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Periprosthetic fracture fixation of the femur following total hip arthroplasty: A review of biomechanical testing

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ABSTRACT

Background: Periprosthetic femoral fracture can occur following total hip arthroplasty. Fixation of these fractures are challenging due to the combination of fractured bone with an existing prosthesis. There are several clinical studies reporting the failure of fixation methods used for these fractures, highlighting the importance of further biomechanical studies in this area.
Methods: The current literature on biomechanical models of periprosthetic femoral fracture fixation is reviewed. The methodologies involved in the experimental and computational studies of this fixation are described and compared.
Findings: Areas which require further investigation are highlighted and the potential use of finite element analysis as a computational tool to test the current fixation methods is addressed.
Interpretation: Biomechanical models have huge potential to assess the effectiveness of different fixation methods. Experimental *in vitro* models have been used to mimic periprosthetic femoral fracture fixation however, the numbers of measurements that are possible in these studies are relatively limited due to the cost and data acquisition constraints. Computer modelling and in particular finite element analysis is a complimentary method that could be used to examine existing protocols for the treatment of periprosthetic femoral fracture and, potentially, find optimum fixation methods for specific fracture types.

Contents

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1. Introduction

Periprosthetic femoral fracture (PFF) is a complication associated with total hip arthroplasty (THA). A concerning increase in the incidence of this condition has been predicted, in line with the increasing number of THA operations (Learmonth et al., 2007; Lindahl et al., 2006; Tsiridis et al., 2009) . PFF can occur intra-operatively or post-operatively, creating a variety of fracture configurations and locations. There are various associated risk factors including the age of the patient (Wu et al., 1999), osteoporosis (Lou et al., 2007), the prosthesis design (Garellick et al., 1999; Toni et al., 1994), and whether a cemented or un-cemented prosthesis is used (Berry, 1999; Schwartz et al., 1989), each of which also influences the method of fixation (Franklin and Malchau, 2007; Lindahl et al., 2006).

A large number of clinical studies have investigated fracture following THA (Garcia-Cimbrelo et al., 1992; Kavanagh, 1992; Löwenhielm et al., 1989; Tsiridis et al., 2004). These have led many authors to classify PFF based on fracture configuration, position along the femur and associated bone quality (Duncan and Masri, 1995; Johansson et al., 1981; Mont and Maar, 1994). One of the most commonly used systems is known as the Vancouver classification (Duncan and Masri, 1995). Here, fractures located within the trochanter region are classified as type A. Fractures located within the stem region are classified as type B, with subsets representing those with a stable implant (B1), a loose implant (B2) and a loose implant plus insufficient bone stock (B3). Finally fractures positioned distal to the stem are classified as type C. Among these, type B fractures represent approximately 80% of all cases (Corten et al., 2009; Lindahl et al., 2006), and these have been the focus of several clinical and laboratory studies.

Treatment of PFF is challenging due to the combination of the fractured bone and the existing prosthesis, with the further complication, in some cases, of the cement used for prosthesis fixation. This creates a different biomechanical scenario to the fracture of an intact femur. Traditional treatment methods such as traction and bracing have been replaced by open reduction and internal fixation, along with revision of the prosthesis in some cases (Lindahl et al., 2006). Many authors have reported the clinical outcomes of different fixation methods (Bryant et al., 2009; Buttaro et al., 2007; Haddad et al., 2000; Partridge and Evans, 1982; Serocki et al., 1992; Tsiridis et al., 2003, 2005). Among these, there are several studies that report the failure of the fixation for these fractures (Fig. 1), indicating that the protocol for classifying PFF cases and subsequent selection of the fixation method are perhaps currently insufficient. Several authors have proposed treatment algorithms for different types of fracture (Masri et al., 2004; Parvizi et al., 2004) in light of available fixation techniques. However, these proposals lack any significant biomechanical evidence.

Biomechanical *in vitro* studies and computer *in silico* models have the potential to assess and optimise the performance of different methods of fixation. These techniques allow certain aspects of the *in vivo* conditions to be replicated in a controlled manner so that the biomechanical effects of various parameters can be assessed both individually and in combination. These types of study have been implemented extensively in the area of THA (Crowninshield et al., 1980; Huiskes, 1980; Prendergast and Taylor, 1990) and there is a growing body of work focusing on fracture fixation (Chen et al., 2010; Krishna et al., 2008; Perren, 1991; Stoffel et al., 2003). A number of biomechanical studies have also investigated PFF, although as yet, not all types of these fractures or fixation methods have been comprehensively evaluated.

The aim of this review is to examine the available literature relating to the biomechanical assessment of PFF. Current experimental and computational methodologies are evaluated and the trends in the results, as well as areas of disagreement, are highlighted. Recommendations for future research and areas which require further scientific investigation are discussed.

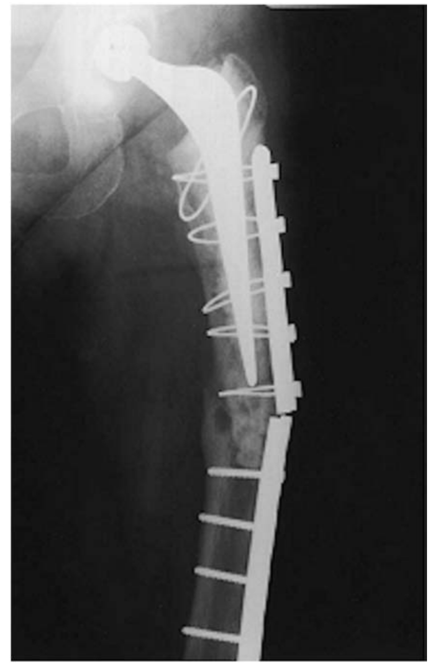

Fig. 1. Anteroposterior radiograph showing the failure of the periprosthetic femoral fracture fixation methods. Dall-Miles plate fracture from Tsiridis et al. (2003) with permission from Elsevier.

2. Methodology

2.1. Experimental methods

2.1.1. Introduction

Experimental *in vitro* studies have been used to assess the biomechanical stability of various methods of PFF fixation. These studies compare the mechanical performance of different fixation methods in the laboratory by stabilising a periprosthetic fracture in a cadaveric or synthetic femur. The following section reviews the methods used and examines their robustness, with a focus on three specific aspects: the type of specimen, the loading protocol and the methods of measurement.

2.1.2. Specimen type and repeatability

Since the majority of studies have made comparisons between different fixation methods, it is necessary for there to be parity between the specimens used so that any differences in outcome can be attributed to the fixation method alone. Both cadaveric and synthetic samples have been used, as is summarised in Table 1.

Cadaveric specimens more closely represent the '*in vivo*' material but there is inherent variability between samples, including, crucially, the bone quality and geometry as well as the potential presence of pre-existing damage in the bone. This variance would likely necessitate large sample sizes to obtain statistically significant results, but this is usually impractical due to the availability of the tissue and, from Table 1, it can be seen that most sample sizes here are small. To overcome this, several authors have adopted an approach of undertaking a number of different procedures sequentially on each specimen, using a randomised ordering method to reduce the effect of any cumulative damage to the specimens. Whilst such an approach is perhaps the best that can be achieved with limited specimens, little work seems to have been undertaken to assess the effect of these repeated tests. Haddad et al. (2003) used a standard fixation method as a control for each specimen, which was retested within each variable group. Whilst this enabled results to be compared to the most recent standard, the variation in the measures of the standard during the tests was not reported.

Table 1
A summary of the specimen preparation and loading protocol in laboratory studies. Note that the isometric loading category was assigned to any study that has loaded the construct 1) under axial loading (including only the adduction angle) or 2) in pure torsion or bending. Any other loading scenario, such as those that included muscle forces or positioned the femur in flexion and adduction, was categorized as physiological loading.

Authors	Specimen number and type	Prosthesis	Fracture	Loading	Femur position
Panjabi et al. (1985)	8 Cadaveric	Cemented ATS Howmedica, NJ Cemented STH Zimmer, IN	Drill hole and reaming defect 90 mm below lesser trochanter	Isometric Axial compression to 417-500 N depending on the neck length and to keep the bending moment 35 N-m in all samples	5° of adduction
Stevens et al. (1995)	27 Synthetic	No prosthesis	Transverse 200 mm distal to the greater trochanter	Physiological Displacement applied monotonically up to 25 mm in 50 sec then 20 displacement cycles between 25 and 15 mm at 1 Hz and finally monotonic displacement increased from 25 to 40 mm in 60 sec	29° of adduction 60° posteriorly relative to the frontal plan with the relative angle of 20°between the loading arm and femur in the anterio-posterior view
Schmotzer et al. (1996)	7 (4 left,3 right) Cadaveric	Cementless, Porous-coated Anatomic [PCA], E Series, Stryker, NJ	Transverse at the tip of the stem	Physiological Via a rigid load arm in 10 N steps up to failure	15° flexion 7° adduction
Han (2000)	11 Cadaveric	Cementless, Straight tapered, collarless Natutal, Sulzer Orthopedics, TX	Induced via a stem one size larger than the templated stem[a]	Isometric Compression to 890 N followed by 1780 N and 2670 N each for 15 sec	Not clear
Dennis et al. (2000)	30 (6 for each test) Synthetic	Cemented, Charnley, DePuy, IN	Oblique 45° to shaft axis distal to the tip of the stem	Isometric Axial compression to 500 N Lateral bending to 250 N Torsion to 200 N	25° of valgus
Dennis et al. (2001)	6 matched pairs Cadaveric	Cemented, Charnley, DePuy, IN	Oblique 45° to shaft axis	Isometric Axial compression to 500 N Lateral bending to 250 N Torsion to 200 N	25° of valgus
Kuptniratsaikul et al. (2001)	5 matched pairs Cadaveric	Cemented, Charnley, DePuy, IN	Spiral	Isometric is not clear	Not clear
Haddad et al. (2003)	16 Cadaveric	No prosthesis	Transverse 100 mm distal to the base of lesser trochanter	Physiological Cyclic cranial-caudal 1.53 BW approx 0-1000 N for 100 cycles at 1 Hz simultaneously loaded under anterior-posterior 0.15 BW approx -100 to80 N at 1.5 Hz	12° of adduction
Peters et al. (2003)	5 Cadaveric	Cemented, Premier Stem, Sulzer Orthopedics Inc, TX	Transverse 15 mm below the tip of the stem	Isometric Axial compression to 2250 N	Two set up tested 1) 21° of varus 2)30° of flexion
Wilson et al. (2005)	6 Cadaveric	Cemented, Charnley-Muller, Stryker, NJ	Transverse at the tip of the stem	Physiological Cyclic cranial-caudal 1.53 BW approx 0-1000 N and anterior-posterior 0.15 BW approx -100 to80 N	12° of adduction
Barker et al. (2006)	14 Synthetic	Cemented Exeter, Stryker, NJ	Cortical perforation at the tip of the standard stem	Physiological Initial loading cycled between 10 and 500 N, then every 100 cycles the peak load increased in steps of 500 N up to 2500 N	12°medially and 8°posteriorly relative to the frontal plan
Fulkerson et al. (2006)	8 matched pairs Cadaveric	Cemented, Charnley, DePuy, IN	Oblique 45° to shaft axis at the tip of the stem	Isometric Axial compression to 500 N Lateral bending to 36 N-m Torsion to 11 N-m Cyclic loading in 400 N compression for 10000 cycle at 3 Hz	25° of valgus
Talbot et al. (2008)	15 (5 for each test) Synthetic	Cemented, Exeter, Stryker, NJ	Transverse at the tip of the stem	Isometric Axial compression Lateral bending Torsion tested with displacement control between 0.5 and 2 with a preload of 50 to 100 N Cyclic between 200 N-1200 N for 100000 at 3 Hz	15° of adduction
Zdero et al. (2008)	20 (5 for each test) Synthetic	Cemented, Exeter, Stryker, NJ	Oblique 45° to shaft axis 25 mm distal to the tip of the stem	Isometric Axial compression to 500 N Lateral bending to 50 N Torsion to 100 N	25° of adduction

Key to content: [a]as a result of this crack occurred through the calcar and propagated distally from 50 mm to 165 mm.

Radiographic assessment has been undertaken by a number of authors to detect any gross abnormalities and determine the appropriate stem size prior to testing (Haddad et al., 2003; Han, 2000; Schmotzer et al., 1996) as well as to verify the position of the prosthesis and cement mantle following stem insertion (Kuptniratsaikul et al., 2001; Peters et al., 2003). This technique could potentially provide a method for evaluating the bone quality or fracture type and perhaps matching specimens between groups. However, radiographic analysis does not yet appear to have been utilised for this purpose.

To avoid the variance issues of cadaveric tissue, a number of studies have opted to use synthetic rather than natural bone for mechanical testing. Such specimens generally represent an average geometry and successive generations have been developed to more accurately match the properties of real bone (Cristofolini et al., 1996). The results from these tests generally show lower standard deviations in the measurements taken than those using cadaveric specimens and sample sizes of 5–10 have been sufficient. However, there are some exceptions. For example, the cadaveric study of Kuptniratsaikul et al. (2001) reported a lower standard deviation compared to the synthetic study of Dennis et al. (2000). Although they both reported a similar bending stiffness (mean 394 N/mm [SD 29 N/mm] and 410 N/mm [SD 100 N/mm] respectively for plate fixation with three proximal uni-cortical screws and three distal bi-cortical screws), which shows a reasonable level of correspondence between synthetic and cadaveric samples in terms of mechanical properties.

The periprosthetic fractures in the specimens have most commonly been generated using a saw. The position and configuration of the fracture has varied between studies, as is shown in Table 1, although in most cases, the type falls within the Vancouver B1 category. In all of the studies, the fracture configuration was kept as constant as possible between specimens, and, as yet, the effect of varying the fracture configuration on specific or different fixation methods has not been investigated.

2.1.3. Representation of the loads and surrounding conditions

A variety of testing setups and loading regimes have been used to examine the performance of the construct. In general, these fall into two categories: physiological loading where typical *in vivo* conditions are represented (Barker et al., 2006; Haddad et al., 2003; Wilson et al., 2005), or isometric loading where a series of isolated loading modes such as pure bending, torsion and compression are applied (Dennis et al., 2000, 2001; Fulkerson et al., 2006; Talbot et al., 2008; Zdero et al., 2008). Examples of these regimes are given in Fig. 2. The latter approach makes it easier to identify if specific loading modes cause instability, but it may be that these cases do not occur *in vivo*. Conversely, physiological loading cannot capture abnormal situations such as falls which may present the highest potential for instability and failure to occur. Both of these loading types have been applied using a single loading step (Panjabi et al., 1985; Schmotzer et al., 1996; Zdero et al., 2008), a cyclic loading regime (Haddad et al., 2003; Wilson et al., 2005) or a combination of both (Dennis et al., 2001; Fulkerson et al., 2006; Stevens et al., 1995; Talbot et al., 2008). However, there has been little consensus on the boundary conditions, magnitudes or directions of the loads applied, as shown in Table 1; here only the value of 500 N has been used repeatedly for non-destructive monotonic tests under axial loading (Dennis et al., 2000, 2001; Fulkerson et al., 2006; Panjabi et al., 1985; Zdero et al., 2008).

Whilst some studies have implemented non-destructive loads to enable the specimen to be re-used, others have evaluated failure through a step-by-step increase in the load. However, the definition of failure is not consistent across existing studies. Schmotzer et al. (1996) defined failure as complete loss of the fixation, permanent displacement/rotation of the fracture fragments or displacement (or rotation) amplitude greater than 2 mm (or degrees), whereas Talbot et al. (2008) and Zdero et al. (2008) define the “clinical failure” as either 10 mm of displacement or the first abrupt drop in load after reaching a peak load. This discrepancy is in part due to the different modes of failure anticipated with the different fixation methods used. As yet, no standard criteria have been adopted and, indeed, it would be difficult to apply a single criterion across the range of testing modes currently used.

2.1.4. Accuracy and repeatability of measurements

The purpose of the experimental studies has been to assess the mechanical stability of the construct, and a number of different approaches have been taken to measure this. These include methods to measure the relative movement across the fracture site as well as the larger scale performance of the whole construct in terms of the overall stiffness.

In order to quantify the movement at the fracture site, camera systems have been used (Haddad et al., 2003; Peters et al., 2003). Since the fracture movement was generally found to be small, the precision of the measurement system is important. Peters et al. used an OptoTrak system (Northern Digital inc, Canada) with a quoted accuracy of 0.1 mm. The measured fracture motions ranged from 0.5 to 2 mm, indicating the potential for the measurement error to be significant. A similar system was used by Haddad et al. (2003), however, by applying and tracking markers away from the fracture site, larger movements could be detected (>1.6 mm) from which the fracture movement was derived.

Evaluation of the whole construct behaviour is less challenging and many studies have evaluated the specimen stiffness from the slope of load–displacement curve generated from the mechanical testing machine. Since the displacements across the whole construct are larger, then the issues of accuracy are less critical.

Finally strain gauging has been performed in a limited number of studies (Barker et al., 2006; Panjabi et al., 1985) to capture the

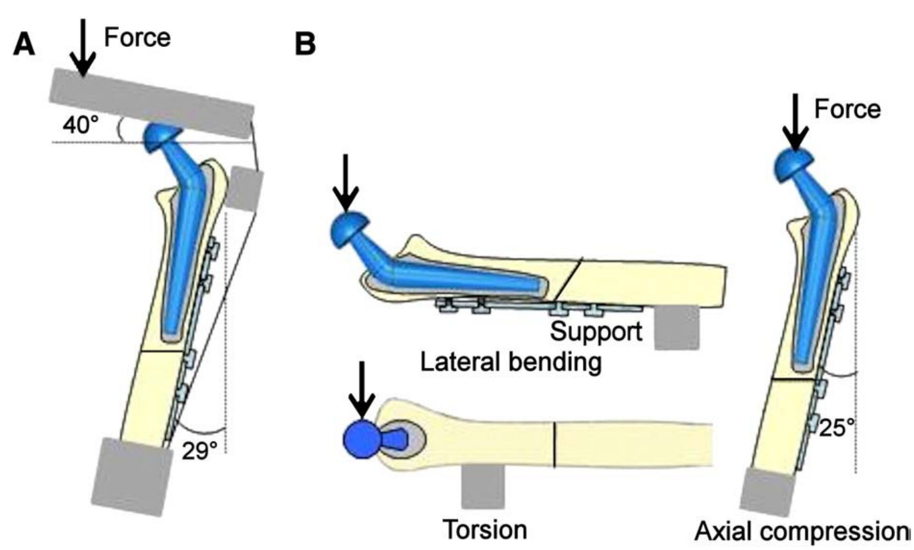

Fig. 2. A schematics of the loading protocol used in previous studies. A: a so-called physiological loading that includes the muscle forces in the model adopted and modified from Stevens et al. (1995); B: a so-called isometric loading that loads the construct under pure axial, bending and torsion adopted and modified from Zdero et al. (2008).

deformation of the construct under loading. This requires the attachment of strain gauges at specific locations on the construct or the bone surface. There are a number of issues with strain gauging that can affect the accuracy and repeatability of measurement. For example, thermal drift can cause changes in the measured voltage due to temperature fluctuations. This has been addressed using temperature compensating gauges and circuits (Panjabi et al., 1985). A high reproducibility rate was also reported in the strain measurements in this study (a standard deviation of approximately 3%) suggesting that repeatable results could be obtained from strain gauges.

2.2. Computational methods

2.2.1. Introduction

Finite element (FE) analysis is a computer modelling technique which allows the prediction of the mechanical behaviour of structures. Its application in the area of orthopaedic biomechanics dates back to early 1970's (see review by Huiskes and Chao, 1983). A great advantage of computer modelling over laboratory experiments is the ability to test large numbers of scenarios with little extra cost per test. This is particularly useful for PFF testing where the wide variety of fracture configurations, bone quality and device types cannot realistically be compared through physical testing alone. As well as the simulation of the force–displacement behaviour and the fracture movement, as can be recorded in physical experiments, the FE method also enables the stress and strain patterns within each of the components to be predicted.

Despite the large number of FE studies on femoral fracture fixation (Beaupre et al., 1988; Cegonino et al., 2004; Cheal et al., 1984; Chen et al., 2010; Helwig et al., 2009), there have been limited investigations that have implemented this technique to explore the biomechanics of PFF fixation (Mihalko et al., 1992; Mann et al., 1997- see also study of Pappas et al., 2006 for plate development for PFF fixation). The following section reviews the computational methods used and examines their robustness. Due to the limited number of computational studies of PFF, techniques and findings from the wider pool of femoral simulation work are drawn upon. Again, three aspects are considered in depth, which reflect those discussed in the previous section: the representation of the femoral bone and the fracture within it, the loading of the construct *in silico*, and the accuracy of the outputs of interest.

2.2.2. Representation of the femoral bone and fracture

Femoral bone has an irregular geometry and is an inhomogeneous, anisotropic structure (that is, the properties vary with both location and loading direction). Computer representations of this bone have ranged from a simple cylinder with homogeneous, isotropic material properties (Krishna et al., 2008), to geometrically accurate models with material properties matched to an individual cadaveric femur (Schileo et al., 2008).

Many authors have approximated femoral bone behaviour as homogenous and isotropic (Peleg et al., 2006) while some have implemented orthotropic properties (Mihalko et al., 1992). In cases where geometry is taken from computed tomography scans, it is possible to assign the bone material properties on an element by element basis (Mann et al., 1997), which enables the inhomogenous nature of bone to be taken into account. Where the intact femur has been modelled, Peng et al. (2006) showed that there was little difference in peak stress and nodal displacement between models using isotropic and orthotropic material properties. However, Gomez-Benito et al. (2005) showed that modelling bone with anisotropic rather than isotropic bone property led to a better prediction of femoral neck fracture when compared to corresponding experimental results. The key objective in a PFF fixation study is likely to be the assessment of the fracture movement, rather than the prediction of the original fracture location as studied by Gomez-Benito et al. (2005) and Schileo et al. (2008). However, as yet, no studies have investigated whether a simplified isotropic material property can provide reliable results or whether more sophisticated material properties should be used in this case.

In order to represent the fracture within the bone, the fracture surfaces can be simulated with contact interactions, which allow transference of the compressive contact stress across the fracture site and, potentially, enable the friction at the interface to be simulated (Krishna et al., 2008). The effect of the properties at this interface on the movement of the fracture is undoubtedly high. However, few studies have experimentally evaluated the friction coefficients of long bones (Shockey et al., 1985) and the friction across the femoral fracture site has not yet been fully characterised. In addition, the use of contact algorithms significantly increases the computational time. An alternative approach involves the use of solid elements, of a much lower stiffness than the surrounding bone, at the fracture site (Chen et al., 2008). This method potentially reduces the computational cost and could be used to simulate the callus formation during healing, which would limit fracture movement. However, low stiffness elements may require re-meshing if large deformations occur, which would increase the computational cost.

2.2.3. Representation of the loads and surrounding conditions

An advantage of the FE method is that, once a model has been developed, it is relatively straightforward to alter the loading and boundary conditions, and both isometric and physiologic loading regimes can be applied.

Where PPF has been investigated, forces representing typical physiological loading have been assigned (Mann et al., 1997; Mihalko et al., 1992). As well as forces applied to the femoral head, physiological loading in FE models has been extended to include the muscle forces, which are difficult to apply experimentally. This commonly includes the abductor muscle (Crowninshield et al., 2004; Mann et al., 1997; Mihalko et al., 1992), but can include all the muscles attached to the femur (Duda et al., 1998). The addition of muscle forces potentially creates a more accurate representation of the *in vivo* scenario but it also adds extra assumptions and complexity to the modelling process. The muscle force information is generally derived from electromyography or multibody dynamic analysis (MDA), which enables the values to be estimated through a walking cycle or other activity (Brand et al., 1986; Duda et al., 1997; Shi et al., 2007). Inevitably, there is considerable variation in these forces from person to person, for example due to size, weight and level of activity. Therefore assumptions have to be made as to whether the derived forces are representative of those that would be applied to the femur being modelled.

In addition to physiological loading, some studies have used finite element methods to investigate the femur under isometric loading. In a similar manner to that used experimentally, the femur model has been constrained at the distal end and loaded under compression, torsion and bending via the proximal section (Krishna et al., 2008; Stoffel et al., 2003; Yoon et al., 1989). The FE method allows the effects of individual isometric loading modes to be evaluated with more certainty than experimentally, because the forces can be controlled to prevent unintentional loading in other modes. To our knowledge, these methods have not, as yet, been applied to models specifically to examine PFF.

2.2.4. Simulation predictions and accuracy

In both experimental and computational PFF studies, the main measurements of interest are the construct stiffness, measured through bulk force–displacement data, and the potential for healing, judged by movement at the fracture site. In addition, some studies have examined the likelihood of construct failure, as well as the stress and strain within the bone and implanted devices. In an FE model, the accuracy of these predictions depends on the choice and sophistication of the input parameters, as well as the number of computations used to derive the solution, which is related to the number and type of elements in the model (Anderson et al., 2007).

The choice of input parameters for a PFF model will depend on how much each input affects the output of interest. For example, Papini et al. found that the effect of the presence of cancellous bone on the axial and torsional stiffness of a simulated synthetic femur was less than 1% (Papini et al., 2007), and therefore reduced the complexity of their model by neglecting this section of the bone. Whilst the geometry and properties of the implanted devices are usually well documented, those of the bone region are more challenging, and are expected to affect the bulk stiffness of the simulated construct. Equally, the representation of the bone fracture is expected to heavily influence the movement at the fracture site.

The resolution of the finite element mesh directly affects the accuracy of the solution, and should therefore be calibrated at the beginning of each study (Viceconti et al., 2005). The two finite element studies simulating PFF fixation, in two-dimensions (Mihalko et al., 1992) and in three-dimensions (Mann et al., 1997), both generated low resolution meshes in comparison to current computational capabilities (Helwig et al., 2009; Schileo et al., 2008), and do not report how the mesh resolution affects the accuracy of their results. More recently, a whole-femur mesh of 44,000 elements was used by Papini et al. to predict bulk stiffness. Although, in that case, increasing the mesh resolution was found to change the displacement by less than 1%, studies which analyse local bone stress would likely need further mesh refinement.

3. Overview of results

The cases studied and the key results of the experimental and two FE studies are summarised in Table 2. Whilst the results of individual studies allow comparison between the different fixation configurations tested, the lack of standardisation in the tests undertaken prevents direct comparison between tests. There are, however, a number of general trends that can be observed. Most of the results indicate, as would be expected, that increasing the overall rigidity of the construct increases the stability of the fracture, as measured by the overall stiffness of the instrumented femur or by the motion across the fracture (see review by Howell et al., 2004). This increase in rigidity has been achieved by:

a) increasing the number of plates and/or struts (Kuptniratsaikul et al., 2001; Peters et al., 2003; Talbot et al., 2008; Wilson et al., 2005)
b) increasing the rigidity of the connectors, by using screws in preference to cables (Dennis et al., 2000, 2001; Wilson et al., 2005), or cables in preference to wires, (Haddad et al., 2003), or double wrapped wires compared to single wrapped (Stevens et al., 1995)
c) using longer revision stems (Barker et al., 2006; Schmotzer et al., 1996)

There are also some contradictions between studies. For example, Haddad et al. (2003) and Wilson et al. (2005) did not find a clear trend between increasing the strut length and reduction in fracture movement (in anteroposterior and mediolateral plane). Haddad et al. (2003) reported that decreasing strut length decreased axial rotation. They suggested that the fit of the strut to the femur could have a major role on fracture stability, with a smaller strut providing better fit to the bone. The closer cable spacing for shorter strut was also thought to be a contributing factor. In contrast, Peters et al. (2003) found that the strength of fixation in mediolateral plane, under single-leg stance loading, was significantly improved with the longer struts. They also did not find any significant difference under stair-climb loading between 160 and 200 mm strut lengths. This could suggest that femoral fixture design could be a potential factor that has led to difference in results of these studies with loading protocol being another important factor (cyclic loading in studies of Haddad et al. and Wilson et al. versus monotonic protocol in study of Peters et al.).

There is also disagreement about the relative benefits of locking and non-locking plates. Fulkerson et al. (2006) found that the locking construct was stiffer than the non-locking, Ogden, construct in axial and torsional loading but not under bending. In addition, although there was no statistical difference between the torsional failure loads, the locking construct failed at the screw holes in the proximal bone, whereas the Ogden construct allowed large rotations of the proximal bone fragment due to cable loosening. However, in similar comparisons between locking and non-locking plates, both Talbot et al. (2008) and Zdero et al. (2008) did not find a marked difference between the two different designs. Zdero et al. (2008) suggested that the discrepancy may lie in the specimens used, since both their study and that of Talbot et al. (2008) tested synthetic specimens whereas Fulkerson et al. used cadaveric tissue. Locking plates may be more effective where the bone quality is poor, a factor that is not captured in the synthetic specimens.

In the case of long stem revision, the results of some studies (Mihalko et al., 1992; Panjabi et al., 1985) show that bypassing the femoral defect by 1.5–2 femoral diameters could reduce the stress riser effect of the implant and Mann et al. (1997) showed that by passing the femoral defect by two femoral diameters would be most effective in minimising the risk of loosening. In contrast, Barker et al. (2006) did not find that longer stems reduced the subsidence. They suggested that this difference in results compared to Mann et al. could be due to the impacted bone layer around the stem that was included in their experiments but not in the computational model of Mann et al.

4. Discussion

The current biomechanical literature relating to periprosthetic fractures was reviewed in this paper. Whilst there is some agreement in the results presented, there were also a number of differences, and it is clear that several issues need to be resolved if biomechanical studies are to provide clinically relevant information.

Firstly, there is currently a lack of standardisation in the methods used. In the experimental studies, there is a lack of consistency in both the testing procedures and the measurements. This means it is difficult to make conclusive comparisons between the findings, which would be particularly useful since each experimental study can only examine a small subset of the patient variables and fixation methods available. A similar issue is also apparent in the computational studies. In this field, there is currently a drive towards the development of minimum standards of model validation, verification and sensitivity analysis to prove the robustness of the model and the conclusions drawn from it (Anderson et al., 2007; Viceconti et al., 2005). As yet, the computational studies undertaken in the area of PFF have not fully demonstrated their compliance with these standards. Without such published information, it is difficult to gauge the validity of the results or the range of cases over which they apply.

Secondly, the level of sophistication in both experimental and computational models is variable. In the experimental tests, there is generally a trade-off to be made between accuracy and consistency. This can be seen in the choice of specimen (cadaveric versus synthetic), and type of loading (physiological versus isometric). In the computational studies, this balance is between realism and time for development and processing, for example in the complexity of the geometry, material properties, fracture representation and loading. In both cases, there has yet to be consensus on the level of complexity required to generate clinically relevant results.

Thirdly, from the methods shown in Table 1, it is clear that at present, biomechanical studies have concentrated on a relatively small subset of the types of periprosthetic fracture seen clinically. In fact, most of the experimental studies have focused on Vancouver type B1 fractures with types B2 and B3 being more challenging to test. Less attention has paid to types A and C fractures perhaps due to

Table 2
A summary of fixation method and results of the current laboratory and computational studies investigating biomechanics of the periprosthetic femoral fracture fixation.

Authors	Test case							Results
	Experimental studies							
	Plate and strut fixation							
	Lateral plate fixation			Strut fixation				
	Proximal		Distal	Position	Strut length (mm)	Proximal	Distal	
	Unicortical Screw	Cable/ wire	Bicortical Screw			Cable/ wire	Cable/ wire	
Schmotzer et al. (1996)	–	–	–	Med & lat	160	3 C	3 C	Long stem with allograft and cable provided the highest force to failure.
	–	–	–	Med & lat	160	3 C	3 C(LS)[b]	
	–	4 C	4	–	–	–	–	
	4 or 5[a]	–	4	–	–	–	–	
	–	–	–	–	160	3 W	–	
	–	–	–	Med & lat	–	–	3 W	
	–	–	–	–	–	–	-(LS)[b]	
Dennis et al. (2000)	–	3 C[c]	–	–	–	–	–	Constructs with screws or screws and cables were more stable than cables alone.
	–	3 C	3	–	–	–	–	
	3	–	3	–	–	–	–	
	3	3C	3	–	–	–	–	
	–	–	–	Ant & lat	120	3 C	3C	
Dennis et al. (2001)	–	3 C	3 C	–	–	–	–	Ogden concept provided a more rigid fixation than the allograft. Screws provided higher fixation rigidity than cables.
	–	–	–	Ant & lat	160	3 C	3 C	
Kuptniratsaikul et al. (2001)[d]	3	–	3	–	–	–	–	Double plating using an anterior and lateral plate provided significantly higher stability compared to single plate fixation methods.
	3	3 W	3	–	–	–	–	
	3[e]	–	3	–	–	–	–	
Haddad et al. (2003)	–	–	–	Ant &lat	200	3 C H T	3 C HT	Cables generated higher stability than wires and increasing the cable tension decrease fracture motion. Increasing the number of cables enhanced the fracture stability and decreasing strut length decreased axial rotation. No clear trend was found between strut length and fracture movement in anteroposterior and mediolateral planes.
	–	–	–	Ant & lat	200	3 C L T	3 C LT	
	–	–	–	Ant & lat	200	3 W HT	3 W HT	
	–	–	–	Ant & lat	200	2 C HT	2 C HT	
	–	–	–	Ant & lat	200	4 C HT	4 C HT	
	–	–	–	Med & lat	200	3 C H T	3 C H T	
	–	–	–	Ant	200	3 C H T	3 C H T	
	–	–	–	Lat	200	3 C H T	3 C H T	
	–	–	–	Ant & lat	160	3 C H T	3 C H T	
	–	–	–	Ant & lat	120	3 C H T	3 C H T	
Peters et al. (2003)	–	–	–	Lat	160	3 W	3 W	Increasing the length and number of struts increase the stability. Allograft struts are biomechanically equivalent to plate fixation using screws and cables.
	–	–	–	Lat	200	3 W	3 W	
	–	–	–	Ant & lat	200	3 W	3 W	
	–	–	–	Med & lat	200	3 W	3 W	
	–	–	–	Lat	200	3 C	3 C	
	–	3 C	4	–	–	–	–	
Wilson et al. (2005)	–	4 C	4	Ant	200	–	4 C	Combining plate and strut graft with two unicortical screws above the fracture side provided the most stable fixation construct.
	2	4 C	4	Ant	200	–	4 C	
	–	4 C	4	–	–	–	–	
	2	4 C	4	–	–	–	–	
	–	–	–	Ant & lat	200	4 C	4 C	
	–	–	–	Ant & lat	120	4 C	4 C	
Fulkerson et al. (2006)	3[f]	–	3	–	–	–	–	The locking plate was stiffer in axial and torsional loading compared the Ogden construct.
	–	3 C	3	–	–	–	–	
Talbot et al. (2008)	2	–	3	Ant	220	2 C	2 C	The allograft strut-plate construct showed higher stiffness in bending and had a higher load to failure than the plate alone.
	4[f]	–	4	–	–	–	–	
	2[f]	–	3	Ant	220	2 C	2 C	
Zdero et al. (2008)	4[f]	–	4	–	–	–	–	Allograft struts with non-locking plates showed higher stiffness compared to other methods.
	2[f]	2 C	4	–	–	–	–	
	2	2 C	4	–	–	–	–	
	2	–	4	Ant	220	2 C	2 C	
	Long stem							
Panjabi et al. (1985)	The effect of stem length (250, 200, 180,170-100 with 10 mm increment) using two different cemented prosthesis design without any additional fixation tested.							By passing the femoral defect by 1.5 FD minimize the stress raiser effect.
Barker et al. (2006)	Femoral defect following impaction bone grafting fixed using X-change femoral mesh, mesh and Dall-Miles plate, mesh and strut graft using standard and long stem prosthesis.							Following impaction bone grafting for revision of THR in the presence of a femoral defect using either a longer stem or extramedullary augmentation reduces the defect strain to a greater degree compared to a standard stem without augmentation.
	Wiring							
Stevens et al. (1995)	Extramedullary plate fixed using six wires in all cases three wires used proximally and three wires used distally and three different							Double wrap with insert plate was more rigid and stable that the other two methods.

(continued on next page)

Table 2 (*continued*)

Authors	Test case	Results
	Experimental studies	
	Wiring	
	wiring methods compared 1) single wrap 2) double wrap 3) double wrap using six plate inserts each time wires passing through eyelet of the insert plate.	
Han (2000)	Four different wiring methods compared 1) a wire applied parallel to the fracture line 2) a wire applied normal to the fracture line 3)a second wire added parallel to the wire in case 2 4)a third wire added parallel to the wires in case 3.	Placement of wires normal to the fracture line reduce the stem subsidence and crack opening in comparison to wiring in parallel to the fracture line.
	Computational studies	
Mihalko et al. (1992)	Three techniques were compared: 1)revision to a long stem prosthesis where effect of 3 stem lengths (passing the fracture site with 1, 2, and 3 FD) studied without any other additional fixation 2)lateral plating with a cortical bone allograft strut and cerclage wires (11 pass proximally and 8 pass distally) 3)plate with 5 proximal Parham bands and distal cortical screws –Ogden concept(rigid fixation considered between the plate and distal bony fragments).	In long stem revision bypassing the fracture by 3 FD can lead to a further stress shielding at the fracture site. In plate fixation allograft strut transfer the stress more evenly than Ogden plate.
Mann et al. (1997)	The effect of stem length (273, 240, 207,173 and 140 mm) using a cemented prosthesis design without any additional fixation tested.	A stem prosthesis bypassing the femoral defect by 2 FD (207 mm) will reduce the risk of loosening of cemented revision cases to higher degree than a short stems.

Key to content: C, cable; W, wire; HT, high tension; LT, low tension; Ant, anterior; Lat, lateral; Med, medial; LS, long stem; FD, femoral diameter.

[a] Based on the description and available data in the paper we are not confident if 4 or 5 unicortical screws have been used for the proximal fixation of the plate.

[b] This test was performed with a long stem compared to the standard stem in other cases.

[c] Plate fixation was performed with three proximal and three distal cables.

[d] A seven hole broad plate has been used. However the authors do not specify exactly how many screws and wires they used. We think that they have used three screws proximally and three screws distally.

[e] Two seven hole broad plates fixed laterally and anteriorly each with unicortical screws proximally and bicortical screws distally.

[f] A locking plate has been used in this case.

lower clinical incidence rates (Corten et al., 2009; Lindahl et al., 2006). Here, there is real potential for a computational approach to test and evaluate the effective fixation methods for much greater range of fracture scenarios.

Finally, and perhaps most importantly, the relationship between the results presented and the clinical situation needs to be better defined. At present, it is not clear how an 'optimum' construct would perform. There is currently a tendency to focus on the construct stiffness, but this may be misleading, especially because a high stiffness plate may cause stress shielding in the underlying bone. From a clinical point of view, there are two main issues: firstly that the fracture heals and secondly that the construct itself does not fail (Buttaro et al., 2007; Tsiridis et al., 2003). At present, neither of these are fully quantified in the biomechanical literature relating to PPF.

If biomechanical studies of PFF are to be used to optimise clinical performance, then all of these issues need to be tackled. One approach would be to use a parallel experimental and computational testing programme to utilise the advantages of each. The computational approach would enable parameters including bone quality, fracture configuration and location to be easily varied so that a much wider range of scenarios could be examined than has been undertaken so far. These models would, however, require validation (Viceconti et al., 2005). This could be achieved through parallel experimental testing which, with careful selection, could be undertaken on a much smaller subset of cases. The computational models could also be used to run more thorough sensitivity analyses to gain a greater understanding of the importance of the different input parameters and the level of sophistication required. The outcomes of the sensitivity analysis could, in turn, be used to determine the most robust experimental procedure and aid in the development of experimental standards.

In order to tackle this final issue, there is a need to draw more information from animal and *in vivo* studies (Claes et al., 1998; Goodship and Kenwright, 1985) and available measuring techniques for fracture healing (Cunningham et al., 1990) to determine the window of conditions across the fracture site that will enable healing to take place. There is also a need to gain a greater understanding of the failure mechanisms occurring clinically, which are not well documented at present. If more information could be gleaned on the likely loading at the time of failure, and whether this were due to repeated cycles of fatigue or single adverse occurrences such as falls, then these situations could be better replicated in the biomechanical models. There is therefore a need for the biomechanics community to work more closely with both biologists and clinicians to ensure that the knowledge in these areas is fully taken into account in the evaluation of PFF.

As the incidences of periprosthetic fracture continue to grow, the need for biomechanically optimised fixation methods becomes more pressing. The literature reviewed in this paper provides a foundation of experimental and computational procedures that evaluate a small subset of the situations seen clinically. This work could now be built on by combining the two approaches, along with better clinical information on fracture healing and modes of failure. This would enable more robust, clinically-relevant models to be developed and biomechanically optimised treatments for these types of fractures to be found.

Conflict of interest

The authors confirm that there is no conflict of interest in this manuscript.

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