Preferential superior surface motion in wear simulations of the Charité total disc replacement

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Abstract Laboratory wear simulations of the dual-bearing surface Charité total disc replacement (TDR) are complicated by the non-specificity of the device's center of rotation (CoR). Previous studies have suggested that articulation of the Charité preferentially occurs at the superior-bearing surface, although it is not clear how sensitive this phenomenon is to lubrication conditions or CoR location. In this study, a computational wear model is used to study the articulation kinematics and wear of the Charité TDR. Implant wear was found to be insensitive to the CoR location, although seemingly non-physiologic endplate motion can result. Articulation and wear were biased significantly to the superior-bearing surface, even in the presence of significant perturbations of loading and friction. The computational wear model provides novel insight into the mechanics and wear of the Charité TDR, allowing for better interpretation of in vivo results, and giving useful insight for designing future laboratory physical tests.

Keywords Wear · Total disc replacement · Charité · Lumbar · Dual-bearing surface

Introduction

Motion-preserving intervertebral total disc replacements (TDRs) are intended to reduce the pain of degenerative disc disease, while avoiding the adjacent level degeneration associated with spinal fusions [3, 7]. Although TDR designs have benefited significantly from advances made in 40 years of experience with hip and knee implants, polyethylene wear is a concern, due to the tribology of the bearing surfaces involved, and the proximity of the implant to the spinal cord. There is evidence supporting the potential for osteolytic reaction to wear debris [6, 20] in the spinal environment, and revision surgery, if necessary, poses a major risk due to the abdominal great vessels. To date, there have been few long-term clinical wear data available, due to the relatively short time that TDRs have been used. Furthermore, wide variation in implant designs makes it challenging to draw broad inferences about clinical performance.

Laboratory simulator tests and, more recently, computational wear simulations, have therefore been increasingly used to predict the long-term wear behavior of TDRs [5, 21, 23]. Consensus standards and a guidance document for wear testing of TDRs have been developed by the ISO (18192-1) and ASTM (F2423-05), respectively. However, for certain implant designs, implementing these standards in a consistent manner is complicated by the potential for a non-specific center of rotation (CoR). In laboratory simulators, the location at which an implant is mounted relative to the angular actuators typically enforces a CoR. While this is straightforward for a fully constrained, single-bearing surface design like the ProDisc-L, it becomes a problem when setting up an unconstrained, dual-bearing surface device such as the Charité. The Charité was designed such that its CoR can move as the implant articulates, much as happens in the natural functional

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spinal unit [12]. Anecdotal evidence both in wear simulators and in cadaver studies, as well as recent retrieval evidence [11], suggests that the Charité is prone to articulating and wearing preferentially on the superior-bearing surface, despite the device's geometry being fully symmetric around the mid-plane [17].

These observations raise questions about how to most appropriately set up Charité physical wear simulators to accurately replicate in vivo motion and wear. The cost and time to run wear tests is prohibitive for conducting extensive permutations of test equipment setup and loading; however, computational wear modeling provides an inexpensive and expeditious alternative. Computational wear models were first developed by Maxian et al. [14, 15] for wear analysis of hip implants. That paradigm has subsequently been expanded and applied to knee [25], shoulder [8], and spinal disc replacements [21], as well as to non-medical applications [19]. Recently, the effect of cross-shear (i.e., sliding motion running counter to the prevailing molecular orientation of the polyethylene) has been incorporated into a wear simulation of the ProDisc TDR [4]. In the present study, a computational wear model is used to investigate the kinematical and wear behavior of the Charité TDR, with respect to changes in the imposed CoR. Further computational testing was performed in which the coefficient of friction at the inferior-bearing surface was lowered, to determine if the preferential articulation of the superior surface could be influenced by lubrication perturbations. Finally, a conceptual mechanical explanation for the preferentially superior surface motion is posited.

Methods

A finite element (FE) model of a Charité TDR was implemented in Abaqus (Fig. 1). The ultra-high molecular weight polyethylene (UHMWPE) core was modeled as a deformable solid and the endplates as rigid surfaces. Forces and displacements were applied to the endplates through reference points initially located at the each endplate's respective center of curvature. Contact was defined between the UHMWPE core and the endplates, with a (baseline) Coulombic friction coefficient of 0.08, representative of CoCr-on-UHMWPE bearing surfaces [18]. Additional details of the FE model and a discussion of limitations can be found in "Appendix".

Wear was incorporated into the FE model through use of the UMESHMOTION subroutine in Abaqus. In this procedure, as in previous studies [4, 13, 14], the relative sliding kinematics and contact pressure of each node against the endplate were tracked for one motion cycle. At the end of the motion cycle, the linear wear depth at each

Fig. 1 **a** CAD and **b** finite element models of Charité TDR

node was calculated based on the Archard/Lancaster [1] wear law, augmented to include cross-shear

$$w = \sum_{i=1}^n k(\text{CSR}) \cdot \sigma_i \cdot |\bar{u}_i|, \quad (1)$$

where w is the local linear wear depth, k and CSR are the instantaneous wear coefficient and cross-shear ratio, σ_i and μ_i are the local contact pressure and sliding distance vector for motion increment i , and n is the total number of increments in the duty cycle. The local linear wear depths were scaled by an update interval of 250,000 cycles, with each node moved accordingly in a direction normal to the surface. Automatic adaptive meshing routines adjusted the internal node positions to preserve computationally well-behaved element geometry. The wear simulations were run to correspond to 10 million cycles physically, involving 40 mesh updates per overall simulation.

CSR was calculated per Kang et al. [9]. Briefly, UHMWPE molecular chains were considered to preferentially align in the direction of the dominant frictional work, the axis of such alignment being designated as the principal molecular orientation (PMO). A CSR was then calculated as the ratio of the frictional work performed in a direction perpendicular to the PMO ($W_{\perp}$) to the total frictional work (W_f)

$$\text{CSR} = \frac{\sum_{i=1}^n W_{\perp i}}{\sum_{i=1}^n W_{f i}} \quad (2)$$

The relationship between the wear coefficient and the CSR was based on the pin-on-plate results of Kang et al. [21] scaled to return wear rates characteristic of full-implant testing (necessitated by the tribologic differences between pin-on-plate and full-implant wear tests), and with a small offset added to replicate the value of $1 \times 10^{-8} \text{ mm}^3/\text{N/m}$ measured by Saikko and Ahlroos [22] for purely reciprocal linear motion

$$k(\text{CSR}) = 2.8 \times 10^{-6} (\text{CSR} + 1.53 \times 10^{-12})^{0.2071}. \quad (3)$$

The input specified by ISO Standard 18192 for lumbar TDR wear testing (Fig. 2) was applied to the implant.

The reference points for these inputs were initially located 12.25 mm above and below the polyethylene core's center, at the center of curvature of the respective endplates. Additional wear simulations were then performed with the reference points moved to positions 8.25, 4.25, and 0 mm away from the implant midline. The location of the superior endplate reference point effectively enforces the

implant's CoR. The 10° shear angle in the ISO Standard was not included.

Preliminary studies with this model, as well as recent observations in cadaveric and retrieval studies [11, 17], have suggested that articulation of the Charité primarily occurs at the superior-bearing surface. To test the sensitivity of this phenomenon to equipment setup, a linear wear balance parameter, defined as the ratio of upper to lower-bearing surface peak linear wear, was calculated for each of the four alternative choices of reference points. In an additional set of simulations testing the sensitivity of the wear balance to lubrication conditions, the coefficient of friction of the inferior-bearing surface was varied from 0.01 to 0.1, while the superior coefficient of friction was held at 0.08. Again, the total wear and the relative wear between bearing surfaces were evaluated.

Results

Regardless of the imposed CoR, initially all articulation of the Charité occurred in the upper-bearing surface (Fig. 3a). Over the course of the simulation, some articulation and wear did eventually occur at the lower-bearing surface,

Fig. 2 Lumbar TDR loading conditions prescribed by ISO 18192-1

Fig. 3 a Total sliding distances of a representative UHMWPE node against the endplates for the superior and inferior-bearing couples, for one duty cycle. **b** Representative cumulative linear wear distributions, after 10 million cycles, from the case with the center of rotation at the centers of curvature of the superior endplate

Fig. 4 **a** Comparison of gross implant movement with center of rotation located at the center of implant (*top*) and at the superior endplate center of curvature (*bottom*). Note the translation of both the implant core and of the inferior endplate in the first case. **b** Comparison of corresponding bearing surface motion loci near the pole

although superior surface articulation still predominated. The linear wear distributions (Fig. 3b) were essentially unchanged throughout CoR perturbations, as were total volumetric wear and wear rate. Additional investigation of implant motion revealed that changing the endplate reference point forced a change in the global motion of the implant (Fig. 4a). However, the degrees of freedom left unconstrained allowed the endplates to compensate such that the local sliding kinematics remained unchanged, as demonstrated by the minimal change of relative surface motion loci (Fig. 4b). The total linear wear rate computed in these simulations was approximately 0.1 mm/year, based on 1 million cycles per year. This is within the range reported by Kurtz et al. [11] for retrievals, although toward the high end of that range. The ratio of the peak superior linear wear depth versus the peak inferior linear wear depth was 1.92.

Changes to the inferior-surface coefficient of friction resulted in non-negligible changes to the Charité wear rate. However, until the lower friction coefficient was very substantially lowered, predominant articulation and wear continued to occur in the superior-bearing couple (Fig. 5). Even with the inferior coefficient reduced to 0.01 (i.e., only 12.5% of that of the superior surface), considerable articulation and wear still occurred at the superior-bearing surface. Figure 6 shows the variation of the sliding distance per duty cycle of a point on each bearing surface, as the inferior coefficient of friction was reduced. Points on the inferior-bearing surface did not begin to experience longer sliding distances than on the superior surface until the inferior coefficient of friction was lowered to less than 60% of the superior surface coefficient of friction.

Discussion

Due to relatively short clinical experience, contemporary designs of TDRs have many unknowns as to in vivo

Fig. 5 Change in peak linear wear and ratio of peak superior to peak inferior linear wear as the inferior coefficient of friction was changed

Fig. 6 Change in total sliding distance of a single point through one motion cycle, as the inferior coefficient of friction is changed. Vertical dotted line indicates symmetric wear coefficients

Fig. 7 Conceptual mechanical explanation for preferential superior articulation of Charité TDR. **a** Inferior articulation requires translation of the inferior endplate. **b** Key forces relevant to implant when articulation is in superior surfaces. **c** Translation of the inferior endplate introduces an additional moment, and requires a lower frictional force generated in inferior-bearing surface to maintain static equilibrium (details in “Appendix”)

function and reliability, and especially as regards long-term clinical performance. The Charité implant is of particular interest because its unconstrained articulation makes it very difficult to predict motion in any given configuration. It is, therefore, difficult to ascertain how well ex vivo tests truly replicate in vivo conditions. Since laboratory wear testing is expensive and time consuming, computational simulations can play a useful role in understanding the mechanics of the design, and the sensitivity of experimental wear results to the test protocol or equipment setup.

Existing literature on wear of the Charité lacks consensus. Initial experimental studies conducted under designer/manufacturer auspices reported that the design had negligible wear propensity [12, 23]. Another experimental study, by Nechtow et al. [16], showed that the Charité could have either very little wear, or significant wear, depending on the input parameters used. Isolated clinical case studies have demonstrated that it is possible for an implanted Charité to function for many years without significant wear [2, 10], duly recognizing the cost and complication risks of revising the implant, should a problem occur. More recently, however, clinical retrieval studies have shown that measurable wear does indeed occur in the Charité [11], and that subsequent wear particle generation invokes a response in periprosthetic tissue [20], although in ongoing clinical series, osteolysis incidence remains low.

The results of the present study support the possibility of appreciable Charité wear in vivo, and provide novel insight into the sensitivity of such wear to factors involved in replicating in vivo conditions experimentally. Deliberate perturbations of the implant CoR revealed that the articulation kinematics, the overall wear rate and the local wear depth distributions were remarkably insensitive to perturbations of the location of the implant relative to the angular actuators. This finding is encouraging, in terms of reducing

the technical challenges for wear testing, in that it suggests that wear behavior is fairly forgiving of experimental test protocol variations. Wear was found to occur predominantly on the superior surface of the UHMWPE core. According to the classification (for retrievals) used by Kurtz et al. [11], the symmetry ratios found in this study (around 0.9) qualify as one-sided wear. As a more direct measure, the results in the present study point to nearly two times greater wear depth on the superior side when the superior and inferior coefficients of friction are equal, regardless of endplate reference point location.

The asymmetry of computed wear was found to be persistent, even when the inferior coefficient of friction was lowered so as to encourage inferior articulation. (This scenario would obviously be difficult to conceive of occurring physically, either in laboratory or in vivo settings, but it is useful for purposes of mechanistic exploration.) Despite reducing the inferior coefficient of friction by over 85%, a substantial amount of the total implant articulation and resulting wear was still taken up at the superior-bearing couple. A simplified, two-dimensional analysis of flexion–extension is helpful as a thought experiment toward a conceptual mechanistic explanation. As illustrated in Fig. 7a, in order for the inferior surface to articulate in the setup used for the present study, the endplates must translate relative to one another. The free-body diagrams of relevant forces in Fig. 7b, c show that such a translation results in an additional moment term not present when articulation occurs purely at the superior-bearing surface. Further analysis (“Appendix”) shows that this additional moment cannot be balanced by the friction in the superior-bearing surfaces unless the coefficient of friction is substantially higher than the inferior surface coefficient of friction. Thus, at least under the conditions typical of laboratory testing, the Charité TDR is kinetically predisposed to articulate and wear preferentially on the superior surface.

The implications of superior-only articulation in vivo are unclear. Kurtz et al. posited that single-surface articulation implied that the CoR of the implant was well below the disc space, in contrast to the situation for the natural intervertebral disc. In this study, we found that even with the CoR constrained to be at the center of the implant, there is still a propensity for superior surface articulation, albeit with a relative translation of the endplates that seems unlikely in vivo. Therefore, there apparently is a disconnect between (a) the intended function of the device, (b) the behavior of the device in laboratory simulators, and (c) the actual function of the device in vivo. Reconciling these variations in behavior and, in particular, designing laboratory biomechanical tests that are representative of in vivo conditions, is critical to being able to predict the long-term function and outcome of the Charité, as well as any other TDR implant.

The results found in this computational study are consistent with recent motion locus recordings of a physical setup closely replicating the computational model [24], in which articulation was found to occur virtually exclusively in the superior-bearing surfaces. Video analysis of another Charité experiment found that the range of angular displacement in the inferior surface was 25% of the total implant flexion/extension range, while the FE model returned a similarly low inferior articulation, up to 10% of the total range. Experimentally, adding an anterior–posterior force invoked significant inferior articulation. Again, corresponding FE tests found similar responses.

As clinical experience with TDRs continues and as further design variations are introduced, it becomes increasingly important to understand the mechanics and long-term wear behavior of the implants. In the case of the Charité, the dual-bearing surfaces provide additional flexibility in accommodating motion in vivo, but complicate in vitro test setup. Although Charité wear is insensitive to the prescribed CoR in vitro, it is unclear whether the corresponding differences in relative translations of the endplates would be physiologically feasible in vivo. The persistent preferential superior surface wear, with corroboration by some retrieval observations, indicates that the Charité may sometimes perform similarly to a single-bearing surface design. Further understanding of the in vivo motion of the Charité, and determining how precisely such motion needs to be replicated in in vitro testing, is vital to reliably predicting the implant's long-term function.

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Conflict of interest statement None of the authors has any potential conflict of interest.

Appendix

FEA details

The FE model of the Charité (size 4, 28.5 mm diameter, 9.5 mm thickness) was created in Abaqus with the endplates modeled as analytical rigid surfaces and the UHMWPE core modeled as a deformable solid. The UHMWPE core was assigned linear elastic material properties with an elastic modulus of 1,400 MPa and a Poisson ratio of 0.3, representative of oxidized UHMWPE. The core was meshed with 4,014 linear hexahedral elements. Abaqus automatically generates contact elements on the faces of elements designated as part of a contact pair. In this case, the contact elements had edge lengths of approximately 0.8 mm. Contact was implemented through a penalty formulation and surface-to-surface discretization was used.

Loads were applied to replicate a simulator recently developed at the University of Leeds [24], in which axial rotation and axial load are applied to the inferior endplate, and flexion–extension and lateral-bending are applied to the superior endplate. Figure 8 shows the loads applied both to the physical simulator and the FE model. Note that in the FE model, the reference point *below* the implant controlled the upper endplate, and the reference point *above* the implant controlled the lower endplate.

As with all FE analyses, this model has certain inherent limitations. The material model was purely linear elastic, and thus did not simulate creep or any type of plastic deformation. In addition, the FE setup required that a CoR be prescribed, by the location of the reference point for the endplates. This is not necessarily a limitation, as it is very analogous to the setup of a laboratory simulator, but the reader should be aware that in vivo loading is somewhat more indirect, subject to muscle forces and moments acting from various interacting anatomic structures. Finally, the realism of the wear process depends on the accuracy of the input data. In this study, the wear factor was modulated to depend on cross-shear, based on the data of Kang et al. This wear formulation had been validated in laboratory tests of other TDR designs. In addition, the linear wear depths found in this study were within the range reported by Kurtz et al. for retrieval specimens, leading us to conclude that the critical wear behavior was captured here.

Fig. 8 Rotational and loading input to **a** the physical simulator, and **b** the finite element model. As indicated by the *dotted lines*, the lower reference point in the finite element model controls the upper endplate, and vice versa

Mechanics analysis

A static equilibrium mechanical analysis of a simplified two-dimensional Charité model can be performed by summing the moments acting on the polyethylene core. Figure 9 shows the relevant forces, vectors, and angles used in the analysis. dN and dF are differential normal and frictional forces, respectively, resulting from the contact pressure, P , at a given point. R is the position vector from the center of curvature of the endplate to the contact point of interest, which can be represented in polar form as $\{R, \theta\}$. ds is the position vector from the polyethylene core center to the contact point, and h is half of the maximum thickness of the core. γ is the angle, from horizontal, to the axis between the endplate centers of curvature, changes of which essentially correspond to changes of the flexion–extension angle. The subscripts s and i denote variables relating to the superior and inferior endplates, respectively.

The contact pressure, P_s , at any point along the superior surface of the polyethylene core results in differential normal and frictional shear forces, which can be calculated (in rectilinear coordinates) as,

$$d\bar{N}_s = P_s \cdot R \cdot d\theta_s \cdot [-\cos(\theta_s)\hat{i} - \sin(\theta_s)\hat{j}], \quad (4)$$

and,

$$d\bar{F}_s = P_s \cdot R \cdot d\theta_s \cdot \mu_s \cdot [-\sin(\theta_s)\hat{i} + \cos(\theta_s)\hat{j}], \quad (5)$$

respectively. To sum the moments about the center of the core resulting from these forces, the position vector from the core center to the point of application of the differential forces is calculated as

$$d\bar{s}_s = [R \cdot \cos(\theta_s) - (R - h) \cdot \cos(\gamma)]\hat{i} + [R \cdot \sin(\theta_s) - (R - h) \cdot \sin(\gamma)]\hat{j}. \quad (6)$$

The corresponding differential moments can then be calculated by cross-product

$$d\bar{M}N_s = d\bar{s}_s \times d\bar{N}_s \quad (7)$$

$$d\bar{M}N_s = P_s \cdot R \cdot (R - h) \cdot \sin(\theta_s - \gamma) \cdot d\theta_s \hat{k} \quad (8)$$

$$d\bar{M}F_s = d\bar{s}_s \times d\bar{F}_s \quad (9)$$

$$d\bar{M}N_s = P_s \cdot \mu_s \cdot [R^2 - R \cdot (R - h) \cdot \cos(\theta_s - \gamma)] \cdot d\theta_s \hat{k} \quad (10)$$

The forces and moments about the core center caused by the contact pressure on the inferior surface can be similarly calculated, this time of course based on a coordinate system attached to the inferior endplate reference point

$$d\bar{N}_i = P_i \cdot R \cdot d\theta_i \cdot [-\cos(\theta_i)\hat{i} - \sin(\theta_i)\hat{j}] \quad (11)$$

$$d\bar{F}_i = P_i \cdot R \cdot d\theta_i \cdot \mu_i \cdot [\sin(\theta_i)\hat{i} - \cos(\theta_i)\hat{j}] \quad (12)$$

$$d\bar{s}_i = [R \cdot \cos(\theta_i) + (R - h) \cdot \cos(\gamma)]\hat{i} + [R \cdot \sin(\theta_i) - (R - h) \cdot \sin(\gamma)]\hat{j} \quad (13)$$

$$d\bar{M}N_i = d\bar{s}_i \times d\bar{N}_i \quad (14)$$

$$d\bar{M}N_i = P_i \cdot R \cdot (R - h) \cdot \sin(\gamma - \theta_i) \cdot d\theta_i \hat{k} \quad (15)$$

$$d\bar{M}F_i = d\bar{s}_i \times d\bar{F}_i \quad (16)$$

$$d\bar{M}F_i = P_i \cdot \mu_i \cdot [-R^2 - R \cdot (R - h) \cdot \cos(\theta_i - \gamma)] \cdot d\theta_i \hat{k} \quad (17)$$

The differential moments thus calculated can then be integrated between the starting and ending points of the contact patch. For simplicity, the contact pressure is assumed to be constant across the contact patch, to

Fig. 9 Differential forces, position vectors, and angles necessary for computing moments about the polyethylene core center for both the **a** superior, and **b** inferior-bearing surfaces

calculate the whole-bearing surface moments. The resulting total moments are, for the normal force contribution on the superior surface,

$$MN_s = \int_{\theta_{s1}}^{\theta_{s2}} d\bar{M}N_s = P \cdot R \cdot (R - h) \cdot [-\cos(\theta_{s2} - \gamma) + \cos(\theta_{s1} - \gamma)] \Rightarrow A, \quad (18)$$

for the frictional forces on the superior surface,

$$MN_s = \int_{\theta_{s1}}^{\theta_{s2}} d\bar{M}F_s = P \cdot \mu_s \cdot [R^2 \cdot (\theta_{s2} - \theta_{s1}) - R \cdot (R - h) \cdot [\sin(\theta_{s2} - \gamma) - \sin(\theta_{s1} - \gamma)]] \Rightarrow B \cdot \mu_s, \quad (19)$$

for the normal force on the inferior surface,

$$MN_i = \int_{\theta_{i1}}^{\theta_{i2}} d\bar{M}N_i = P \cdot R \cdot (R - h) \cdot [\cos(\gamma - \theta_{i2}) - \cos(\gamma - \theta_{i1})] \Rightarrow C, \quad (20)$$

and for the frictional force on the inferior surface,

$$MF_i = \int_{\theta_{i1}}^{\theta_{i2}} d\bar{M}F_i = P \cdot \mu_i \cdot [-R^2 \cdot (\theta_{i2} - \theta_{i1}) - R \cdot (R - h) \cdot [\sin(\theta_{i2} - \gamma) - \sin(\theta_{i1} - \gamma)]] \Rightarrow D \cdot \mu_i \quad (21)$$

Fig. 10 Change in ratio of μ_s to μ_i necessary to balance moment actions on polyethylene core

Static equilibrium of the polyethylene core requires that these moments sum to zero. (Despite being evaluated in different coordinate systems, the moments are all calculated about a common point and result in moment vectors about the k axis (perpendicular to the plane of analysis), so they can be summed together). Using this fact, and for brevity using the nomenclature substitutions in (18)–(21), the relationship between the inferior coefficient of friction and the necessary superior coefficient of friction can be written as

$$\sum M = 0 \Rightarrow A + B \cdot \mu_s + C + D \cdot \mu_i = 0 \Rightarrow \mu_s = \frac{-(A + C + D \cdot \mu_i)}{B}. \quad (22)$$

In Fig. 10, μ_i is held constant at 0.08, while γ is changed from 90° to 110°. The figure demonstrates that unless the endplates remain directly above one another (i.e., γ equal to 90°), the superior coefficient of friction must be larger than the inferior coefficient of friction in order to maintain static equilibrium while articulating along the inferior surfaces. Since a normally functioning implant would have equal coefficients of friction in both surfaces, this means the superior surface must preferentially articulate to satisfy static equilibrium.

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