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Analysis of hip joint cross-shear under variable activities using a novel virtual joint model within Visual3D.

Abstract

Cross-shear forces occur between bearing surfaces at the hip and have been identified as a key contributor to prosthesis wear. Understanding the variation in relative motion paths between both individuals and activities, is a possible explanation for increased revision rates for younger patients and could assist in improved pre-clinical testing regimes. Additionally, there is little information for the pre-clinical testing of cartilage substitution therapies for younger more active individuals. The calculation of motion paths has previously relied on computational modelling software which can be complex and time-consuming. The aim of this study was to determine whether the motion paths calculations could be integrated into gait analysis software to improve batch processing, reduce analysis time and ultimately improve the efficiency of the analysis of cross-shear variation for a broader range of activities.

A novel Virtual Joint model was developed within Visual3D for calculating motion paths. This model was compared to previous computational methods and found to provide a competitive solution for cross shear analysis (accuracy <0.01 mm error between methods). The virtual hip model was subsequently applied to 13 common activities to investigate local aspect ratio's, velocities and accelerations. Surprisingly walking produced the harshest cross shear motion paths in subjects. Within walking, of additional interest was that the localised change in acceleration for subjects was 6 times greater compared to the same point on an equivalent smoothed simulator cycle.

The Virtual hip developed in Visual 3D provides a time saving technique for visualising and processing large data sets directly from motion files. The authors postulate that rather than focussing on a generalised smoothed cross-shear model that pre-clinical testing of more delicate structures should consider localised changes in acceleration as these may be more important in the assessment of cartilage substitutes sensitive to shear.

Key Words: Gait Analysis, Hip Biomechanics, Hip Protheses, Hip Simulators, Wear Analysis/ Testing [Biomechanics], Tribology of Materials

Word count: 4816

1.0 Introduction

From the 1st of January 2017 to the 31st of December 2019 281196 primary total hip replacements were implanted in the UK with the majority utilising a metal or ceramic on polyethylene bearing combination [1]. Revision rates, generally caused by wear debris induced osteolysis, showed an inverse relationship in comparison to the age of the patient [1, 2]. This is believed to be related to patient activity and has raised concerns surrounding the reasons behind the increased risk of prosthesis failure for some individuals [1]. The decrease in implant longevity for this younger and potentially less symptomatic group is thought to be linked to the greater physical demands placed on joints, with a corresponding increase in wear [3, 4]. The corresponding wear of the cup has been shown to be proportional to the degree of cross-shear motion occurring between the bearing surfaces along with the load, and the relative sliding distance [5, 6].

During unidirectional motion the polyethylene material on the surface of the acetabular cup will experience strain hardening, whereby the polyethylene molecules are stretched and re-orientate in the principal direction of sliding, ultimately increasing the materials resistance to wear in that direction [5-10]. Prolonged and repetitive multi-directional motion causes cross-shear of the polyethylene cup, due to the crossing and overlapping of motion paths leading to increased wear [7, 11]. In 2013, Schwenke and Wimmer suggested that 6.4 times more work was required to remove 1 mm^3 of wear in the principle molecular orientation, compared to at an angle of 90° [12]. Investigation of cross-shear involves the analysis of the trajectory (motion path) of a singular point on the femoral head moving against the surface of the polyethylene acetabular cup. During walking, these motion paths have generally shown to be quasi-elliptical, arc, or complex figures of eight in shape that vary dramatically depending on location [7, 11,13-15]. The shape, length and the crossing of motion paths will therefore influence wear of polyethylene in a hip replacement. For this reason, the authors postulate that the activity a patient is undertaking may be very important.

Simulation of motion paths has previously involved initial gait analysis followed by subsequent computational modelling [13, 15-17]. Previous work has assessed motion paths using input angles from hip simulator ISO cycles, total hip replacement patients and healthy patients[18]. Results have shown variation across selected points on the femoral head and between individuals for walking gait. However, little detail has been published with regards to variation between and within subjects, for different activities and for larger cohorts [13]. This is likely because the analysis can be extremely time consuming to organise data and analyse motion path variation in detail, particularly if motion paths are being calculated one trial at a time.

Visual3D (V3D) is largely regarded as the gold standard for the processing of gait data and may be an appropriate software to improve the current method for calculating motion paths. Raw gait data can be imported directly from motion capture software such as Qualisys (*Qualisys TM Medical AB, Goteborg, Sweden*) and Vicon (*Vicon Motion Systems, Oxford, England*). Large sets of motion data can be organised, processed and biomechanically analysed within the software. The integration of motion path analysis into V3D would provide a time saving technique that facilitates batch processing of the relative motion occurring at the hip during a range of activities.

The primary aim of this study was to determine whether the analysis of hip motion paths can be integrated into gait analysis software. Specific objectives were to: 1) Integrate motion path calculations within V3D by creating a virtual hip joint; 2) validate the V3D method (Virtual Joint motion path method) against previous methods; 3) utilise the new model for the analysis of cross shear for a range of activities 4) and to consider the potential for analysing both local and global wear using the new method for the application of both joint replacements and more delicate cartilage substitutional therapies.

2.0 Methods

2.1 Motion Capture Analysis

All subjects were recruited from staff and students at the University of Leeds. 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.

Validation of the virtual hip model was undertaken using 5 subjects. Three males and two females (Mean ±Standard Deviation; Age: 48 ±19 y; Height: 1.72 ± 0.1 m; Mass: 73 ±8 kg). For later application of the model a total of 18 subjects were recruited

(Mean ±Standard Deviation; Age: 44 ±19 y; Height: 1.7 ±0.1 m; Mass: 76.3 ±13.1 kg) (Table 1).

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

Subject demographics	
N	18
Sex (Male: Female)	10 Male 8 Female
Age Range	20 to 70
Age (Mean ±SD)	44 ±19
Weight Range (kg)	50.2 to 106.1
Weight (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

Validation was performed using two extremes of activity, Level Walking and Sitting Down (chair height 47 cm) representing the lower and higher extremes of expected motion path aspect ratios respectively.

Lab set-up

Twenty-eight 15.9 mm diameter retro reflective markers were attached to lower limb anatomical landmarks. Additionally, four semi-rigid thermoplastic shells, fitted with a total of sixteen tracking markers, were attached to the thigh and shank (Table 2) [19-20]. The 15 by 15 meter movement analysis laboratory allowed for the set-up of a thirteen-camera Qualisys Oqus 3-D motion capture system (*Qualisys TM Medical AB, Goteborg, Sweden*) and two force platforms (*AMTI, Advanced Mechanical Technology*

Inc., Watertown, MA, USA). Kinematic and kinetic data was synchronised and collected at 400 Hz and 1200 Hz, respectively.

Table 2. Location of external skin markers. Co-ordinate system 'A' refers to the anatomical markers and 'T' to the tracking markers. Those with both 'A' and 'T' were used to define both anatomical and technical co-ordinate systems. Markers were mirrored on the left and right side.

Marker	Co-ordinate System	Location
ASIS	A	Anterior superior iliac spine
PSIS	A	Posterior superior iliac spine
GT	A	Most lateral aspect of the femoral head (Greater trochanter)
THI1-4	T	Lateral aspect of the thigh
MKNE	A	Most medial projection of the medial femoral condyle
LKNE	A	Most lateral projection of the lateral femoral condyle
SHK1-4	T	Lateral aspect of the shank
MANK	A	Most medial projection of the medial malleolus
LANK	A	Most lateral projection of the lateral malleolus
MCAL	A, T	Medial aspect of the calcaneus
CAL	A, T	Aspect of the Achilles tendon insertion on the left calcaneus
LCAL	A, T	Lateral aspect of the calcaneus
MT1P	A, T	Most medial projection of the base of the first metatarsal head
MT5P	A, T	Most lateral projection of the base of the fifth metatarsal head
MT1D	A, T	Most medial projection of the head of the first metatarsal head
MT5D	A, T	Most lateral projection of the head of the fifth metatarsal head

Data collection

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 13 activities namely Walk, Walk Turn, Incline Walk, Decline Walk, Stand to Sit, Sit to Stand, Sit Cross Legged, Squat, Stand Reach, Kneel Reach, Lunge, Golf Swing, Cycling. Activities were chosen specifically to represent the movements that occur during common household activities.

Data processing

Kinematic markers were filtered at 10 Hz and body segments were modelled on Visual3D, as described in previous work [21]. Bell and Brand's predictive method was utilised to define the location of the hip joint centre [22-23]. It is important to appreciate that error will occur within all hip centre regression calculations, thus it is crucial to appreciate that alternative methods may yield various errors ranging from ∼15 to 35 mm [24]. Hip joint angles were defined through the orientation of the thigh segment in relation to the pelvis.

2.2 Virtual Joint motion paths model

The basic analytical capabilities of Visual3D (V3D) were utilised to allow a Virtual Joint to be constructed within the model and to allow the calculation of motion paths to be integrated into the program. Similar to previous methods, twenty points were defined to represent the hemisphere of a 28 mm diameter femoral head (*X*: anterior (+) posterior (-); *Y*: medial (-) lateral (+); *Z*: inferior (-) superior (+)) [13, 15]. Ten points (*X*, *Y*, *Z*) ran in an arc from posterior (0, -14, 0) to anterior (0, 14, 0) and ten points ran from medial (-14, 0, 0) to lateral (14, 0, 0). This was achieved by creating a hemisphere of equally spaced landmarks, relative to the thigh segment, around the hip joint centre (Figure 1). Angular motion of the thigh segment influenced the three dimensional displacement of each landmark. The motion of the twenty landmarks were then calculated relative to the pelvis coordinate system, using a transformation pipeline within V3D, in order to include pelvic tilt within the motion paths. Resulting data was subtracted from the position of the hip centre, therefore scaling the motion paths within the space of a 28 mm diameter hemisphere. A Virtual Joint motion paths model (MDH file) was thus created, meaning that this method could be simultaneously applied to any number of motion trials.

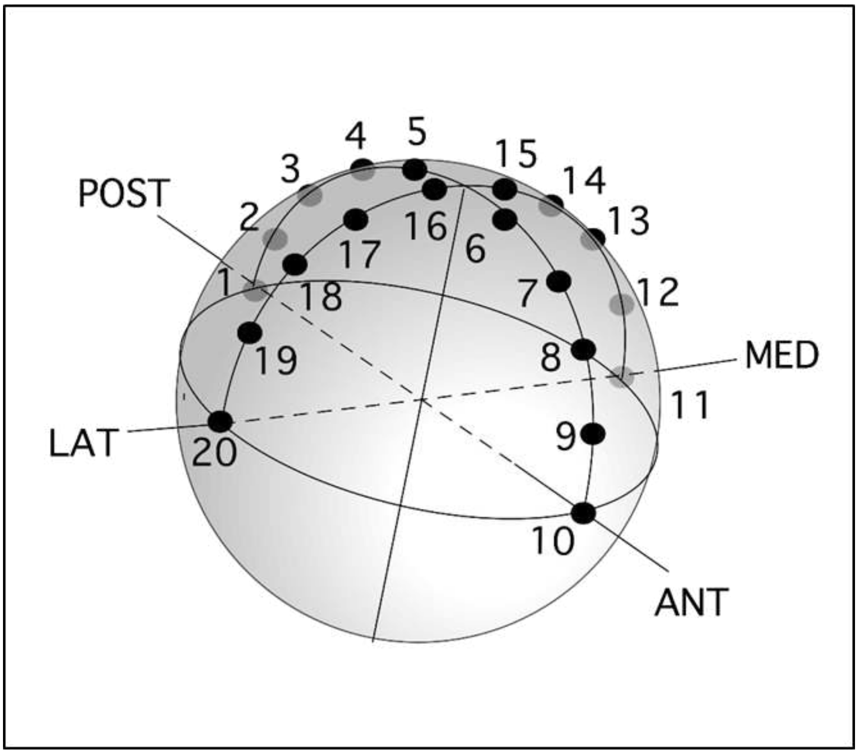

Figure 1. Virtual joint model construction.

Validation

The Virtual Joint motion paths method was validated against a computational model previously developed by Budenberg and colleagues in 2012 (*MATLAB, 2016, MathWorks, Natick, MA, USA*) [13]. Budenberg's method incorporated a number of matrices, alongside input values for one cycle of hip angular data [13,25]. Twenty points were defined for a 28 mm diameter femoral head. The same twenty points were used in V3D to allow for comparison [15,26].

Each point on the femoral head was defined and tracked in relation to the gait cycle. Although there are contrasting views in the literature, the study incorporated a Cardan sequence of rotations in which abduction/adduction is followed by internal/external rotation and finally flexion/extension [13]. Angular data was derived in this way, before inputting to the MATLAB program, to ensure that motion paths matched up to those calculated directly from V3D.

Rather than manually implementing the transformation matrix, as used previously in Budenberg's model, the displacement of points on the femoral head were calculated automatically within a V3D pipeline, directly from the motion file. Relevant motion files

(C3D) were imported to V3D, the MDH file was applied and motion path data was then exported for all trials.

In order to validate V3D results, two motion files were processed within a Matlab program and using the V3D virtual hip and the level of error was compared between motion paths for each of the twenty points. A number of variables were matched in order to validate the program, including: the position of points on the femoral head, the diameter of the femoral head and the coordinate system in which motion paths were calculated.

3. Results

3.1 Validation

When comparing the motion paths for the twenty femoral head points calculated from the past computational model, against the new Virtual Joint model (Visual3D) for walking and for rising from a chair the error and standard deviation was negligible in all cases. The Sliding distances that were predicted in both models were within < 0.01mm demonstrating that the same calculations were being replicated. Visual3D retains its significant figures within internal calculations. This is a potential benefit of using the Visual3D method as it is less likely to cause errors associated with data transfer. The suitability of Visual 3D was expected as it is essentially a mathematical model specially designed for analysis of motion in geometric shapes and is hence perfectly suited to the analysis of motion paths.

3.2 Application of the Virtual model to 13 activities.

The Visual 3D model can represent the activity of each subject in a skeletal format which is a useful feature to assist in visualising the data (Figure 2).

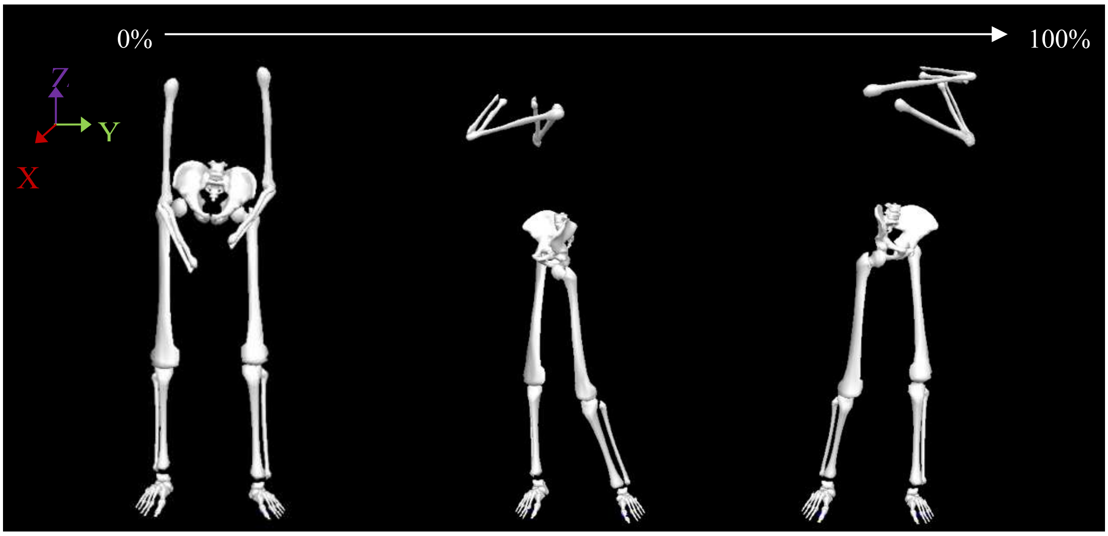

Figure 2. Visual3D model of a golf swing from start (Left) to end (Right); the person is rotating about their left hip. Under pure rotation the local motion path at the superior pole of the left femoral head will be a small sphere whereas the motion path at a lateral radius of the head will be a long linear arc. As these movements occur over the same time period this induces variations in velocity and acceleration across the joint surface.

The range of motion of the 13 activities is shown in Figure 3 along with the subsequent motion paths of each activity in Figure 4.

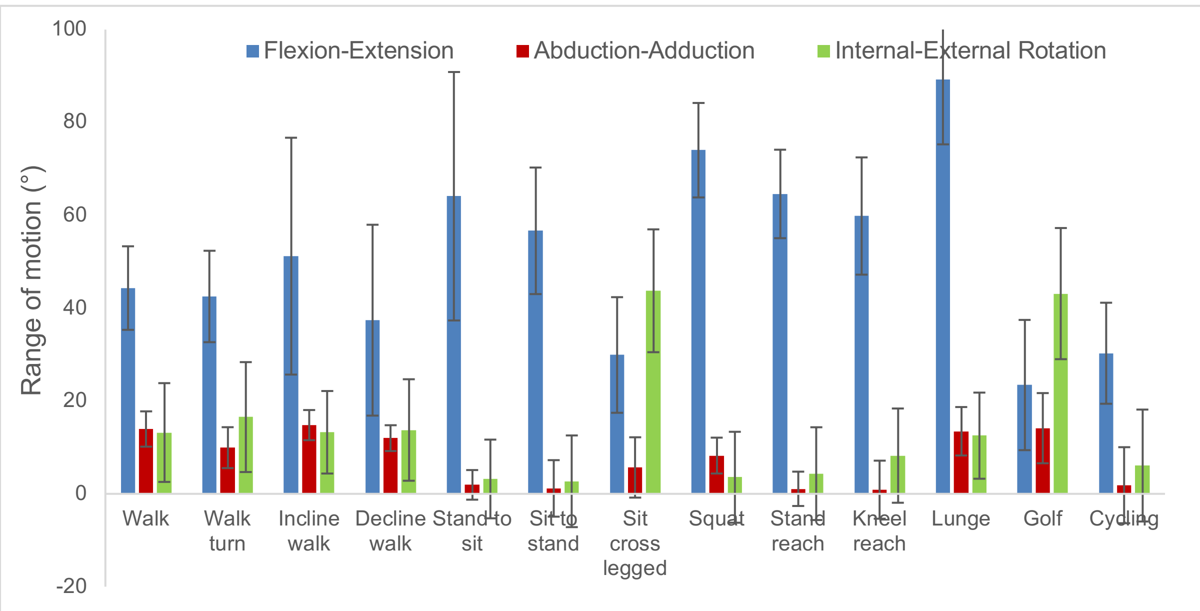

Figure 3. Average hip angular range of motion, in three axes, for thirteen common activities (n=18). Error Bars represent average standard deviation.

All versions of walking produced similar ranges of motion, whereas other activities involving squatting or sitting had much greater flexion. Internal and external rotation was noticeably greater in sitting cross legged and in playing golf. Interestingly the motion path for the walk turn had more of a helix pattern compared to other forms of walking, however the aspect ratio was comparable. Activities involving squatting or sitting had much more linear motion paths with aspect ratios 2-5 times greater than walking. The exception to this was the lunge which despite having greater flexion also had comparable levels of ab/adduction and rotation to walking and thus a lower aspect ratio. The motion paths for cycling and golf were reasonably linear.

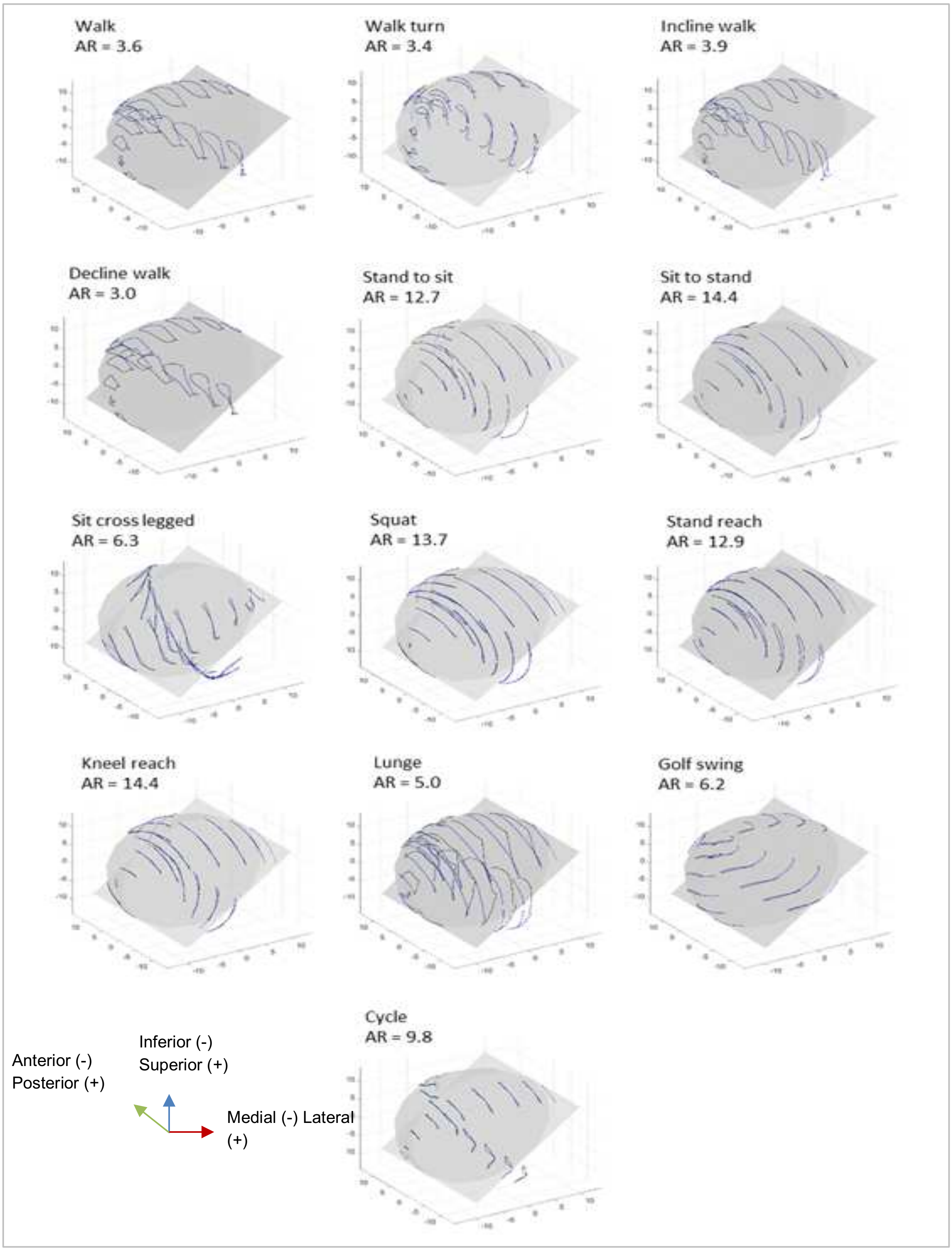

Figure 4. Mean motion paths for thirteen common activities. Mean aspect ratio (AR) (motion path height divided by perpendicular width) is shown above each individual graph.

A further advantage of the Visual 3D model is that the kinematics can also be determined within the model at any local area within the contact. The range of angular velocity at the superior pole of the femoral head under each activity is shown in Figure 5. This is compared to the levels produced in a typical hip joint simulator, in this case the Leeds Prosim [13]. Velocity levels were lower for the golf swing as this action contains more internal rotation, occurring about the superior pole of the head where there is less sliding distance. Hence, if a different point is considered the local velocity will change.

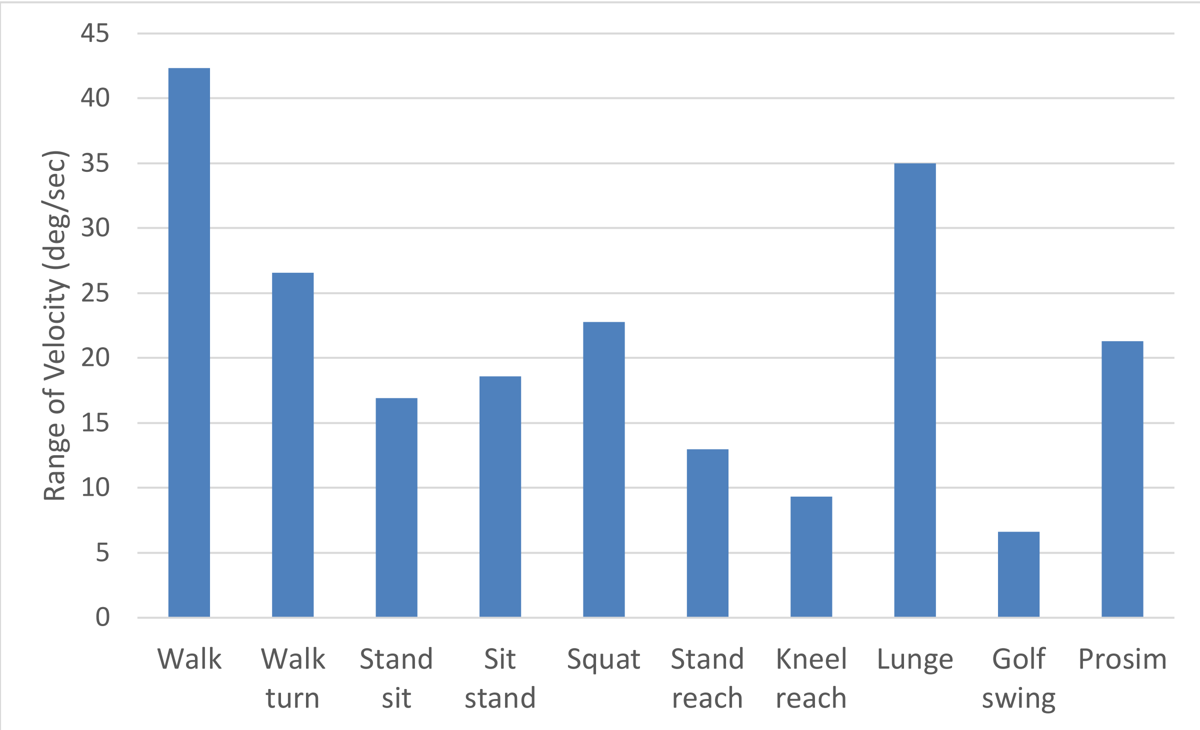

Figure 5. Range of hip velocity (degrees per second) for a single point on the superior pole for nine activities versus a hip simulator (Prosim).

More important perhaps for softer materials sensitive to shear is the acceleration of the relative surfaces, as shown in Figure 6. Surprisingly walking produced the greatest level of acceleration with the lunge being the next highest and the remainder of the activities being much lower. The range of acceleration of the simulator was found to be much lower than during the gait analysis assessment of walking despite the simulator being setup to represent an ISO walking cycle [18]. This is related to the physical limitations of the simulator that has large masses to accelerate and decelerate, hence the movements of the simulator are smoothed.

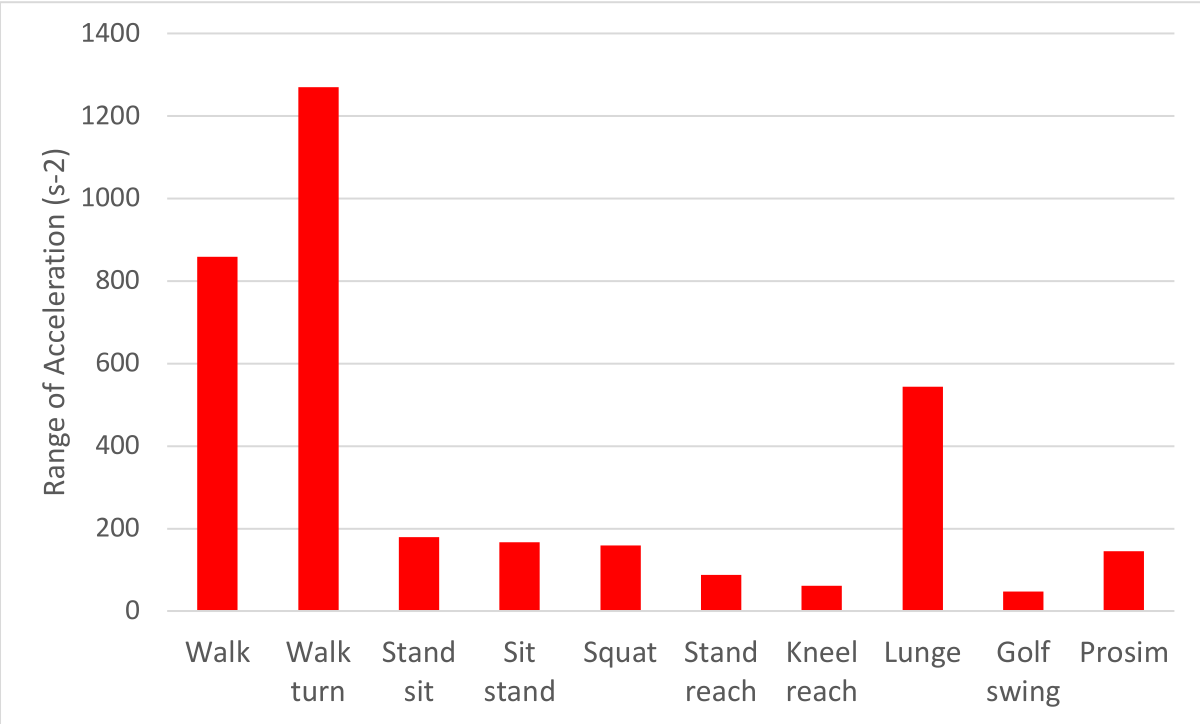

Figure 6. Range of sliding acceleration occurring across the superior pole surfaces at the hip for nine activities versus a hip simulator.

4. Discussion

Subjects were chosen to represent the lower age of the total hip replacement spectrum that generally return to a normal dynamic gait and activity following surgery and historically have poor success rates [1]. This age group also represents patients who may require soft tissue repair following a cartilage injury at a younger age, thus the study was focussed on patients that place the greatest demand on their joints to consider this group in comparison to the ISO hip replacement testing standard that was developed from a comparable cohort size [18].

The Virtual Joint motion paths model was able to replicate the motion paths produced in the previous computational model [13]. The Virtual Joint model also provides potential to easily recreate any patient specific positioning or implant design factors (diameter, head centre, femoral offset) and could therefore be used to analyse motion paths and wear implications from clinical data more effectively. Budenberg (2012) validated the original method against a simulator for walking [13]. However, it is important to appreciate that simulator motion paths will only match to computational work when the Carden sequences are matched and angular input angles are replicated.

The range of motion observed across the 13 activities (Figure 3) was substantial with an 89 degree variation in flexion, 15 degrees in abduction and 44 degrees in rotation making it challenging, but not impossible for a simulator to replicate. The variability over the 18 person cohort was very large in all of the activities, especially activities requiring deep flexion, this was expected perhaps with the age range of the cohort 20-70.

Figure 3 demonstrates the variation in motion paths between activities. Walking motion paths were generally quasi-elliptical and figure of 8 shapes, indicating a potential for high levels of cross-shear. This was in keeping with previous research [11, 13, 17]. Points lying on the most medial and lateral aspect of the femoral head showed the highest potential for cross-shear, due to their circular shapes, with 'complex tails'. Although the position of these points will change with alterations in the orientation of the femoral head in the acetabular cup. Further analysis into the variation of this position, between subjects and activities, may assist in understanding the key mechanisms for wear of polyethylene in a total hip replacement. The stand-to-sit activity showed linear patterns with 'complex tails'. The linear motion paths suggest that strain hardening may occur when sitting down. However, the 'complex tail' seen at the end of the sitting cycle may indicate a risk to high instantaneous cross-shear. It is important to acknowledge that the magnitude and direction of hip loading will contribute to the degree of wear occurring and must therefore be taken into account alongside motion path analysis [27]. The two movements (walking and stand-to-sit) are biomechanically very different, hence the difference in motion path aspect ratios.

The consideration of variable activities highlighted the variation of movement that occurs in walking, that is characterised by highly multidirectional sliding, versus activities that involve high flexion that have much more linear movements. Thus, whilst squatting (chair rise), is known to have greater reaction forces and thus potential for higher implant wear, squatting was found to have very linear motion paths and is perhaps less important than previously thought. The authors thus suggest that simulation of walking is perhaps the most important activity to replicate in testing as it has very multidirectional motion and variable load [28, 29]. In fact, as walking is the most common activity involving large amounts of movement, this is why it was chosen as the activity to replicate in the ISO pre-clinical test standard[18].

However, an interesting finding of the results is that the smoothing of kinematic inputs (to run efficiently on a hip simulator) alters the motion path trajectories but also more importantly dramatically reduces the local acceleration and thus the effectiveness of the hip simulator in replicating the harsh conditions of true walking [16,27]. For this reason, it is important to consider the local conditions occurring in activities and how these are replicated in simulator cycles. For polyethylene bearings this might not be as important as the average cross shear is comparable to walking. However, for softer surfaces like articular cartilage substitutes it is likely crucial to to ensure that the local acceleration/shear between the surfaces is replicated in pre-clinical assessment.

A benefit of the Virtual Joint motion model is the capability to visually navigate large data sets and batch process. For example, when motion analysis involves multiple individuals, activities and repeated trials. Traditionally these are averaged in order to predict representative motion, however this has the potential to lose important features within the motion (and motion paths). The Virtual Joint motion model allows all trials to be processed simultaneously. By processing large cohorts within V3D, further programming and organisation of input files within directories can be avoided. This provides a novel tool for processing large sets of cross-shear data quickly and easily, whilst avoiding the potential for error/ time when using multiple software. Additionally, motion paths can be viewed alongside the corresponding motion file and hip angular data, allowing the researcher to visualise the influence that specific hip angular patterns may have on motion paths. In a patient where an instrumented hip is utilised, whereby force data is recorded in-vivo, the modelling of a virtual joint will aid in the future for real-time wear modelling of implants [30, 31].

5. Conclusion

A novel Virtual Joint motion model was developed within Visual3D gait analysis software. The model facilitates the production of joint surface motion path calculations and thus provides a more holistic view of the influence of body movement/ activity on implant wear. Of 13 activities assessed using the model walking was confirmed to be an excellent activity for wear assessment due to its complex multidirectional motion paths. However, when considering simulation of movements for wear assessment the details within the motion path such as the localised acceleration/shear between joint surfaces may be more important than the global shape of the path itself.

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