion [22]. The anatomical directions of movement are lateral (LAT), medial (MED), anterior (ANT) and posterior (POST).

Stand, Squat, Stand Reach, Kneel Reach, Lunge and Golf Swing. Activities were chosen specifically to include the movements that occur during common activities, specifically focussed on larger ranges of movement. Data from the right hip was evaluated in all cases.

Hip joint angles were defined through the orientation of the thigh segment in relation to the pelvis [23]. The number of subjects considered in the movement results for each activity varied due to the subjects ability to complete the activity and the degree of marker drop-out (camera screened by a portion of the subject). Analysis for hip joint reaction force also restricted subjects whom did not have a reliable force plate contact, hence subject numbers are reported where required.

Localised movement was evaluated using a novel Virtual Joint Model, developed by the authors, consisting of a virtual sphere that was constructed within Visual 3D located at the hip centre [1,2]. Twenty virtual markers were evenly placed across the spherical surface, 10 anterior-posterior and 10 medial-lateral to capture the localised paths of motion. Hence, whilst the global rotations of the hip centre were evaluated following a traditional analysis, the additional Virtual Joint Model facilitated the calculation of localised surface motion paths over the joint to be integrated into the motion analysis program and batch calculated [3,14]. Whilst motion path analysis has been completed in the past using matlab or similar computational models, the encorporation of this into Visual 3D by the authors offers a novel method to broaden the scope/impact of gait analysis.

To determine the sliding distance the diameter of the virtual hip was set to a 28 mm diameter sphere for all subjects and all activities to generate the individual motion paths for a standardised geometry comparable to past studies of joint replacements; this is smaller than a natural hip. The sliding velocity/acceleration were determined by differentiation of sliding distance/velocity respectively with an example shown in Fig. 3. More sudden changes in velocity lead to corresponding increases in local acceleration. Whilst a localised velocity change will influence tribology, a localised acceleration change is more likely linked to a change in direction and of interest when additionally considering strain/shear.

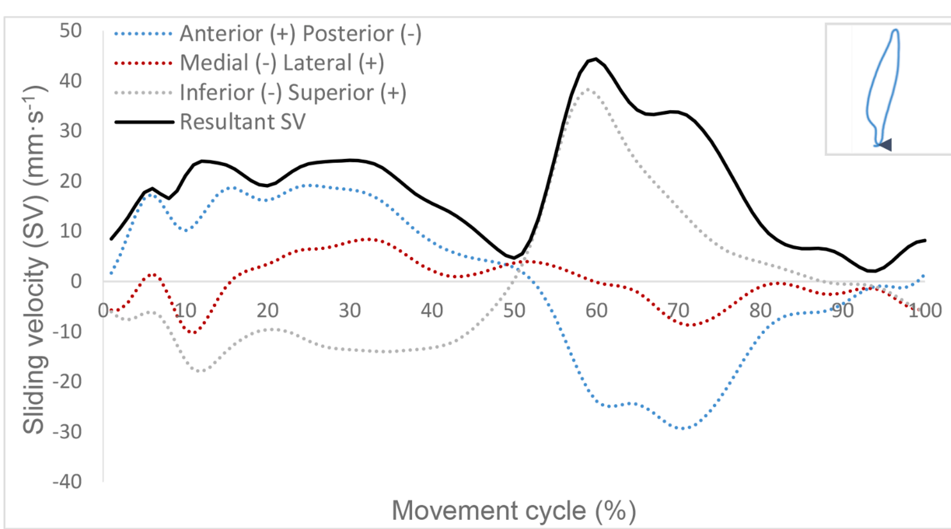


Fig. 3. Typical Sliding Velocities (Anterior-Posterior in Blue, Medial-Lateral in Red,Inferior-Superior in grey, and Resultant in Black) during walking for a point on the superior pole of the head. The small caption in the right corner depicts the displacement of the movement path in blue with the arrow depicting the start and direction of the movement cycle. Adopted from Layton [22].

2.2. Activities

For the Walk activity the subject was asked to walk at their normal pace with the cycle of the stride encompassing from right heel-strike to the next right heel-strike. The movement data collected from walking produced normal healthy biomechanics as reported in the literature with a range of motion of 44 +/- 9° flexion/extension, 6 +/- 11° internal/external rotation and 14 +/- 4° abduction/adduction [26–28].

In the Walk Turn the subject started perpendicular to the final direction of motion. After a right foot contact at ~30° flexion and ~8° internal rotation, the next left step occurred at 90° to the right foot, externally rotating about the right hip ~8°, until the subsequent left foot contact occurred. When the person turned 90° to the left with their right foot planted in stance the majority of the rotation was observed to occur around the right foot in the loaded limb and about the left hip (unloaded), hence movement in the right hip was limited. The standad deviation in movement of the right hip during the Walk Turn was +/- 10° flexion/extension and +/- 12° in internal/exteral rotation. In the Sit to Stand and Stand to Sit activities the subjects used a 47 cm stool and completed the activities un-aided by their hands (maximum hip flexion 81 +/- 26°). In the Squat the subjects were requested to squat down to as close to 90° as they felt comfortale with their back straight and arms forwards, pause, and then to return to a standing position; the mean hip flexion angle was 80 +/- 14°. In the Stand Reach (simulating picking something up) subjects, from a standing position, reached downwards in front of them to achieve full hip flexion but not to touch the floor, pause, and then return to a full upright position (Hip flexion 74 +/- 6 dedrees). In the Kneel Reach the subject started from an upright kneeling position and then reached forwards as far as they could, paused, and then returned to a vertical torso; this simulated washing a floor or gardening (Hip flexion 76 +/- 14°). In contrast in the Lunge, from standing upright, the subject led with their right leg and lunged forwards with their left knee touching the floor (Hip flexion 74 +/- 6 dedrees). In the Golf Swing the subjects repeated a tee-off manoeuvre, including standing, the internal rotation of the right hip (backswing) ending with the club over the right shoulder, external rotation for the club impact and finally the full follow-through with the club ending up over the left shoulder. Hip flexion during the Golf Swing was relatively low (21–24° backswing) with motion in all planes including internal then external rotation (22 and 21° +/- 14° respectively) and adduction (11° +/- 10°).

Abduction/Adduction, and Internal/External rotation movements were generally less than 5° in most activities with the exception of the Golf Swing and Walking activities where values were larger and in the region of 10°. Thus the most significant movement axes observed in the activities within the study was flexion.

2.3. Hip joint reaction forces

Motion capture and ground reaction force data was imported into the AnyBody™ multi-body dynamics modelling system (AnyBody, version 6.0, AnyBody Technology, Aalborg, Denmark). The Twente Lower Extremity Model (TLEM), taken from the open access, AnyBody Repository, was used for analysis. The previously validated musculoskeletal lower-extremity model was adapted and used to perform inverse dynamic calculations, in order to estimate hip reaction forces [29–32]. The 6° of freedom model incorporates 159 muscles and 11 rigid bodies representing the talus, foot, shank, patella and thigh for both legs, plus the pelvis. Trunk segments were also included within the model in order to provide attachment sites for the psoas major muscles, and were constrained to the pelvis. Muscles, joint centres and inertial parameters for the model are based on an anthropometric data set from the University of Twente [32]. The virtual joint model was independent of the analysis of joint reaction forces.

Forces were analysed at the same normalised time period for all subjects and then averaged for the number of subjects with a standard deviation reported in error bars. The magnitude of force was additionally normalised to body weight (ρBWT) in all cases. Impulse was calculated at each of the normalised time periods where load x time was integrated for each subject and then averaged for the cycle to allow the loading rate to be considered [33]. Duration of loading was normalised as percentage of the gait cycle in all cases with 100 time steps to facilitate analysis. Statistical analysis was completed using Microsoft Excel to investigate the effects of age and gender within the subjects using a paired T-Test with a two-tailed analysis of variance and a confidence internal of 0.05. Mean and standard deviation was calculated in all cases to understand the extent of the variability in measurements and activities. The number of sucessfult trials for each subject used within the calculations are shown in Table 2.

3. Results

3.1. Force magnitude

The average resultant peak hip reaction force, normalized to body weight, for the 9 activities is shown in Fig. 4 and listed as a vector in Table 2. Peak hip joint reaction forces varied from 0.5 to 9 times body weight. Sit to stand produced the greatest average peak reaction force in excess of 6 body weight to generate lift without the use of arms. Kneel reach produced the lowest hip reaction forces ~0.5 body weight. The remainder of the activities produced peak reaction forces of 3–4 body weight similar to walking. Peak hip reaction force vectors were generally medial, posterior and proximal.

The variation in the average hip reaction force over the entire movement cycle for the 9 activities in each of the three anatomical axes is shown in Fig. 5. The greatest magnitude of forces occurred proximally in all cases up to 8 body weight, with medial forces 1-2 body weight and posterior loads generally less than 0.5 body weight present at various periods in 7 of the 9 activities.

Hip peak loading occurred when ground reaction forces were high, if not maximal. The Walk peak loading occurred at 46% of the cycle, which corresponded approximately to the timing of heel-off (in anticipation for the propulsive phase of gait). It is noteworthy that a similar magnitude of load resulted at heel-strike. In the Walk Turn peak force occurred following heel-strike (13%), when the lower limb loading rate was high. Stand to Sit hip force was highest following initial contact with the seat. In contrast, the Sit to Stand hip reaction force was highest just before leaving the seat (58%). The Squat peak hip reaction force occurred at approximately 50% of the movement cycle, when the centre

Table 2
Average values for peak resultant hip reaction force and vector orientation (relative to *X*: Horizontal axes = Flexion; *Y*: Anterior axes = Abduction; *Z*: Vertical axes = Rotation) are shown along with the associated hip angle for the 9 activities along with the number of subjects where the analysis of reaction force was feasible. Force has been normalized to body weight (ρBWT). Error bars represent 1 standard deviation above and below the mean.

Activity	Peak resultant force ± 1 SD (Vector angle: *X, Y, Z*) (ρBWT)	Hip Angle(X,Y, Z) (degrees)	Number of Sucessful Subjects
Walk	3.1 ± 1.1 (110°, 82°, 22°)	(−5°, 5°, 6°)	16
Walk turn	4.1 ± 1.1 (113°, 80°, 25°)	(28°, −1°, 0°)	17
Stand to sit	4.2 ± 1.1 (113°, 83°, 24°)	(80°, −2°, 2°)	7
Sit to stand	6.4 ± 2.6 (112°, 83°, 23°)	(74°, −4°, 5°)	7
Squat	3.5 ± 1.9 (111°, 86°, 21°)	(80°, −9°, 1°)	9
Stand reach	4.3 ± 1.0 (104°, 87°, 15°)	(70°, 0°, 5°)	12
Kneel reach	0.5 ± 0.1 (104°, 80°, 17°)	(68°, −7°, 10°)	10
Lunge	3.1 ± 1.3 (97°, 86°, 8°)	(−5°, 7°, 7°)	13
Golf swing	4.0 ± 1.9 (115°, 86°, 25°)	(23°, 4°, 11°)	16

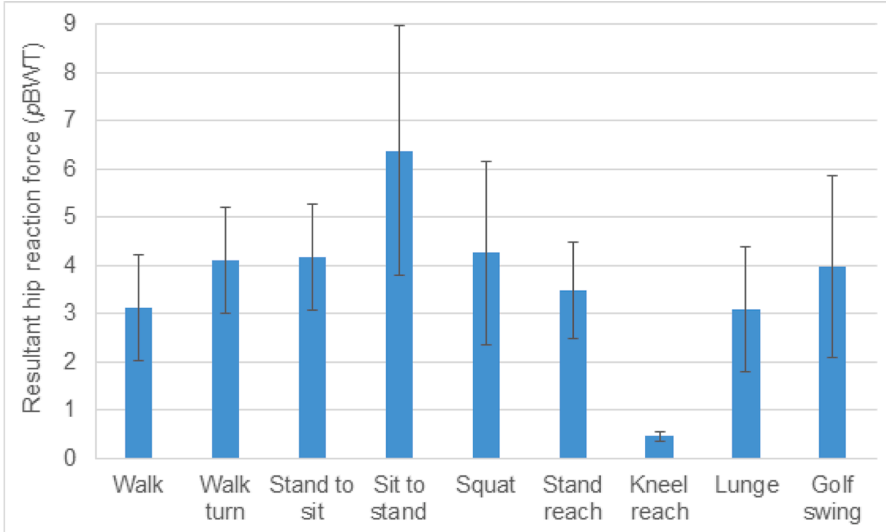


Fig. 4. Average Peak resultant hip reaction force for a range of activities normalized to body weight (ρBWT). The greatest hip reaction forces were observed in the Sit to stand activity. Error bars represent 1 standard deviation above and below the mean.

of mass was at its lowest and the hip was fully flexed. Stand Reach and Kneel Reach hip forces peaked at 35% and 31%, respectively, at points approaching full hip flexion. The Lunge reached peak forces at 100% of the cycle, when the right heel had lifted off the floor and the body was propelled upwards out of the lunge position (this corresponded to the timing of the maximum ground reaction force). The Golf Swing peak hip force occurred during the downswing, at 63% of the movement cycle (when the ground reaction force was also maximal).

3.2. Duration

The duration of an activity can provide an indication of the general time that the load is applied for [Fig. 6]. For the 9 activities the longest duration activity was the Kneel Reach (5.9 ± 1.6 s) that simulated common gardening activities. Other Squat and Lunge activities had durations of 2–4 s, with the walking activities generally lasting ~1 s. There was a large variation in the duration of the reach/lunge/golf activities depending on how the person preferred to complete the activity.

3.3. Impulse

The average of the integration of the reaction force and duration of the activities is shown in Fig. 7. The Golf Swing, Lunge and Stand Reach produced the greatest impulse. These were up to 3 times greater than for walking. The Kneel Reach, despite having a very long duration, had very small hip reaction forces, thus its impulse was the lowest.

3.4. Timings of loading and motion

The sliding velocity and acceleration occuring at a point on the superior pole of the head is plotted against the hip joint reaction force in Fig. 8 for all of the activities. The point was chosen as it was within the contact area for all activities (Point 7, Fig. 2). The range of the load is shown over the horizontal axes plotted against the range of velocity (red) and the range of acceleration (blue) across the vertical axes. Hence, by considering the load and local movement the potential effects on tribology can be considered more holistically [20,34]. The worst potential tribological conditions were observed in the Walk Turn and Walk where high load and high acceleration (shear) is present simultaneously. The Sit to Stand activity is also a concern with high loads and low velocities that may limit fluid entrainment.

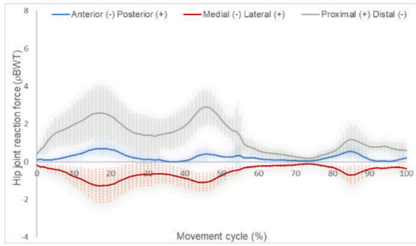
The Leeds Prosim simulator, that represents and an ISO cycle (Fig. 8j) applies a smoother application of movement with a reduced acceleration.

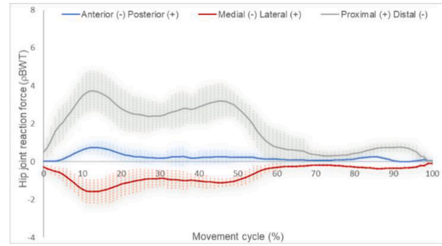
4. Discussion

The peak hip joint reaction forces determined in this study for walking were similar to the magnitudes reported is the literature for healthy subjects [26–28]. The average peak hip reaction forces found within the other activities ranged from 0.5 to 6.4 body weight, greater than observed in walking in many cases and the greatest in the Sit to Stand activity (Fig. 4). Deviations were observed in the manner in which the activities were completed by the subjects, however, this is common in the literature. Three times body weight is likely the most commonly used reaction force in mechanical testing thus the larger values found here may be important (Fig. 5). The magnitude of forces is important for materials testing to consider their mechanical strength, subsurface shear, and general suitability to an application. In engineering terms a safety factor would also normally be multiplied to these values. However, the use of safety factors in medical implant design has not been broadly published since the biomechanical constraints required by the surgeon and patient are often the controlling design influences and thus long term durability testing is required by regulators to confirm structural integrity.

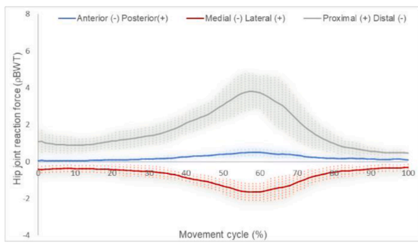
People who are receiving cartilage repair procedures are often very active and following recovery will be returning to normal activity. It is important to consider that the results of the study thus represent an analysis of well-functioning normal healthy subjects and are thus not applicable to older patients with compromised gait or contralateral joint problems more typical of a joint replacement cohort [26–28,35]. Considering subject variation, in the present study within the cohort older persons >55 showed increased average peak hip reaction forces (1.1 body weight) only during the Golf Swing activity; only 1 of the subjects in this age group was a regular golfer and none of the other subjects participated in regular sports. Increased average peak hip reaction forces (1.2 body weight) were also observed in females during the squat with a corresponding increase in hip flexion (10°) compared to males. No other significant age or gender differences were observed in the study for any of the activities making conclusions regarding heterogeneity limited.

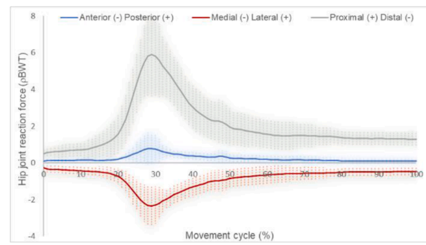
Recent studies reporting the biomechanics of hip joint replacements have presented a wide variability in results with magnitudes of forces increasing with the functional ability of the subject in completing the



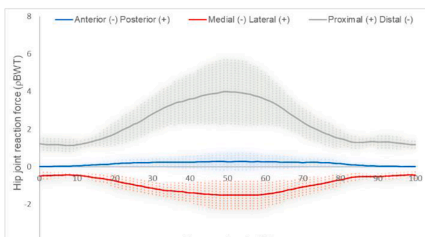


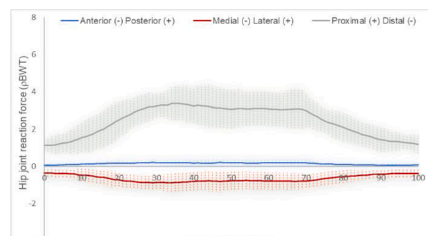
a. Walk, n=16 b. Walk Turn, n=17



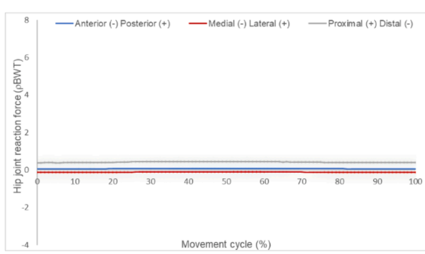


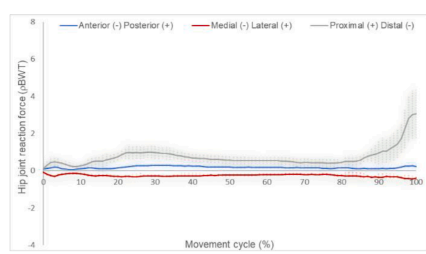
c. Stand to Sit, n=7 d. Sit to Stand, n=7



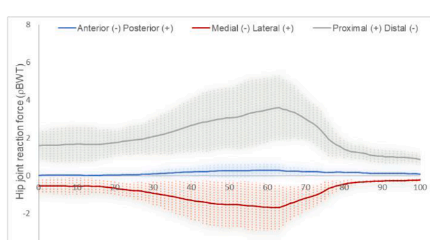


e. Squat, n=9 f. Stand Reach, n=12





g. Kneel Reach, n=10 h. Lunge, n=13



i. Golf Swing, n=16

Fig. 5. (a–i). Mean hip reaction force during one movement cycle for 9 activities normalized to body weight (ρBWT). Shaded areas represent one Standard Deviation above and below the mean for the number of subjects indicated.

activity [26–28,35,36]. The results of the present study showed the same general trends in the variability of the magnitudes of force and were comparable, perhaps, to a high functioning hip replacement patient. It is important to remember that these activities will be performed much less dynamically in older persons or those with symptomatic gait problems, but more importantly will be much more dynamic in a healthy normal subject.

With regards to the duration of the application of forces within the hip joint, the walking activities were the shortest with all others lasting much longer (Fig. 6). Contacts under relative motion, particularly

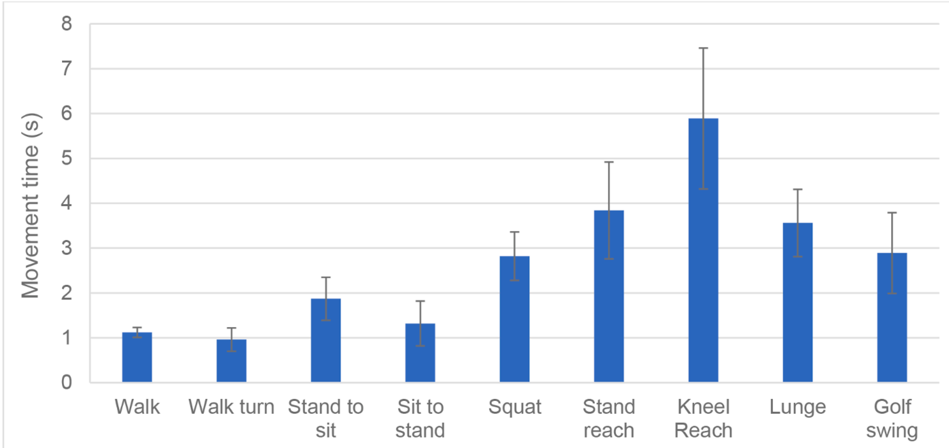


Fig. 6. Mean movement time for a range of activities. Several activities did not follow the typical 1 Hz used to represent most pre-clinical assessment tests. Error bars represent 1 standard deviation above and below the mean.

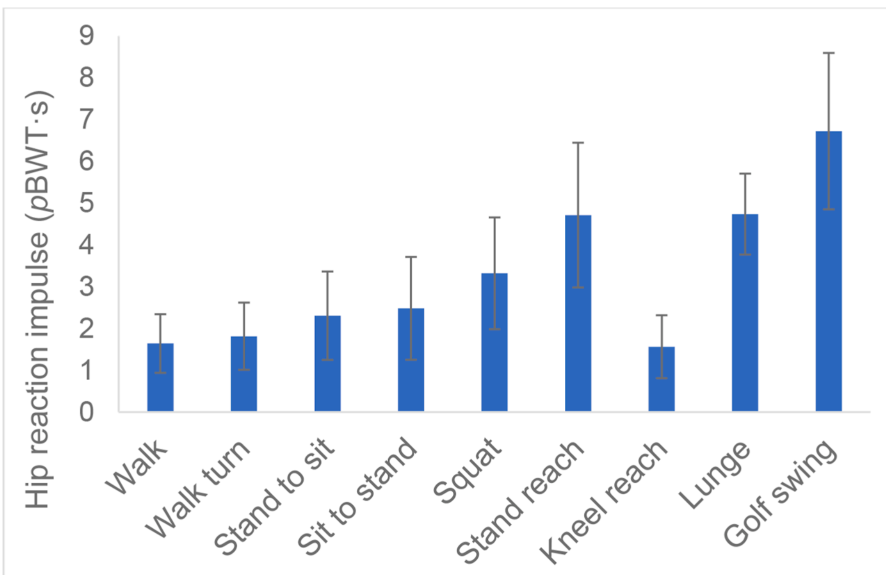


Fig. 7. Average Peak hip reaction impulse for a range of activities normalized to body weight (ρBWT). Most of the activities had a greater impulse compared to walking. Error bars represent one standard deviation of the mean.

between soft surfaces, can cause time dependant depletion of fluid films, whilst large movements under low loads may replenish fluid films. The duration of loading is, therefore, crucially important for the consideration of the tribological effects of load influencing the lubricant film [20]. In past cyclic load cartilage friction testing studies, the duration of testing has generally been set at 1 s to reflect a typical walking stride. This generates a beneficial fluid film lubrication in natural joints as modelled by Jin et al. [34,37]. Cartilage surface friction testing, however, has been completed under longer durations up to several minutes to consider depletion of lubricant within the material or contact in biphasic materials [9,15,16,21] . These longer duration tests perhaps represent other activities to walking that have longer durations. However, the results of the present study suggest that typical repetitive cyclic activities would be unlikely to last more than ~6 s.

Impulse was defined, in this study, as the force applied to the cartilage surface integrated over the time period of the activity (Fig. 7). The characteristics of force and time are important in the assessment of contact mechanics and fatigue in terms of the resulting displacement / work / energy [7]. The impulse imparted in the hip joint contact during the 9 activities was the lowest during simple walking and the greatest during the Golf Swing. This suggests that the lifestyle of the person might be important in choosing a suitable treatment or pre-clincial testing regime [33].

In vitro studies of cartilage friction/wear, have shown that the magnitude and duration of load application is important, providing evidence supporting the theory of the bi-phasic nature of cartilage [15,16,19,38–40]. Examining the detailed movement at a single point on the joint surfaces provides a holistic view of the timings of loads and movements occurring simultaneously and this can help understand the important parameters for implant development / testing (Fig. 8). In addition the cartilage surfaces of the hip are soft and very conforming, producing a large contact area. Under pure rotation only there will be variable motion from the centre to the periphery of the contact. This variability will only increase under complex activities. In this study the authors encorporated a virtual hip model within Visual 3D that has the ability to consider any localised point on the surface of the head and its relative movement against the acetabulum.

Present pre-clincial testing for hip implants utilizes a smoothed walking cycle for the application of load over a 1 s cycle time. The smoothing is completed to facilitate the programming of a machine to

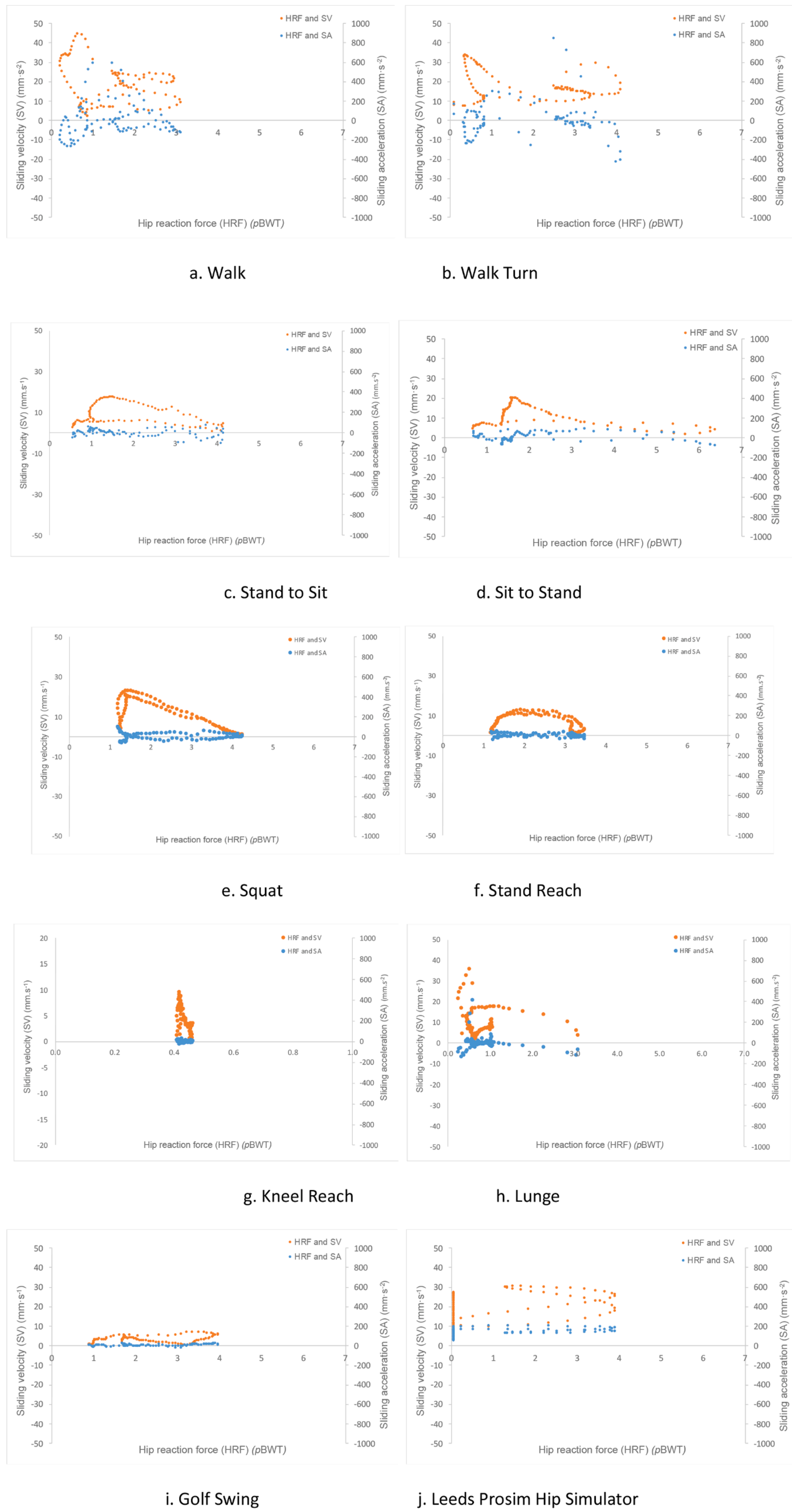


Fig. 8. (a–i). Typical timings of hip joint reaction force (HRF, normalized to body weight (ρBWT)) at a fixed point on the femoral head plotted against the corresponding sliding velocity (SV, red, mm/s) and sliding acceleration (SA, blue, mm/s^2) for 9 activities compared to the Leeds Prosim ISO hip simulator.

replicate the cyce of motion / load. The use of the Virtual Hip Joint model allows the detailed path of movement of a surface point within the joint to be investigated. This has demonstrated the variability in acceleration that occurs locally in our joints but is traditionally ignored (Fig. 8). With regards to tribology high loads with low velocity, such as observed in the Sit to Stand activity, would likely cause depletion of protective fluid films. Likewise the combination of high load with high acceleration, as observed in the Walk Turn may lead to excessive shear forces. These localized conditions may be crucially important depending upon where a specific cartilage repair mechanism is placed and the activity being completed. The study has demonstrated that it is important that when alternative activities are simulated for testing purposes that the characteristics of the activity that might influence the tribology or stress are fully understood and not lost by the time they are adopted as a standard.

Whilst magnitudes of load are commonly reported in the literature researchers often miss out on the actual characteristics of the load that occur in our joints. The future development of testing standards for tissue engineering or for novel biological materials/composites produced through processes such as bioprinting presents a challenge to the research community [41–43]. Fundamental research has shown that compliant surfaces are more sensitive to lubrication and shear thus the holistic effects of the lower and higher extremes of the magnitude/-timing of load and motion will be important in the tribology at the implant surface [39]. Maximum sub-surface shear stresses in these surfaces will additionally likley occur at or near fixation interfaces with the bone, making surface analysis very important. It should be remembered that joint replacements do not perform well in young active persons and that any alternative solution should undergo rigorous evaluation [13].

4.1. Study limmitations

The paper is intended to raise awareness to additional considerations that might be important for the assessment of cartilage mechanics. It is only through testing of cartilage repair devices themselves that researchers can determine the critical variables limiting their application and thus develop an informed testing protocol.

Subjects were asked to complete the activity to the best of their ability, but this was not prescribed in detail thus there was significant variability in the magnitudes and timings. This type of variability is common when considering a normal or a patient population where more active subjects often complete the activity more dynamically. Additionally, variation in marker drop-out had an effect on the numbers of subjects that produced reliable data for each activity that ranged from 7 to 16; a potential weakness of the study.

Analysis of impulse presented in the paper is a very crude approximation and represents only the contribution of the reaction force and timing of the activity. More complex analysis that includes subject specific modelling may increase the accuracy of the force results reported, however these generally are much more time consuming and less favourable for larger cohorts [27]. The work of Bergmann (Orthoload.com) oferrs a repository of data from instrumented hip joint replacement prostheses [26]. The hip joint reaction forces reported in this database were lower for the sitting activities compared to this study, however the movement times were also greater. It should be noted that healthy subjects complete activities much more dynamically than a person with a hip replacement and the authors feel that the values from Orthoload may be inappropriate for testing a cartilage substiture being implanted into a young person.

Whilst motion path analysis has been completed in the past using matlab or similar computational models, the encorporation of this technique into Visual 3D by the authors offers a novel method to broaden the impact of gait analysis. The authors applied 20 points of analysis evenly across the anterior-posterior and medial-lateral planes of the joint surface. Whilst this point spread covers a broad range of the surface, it is not all encompassing. However, the method allows a specific area of interest to be focussed upon by simply moving the selected points / planes within the Virtual Hip Joint model. Additionally within this publication the authors have only reported the conditions of movement at a single point. This point was selected near the superior pole of the joint surface where the maximum joint reaction force was more commonly located within the analsyis.

The paper focusses on the hip as the virtual joint model was developed specifically for this joint and it presents a fundamentally spherical contact with lower stresses than the knee or ankle. The knee has challenges associated with detecting larger translations (roll-back) in the gait model and capturing an accurate representation of the natural medial-lateral asymmetry. The clinical need in the knee is additionally far greater than within the hip since knee arthroscopy and cartilage repair is commonplace [44,45]. Cartilage repair in the ankle is perhaps the most challenging as the space is limited, joint replacement being less successful, and fusion known to transfer biomechanical stresses to other joints within the foot [46,47]. Clinical solutions such as joint distraction have also been introduced in attempts to prevent loading and motion from occurring within the joint space to allow cartilage to repair, thus highlighting the importance of movement and load. In future it is hoped that the virtual model can be refined to include translational components of joint micro-separation and soft tissue impingement to be studied.

5. Conclusion

The authors have demonstrated that using a novel Virtual Hip Joint model within gait analysis offers the potential to better understand the holistic nature of activities in more detail by being able to visualise the timings of loading and movement. The hip reaction forces and their characteristics were demonstrated at times to be more severe than existing standards represent. The study found that during normal activities average peak joint forces varied from 0.5 to 6.4 times body weight. The duration of these activities also varied from 1 to 6 s. Person to person variability in a healthy cohort accounted for ~30% variation. The differences in activities accounted for up to a 3-fold variation in expected impulse compared to simple walking.

Ethical approval

The study protocol was ethically approved according to the guidelines of The University of Leeds Ethics Committee (MEEC 16–021).

Author contributions

The work represents the output of the 1st authors PhD thesis with the co-authors being the supervisors. Hence, all authors were direcly involved in all aspects of the submission.

Declaration of Competing Interest

None declared.

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References

[1] Ramamurti BS, Bragdon CR, O'connor DO, et al. Loci of movement of selected points on the femoral head during normal gait: three-dimensional computer simulation. J Arthroplasty 1996;11:845–52. https://doi.org/10.1016/s0883-5403(96)80185-9.

[2] Bennett DB, Orr JF, Baker R. Movement loci of selected points on the femoral head for individual total hip arthroplasty patients using three-dimensional computer simulation. J Arthroplasty 2000;15(7):909–15. https://doi.org/10.1054/arth.2000.9195.
[3] Budenberg S, Redmond A, White D, et al. Contact surface motion paths associated with leg length inequality following unilateral total hip replacement. Proc Inst Mech Eng Med 2012;226:968–74. https://doi.org/10.1177/0954411912453245.
[4] Shen F, Mckellop H. Surface-gradient cross-linked polyethylene acetabular cups oxidation resistance and wear against smooth and rough femoral balls. Clin Orthop Relat Res 2005;430:80–8.
[5] Wang A, Sun DC, Yau SS, et al. Orientation softening in the deformation and wear of ultra-high molecular weight polyethylene. Wear 1997;203:230–41. https://doi.org/10.1016/S0043-1648%2896%2907362-0.
[6] Galvin A, Kang L, Tipper J, et al. Wear of crosslinked polyethylene under different tribological conditions. J Mater Sci Mater Med 2006;17(3):235–43. https://doi.org/10.1007/s10856-006-7309-z.
[7] Johnson KL. Contact mechanics. Cambridge University Press; 1987. https://doi.org/10.1017/CBO9781139171731.
[8] Bartel D, Burstein AH, Toda MD, et al. The effect of conformity and plastic thickness on contact stresses in metal-backed plastic implants. J Biomech Eng 1985;107(3):193–9. https://doi.org/10.1115/1.3138543.
[9] Li J, Stewart TD, Jin ZM, Wilcox RK, Fisher J. The influence of size, clearance, cartilage properties, thickness and hemiarthroplasty on the contact mechanics of the hip joint with biphasic layers. J Biomech 2013;46(10):1641–7. https://doi.org/10.1016/j.jbiomech.2013.04.009.
[10] Stewart T, Jin ZM, Fisher J. Analysis of contact mechanics for composite cushion knee joint replacements. Proc Inst Mech Eng Part H 1998;212(1):1–10. https://doi.org/10.1243/0954411981533773.
[11] BS ISO 14242-1:2014+A1:2018. Implants for surgery - wear of total hip prostheses - Part 1: loading and displacement parameters for wear-testing machines and corresponding environmental conditions for tests. Int Organ Stand 2018.
[12] Liu F, Fisher J, Jin ZM. Effect of motion inputs on the wear prediction of artificial hip joints. Tribol Int 2013;63:105–14. 10.1016%2Fj.triboint.2012.05.029.
[13] Ben-Shlomo, Y.; Blom, A.; Boulton, C., et al. The national joint registry 17th Annual Report. reports.njrcentre.org.uk 2020; London.
[14] Layton R, Messenger N, Stewart T. Analysis of hip joint cross-shear under variable activities using a novel virtual joint model within Visual3D. Proc Inst Mech Eng H 2021;235(10):1197–204. https://doi.org/10.1177/09544119211025869.
[15] Katta J, Jin ZM, Ingham E, Fisher J. Biotribology of articular cartilage—a review of the recent advances. Med Eng Phys 2008;30(10):1349–63. https://doi.org/10.1016/j.medengphy.2008.09.004.
[16] Forster H, Fisher J. The influence of loading time and lubricant on the friction of articular cartilage. Proc Inst Mech Eng Part H 1996;210(2):109–19. https://doi.org/10.1243/pime_proc_1996_210_399_02.
[17] Rosetti L, Desando G, Cavallo C, Petretta M, Grigolo B. Review, articular cartilage regeneration in osteoarthritis. Cells 2019;8(11):1305. https://doi.org/10.3390/cells8111305.
[18] Yucelkul A, Ozdil D, Kuylu NH, Erdemli E, et al. Tri-layered composite plug for the repair of osteochondral defects *in-vivo* study in sheep. J Tissue Eng 2017;8:1–10. https://doi.org/10.1177/2041731417697500.
[19] Stewart T, Jin ZM, Fisher J. Friction of composite cushion bearings for total knee joint replacements under adverse lubrication conditions. In: 211; 1997. p. 451–65. https://doi.org/10.1243/0954411981534574. journal of engineering in medicine.
[20] Higginson G. Squeeze films between compliant solids. Wear 1978;46(2):387–95. https://doi.org/10.1016/0043-1648(78)90042-X.
[21] Scholes SC, Burgess IC, Marsden HR, Unsworth A, et al. Compliant layer acetabular cups: friction testing of a range of materials and designs for a new generation of prosthesis that mimics the natural joint. In: ; 2006. p. 583–96. https://doi.org/10.1243/09544119H06404. Vol. 220.
[22] Layton, R.B. Understanding movement and its influence on tribology of the human hip. White rose E-Thesis on-line. University of Leeds PhD Thesis 2020.
[23] Franjcic, Z.; Wozniak, P.W.QW: Towards optimising feedback for motion capture system calibration. In: Graphics interface 2014; Poster Session Proceedings, pp.5–6.
[24] Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. J Biomech 1990;23(6):617–21. https://doi.org/10.1016/0021-9290(90)90054-7.
[25] Meftah M, Yadav A, Wong AC, et al. A novel method for accurate and reproducible functional cup positioning in total hip arthroplasty. J Arthroplasty 2013;28(7):1200–5. https://doi.org/10.1016/j.arth.2012.09.018.
[26] Bergman nG, Bender A, Dymke J, Duda G, Damm P. Standardized loads acting in hip implants. PLoS One 2016;11:e0155612. https://doi.org/10.1371/journal.pone.0155612.
[27] Depieri E, Lunn DE, Chapman GC, Rasmussen KP, et al. Patient characteristics affect hip contact forces during gait. Osteoarthr Cartil 2019;27:895–905. https://doi.org/10.1016/j.joca.2019.01.016.
[28] Lunn DE, Depieri E, Chapman GC, et al. Current pre-clinical testing of new hip arthroplasty technologies does not reflect real-world loadings: capturing patient-specific and activity-related variation in hip contact forces. J Arthroplasty 2020;35:877–85. https://doi.org/10.1016/j.arth.2019.10.006.
[29] Carbone V, Fluit R, Pellikaan P, et al. TLEM 2.0–A comprehensive musculoskeletal geometry dataset for subject-specific modeling of lower extremity. J Biomech 2015;48(5):734–41. https://doi.org/10.1016/j.jbiomech.2014.12.034.
[30] Forster, E. Predicting muscle forces in the human lower limb during locomotion. Universität Ulm PhD Thesis 2004. 10.18725/OPARU-214 .
[31] Manders C, New A, Rasmussen J. Validation of musculoskeletal gait simulation for use in investigation of total hip replacement. J Biomech 2008;41:pS488.
[32] Horsman, M.K. The twente lower extremity model. consistent dynamic simulation of the human locomotor apparatus. University of Twente PhD Thesis 2007.
[33] Patel JM, Wise BC, Bonnevie ED, Mauck RL. A sytematic review and guide to mechanical testing for articular cartilage tissue engineering. Tissue Eng 2019;V25 (10). https://doi.org/10.1089/ten.TEC.2019.0116.
[34] Hamrock BJ, Dowson D. Elastohydrodynamic lubrication of elliptical contacts for materials of low elastic modulus I—fully flooded conjunction. J Lubr Technol 1978;100(2):236–45.
[35] Li J, Redmond AC, Jin Z, et al. Hip contact forces in asymptomatic total hip replacement patients differ from normal healthy individuals: implications for preclinical testing. Clin Biomech 2014;29:747–51. https://doi.org/10.1016/j.clinbiomech.2014.06.005.
[36] Bergmann G, Deuretzbacher G, Heller M, et al. Hip contact forces and gait patterns from routine activities. J Biomech 2001;34(7):859–71. https://doi.org/10.1016/s0021-9290(01)00040-9.
[37] Jin Z, Dowson D, Fisher J. Fluid film lubrication in natural hip joints. Tribol Ser 1993;25:545–55. https://doi.org/10.1016/S0167-8922(08)70410-3. Elsevier.
[38] Lipshitz H, Etheredge R, Glimcher MJ. *In vitro* studies of the wear of articular cartilage—III. The wear characteristics of chemically modified articular cartilage when worn against a highly polished characterized stainless steel surface. J Biomech 1980;13(5):423–36. https://doi.org/10.1016/0021-9290(80)90036-6.
[39] Murakami T, Sakai N, Yarimitsu S, et al. Evaluation of influence of changes in permeability with aging on friction and biphasic behaviors of artificial hydrogel cartilage. Biotribology 2021;26:1–12. https://doi.org/10.1016/j.biotri.2021.100178.
[40] Mow V, Ateshian G. Lubrication and wear of diarthrodial jointsMow VC, Hayes WC, editors. Lubrication and wear of diarthrodial joints. Basic orthopaedic biomechanics 1997. ISBN 0397516843.
[41] Liu Y, Zhou G, Cao Y. Recent progress in cartilage tissue engineering - our experience and future directions. Engineering 2017;3(1):28–35. https://doi.org/10.1016/J.ENG.2017.01.010.
[42] Hu F, Lu H, Ye Z, et al. Slow-release lubrication of artificial joints using self-healing polyvinyl alcohol/polyethylene glycol/graphene oxide hydrogel. J Mech Behav Biomed Mater 2021;124. https://doi.org/10.1016/j.jmbbm.2021.104807.
[43] Agarwal T, Chiesa I, Presutti D, et al. Recent advances in bioprinting technologies for engineering different cartilage-based tissues. Mater Sci Eng C Mater Biol Appl 2021;123. https://doi.org/10.1016/j.msec.2021.112005.
[44] Khajehsaeid H, Abdollahpour Z. Progressive deformation-induced degradation of knee articular cartilage and osteoarthritis. J Biomech 2020;111. https://doi.org/10.1016/j.jbiomech..109995.
[45] Manda K, Ryd L, Eriksson A. Finite element simulations of a focal knee resurfacing implant applied to localized cartilage defects in a sheep model. J Biomech 2011;44:794–801. https://doi.org/10.1016/j.jbiomech.2010.12.026.
[46] Astephen Wilson JL, Kobsar D. Osteoarthritis year in review 2020: mechanics. Osteoarthr Cartil 2021;29(2):161–9. https://doi.org/10.1016/j.joca.2020.12.009.
[47] Hlavácek M. Lubrication of the human ankle joint in walking with the synovial fluid filtrated by the cartilage with the surface zone worn out: steady pure sliding motion. J Biomech 1999;32(10):1059–69. https://doi.org/10.1016/s0021-9290(99)00095-0.