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Review

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

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ABSTRACT

Background: Periprosthetic femoral fracture is a severe complication of total hip arthroplasty. A previous review published in 2011 summarised the biomechanical studies regarding periprosthetic femoral fracture and its fixation techniques. Since then, there have been several commercially available fracture plates designed specifically for the treatment of these fractures. However, several clinical studies still report failure of fixation treatments used for these fractures.
Methods: The current literature on biomechanical models of periprosthetic femoral fracture fixation since 2010 to present is reviewed. The methodologies involved in the experimental and computational studies of periprosthetic femoral fracture fixation are described and compared with particular focus on the recent developments.
Findings: Several issues raised in the previous review paper have been addressed by current studies; such as validating computational results with experimental data. Current experimental studies are more sophisticated in design. Computational studies have been useful in studying fixation methods or conditions (such as bone healing) that are difficult to study in vivo or in vitro. However, a few issues still remain and are highlighted.
Interpretation: The increased use of computational studies in investigating periprosthetic femoral fracture fixation techniques has proven valuable. Existing protocols for testing periprosthetic femoral fracture fixation need to be standardised in order to make more direct and conclusive comparisons between studies. A consensus on the 'optimum' treatment method for periprosthetic femoral fracture fixation needs to be achieved.

1. Introduction

Periprosthetic femoral fractures (PFF) is a severe complication following total hip arthroplasty (THA); the rate of intraoperative PFF ranged from 0.1–27.8% and of postoperative from 0.07–18%. PFF are more frequent in uncemented than cemented both in primary and revision THA (e.g. Biggi et al., 2010; Dubov et al., 2011; Fleischman and Chen, 2015; Kenanidis et al., 2018). PFF account for approximately 6% of revision cases and are the third most common reason for revision surgery after aseptic loosing and infection (e.g. Lewallen and Berry, 1998; Lindahl et al., 2006; Marsland and Mears, 2012). This number is expected to rise substantially by 2030, with the increase in life expectancy of the general population also leading to a rising incidence of total hip arthroplasties (THAs), with PFF also expected to rise proportionally (Della Valle et al., 2010).

PFF can occur intra-operatively or post-operatively, creating a variety of different fracture configurations at different locations; many researchers classify PFF based on fracture type, position on the femur, and bone quality. The Vancouver classification system is the most widely used and accepted classification system for PFF (Duncan and Masri, 1995; Learmonth, 2004; Moazen et al., 2011). Fractures classified as Type A are fractures involving the trochanteric area. The majority (approximately 75%, – Lochab et al., 2017; Lever et al., 2010) of PFF, however, are Type B; located around and just distal to the tip of the stem, and are subdivided as B1 with the stem stable and good bone stock, B2 with the stem unstable and good bone stock, and B3 with stem unstable and significant bone loss. Type C are fractures located distal to the stem (Capone et al., 2017; Leonidou et al., 2013; Tsiridis et al., 2009).

These fractures can be challenging to manage and treat, and are

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most commonly found in osteopenic elderly women, or in patients who have experienced loosening of the femoral stem following low energy trauma (Kenanidis et al., 2018; Shah et al., 2011). Given the complex nature of PFF treatment, due to the combination of the fractured bone and existing prosthesis (Moazen et al., 2011), many factors are required to be taken into consideration in the treatment of PFF; e.g., sex, age, bone quality, fracture topography, previous hip revision procedures, implant stability, and types (e.g. cemented vs. uncemented stem - Della Valle et al., 2010). The Unified Classification System (UCS); a recently proposed treatment algorithm developed by Duncan and Haddad (Duncan and Haddad, 2014), outlines the principles of PFF treatment. Treatment for Type A fractures is dependent on two factors; fracture displacement and the importance of soft tissue attached. Non-displaced Type A fractures are typically non-operative and treated conservatively. In cases of displacement of the greater trochanter, surgical treatment typically uses cerclage wires or hook cable plates for fixation. In cases of the lesser trochanter, if the fracture compromises the stability of the implant, cerclage wiring and implant revision may be considered (Biggi et al., 2010; Schwarzkopf et al., 2013). Management of Type B fractures is determined by subtype. B1 fractures can be treated by reduction and fixation using minimally invasive plate osteosynthesis (MIPO). In B2 fractures, revision surgery with a longer stem is commonly used. B3 fractures require more complex reconstruction or salvage procedures (megaprosthesis, allograft/stem composite). Type C fractures can be treated as a non-periprosthetic fracture. Specialized techniques can be used in some cases if hardware required for fixation will extend towards the implant, such as cerclages and unicortical screws (Capone et al., 2017; Duncan and Haddad, 2014).

While the Vancouver classification determine the treatment for PFF, many clinical cases still report failure of femoral fracture fixation due to mismanagement; the misclassification of B1 and B2 fractures is the main reason for the greater reported failure of B fractures (Kenanidis et al., 2018). For example, up to 20% of loose stems are missed on preoperative radiologic evaluation; many surgeons also fail to adequately test stem stability in the operating room leading to inappropriate selection of surgical methods for treatment (Fleischman and Chen, 2015; Niikura et al., 2014). This suggests that protocol for classifying PFF and subsequent fixation method is still insufficient. Indeed the reliability of any classification system depends on inter-observer and intra-observer consistency (Rayan et al., 2008). Optimal management of PFF remains controversial and debated, given that adequate fixation needs to be achieved without compromising the stability of the hip prosthesis. Although PFF is a rare complication, understanding risk factors and optimum treatment for fixation is still of high importance, as one study documented a higher risk of death after PFF compared with a similar population of patients undergoing uncomplicated THA (Della Rocca et al., 2011; Lindahl et al., 2007).

Finite element (FE) analysis is a computational modelling technique that allows prediction of the mechanical behaviour of structures. Used for orthopaedic biomechanics since the early 1970's it has been increasingly utilized by a number of authors to study structural-mechanical problems such as stress and strain analysis of bone, joints, and load-bearing implants (Huiskes and Chao, 1983; Kluess et al., 2010). Computer modelling allows a large number of scenarios to be tested with little extra cost per test making it advantageous over traditional experimental studies. To optimise management of PFF fixation, there have been a number of computational studies dedicated to simulating their biomechanics.

In 2011, Moazen et al. summarised the biomechanical research investigating PFF fixation following THA and its treatment methods. However, since then, there has been a large influx of biomechanical and computational studies carried out, and this is the basis of this paper. The aim of this paper was to provide an updated review of current research relating to PFF following THA published since 2011; currently, available literature pertinent to the biomechanical analysis of PFF treatment methods will be examined. Results of the experimental and computational studies conducted from 2010 to present and their trends were evaluated. Results from this review were critically compared to previous studies, highlighting any evolutions in biomechanical analysis of treatment methods for PFF.

2. Methodology

Computerised scientific journal databases, i.e. Scopus, Google Scholar, PubMed, and Web of Science were searched with the following keywords: Biomechanical testing, analysis, Finite element analysis, computational modelling, periprosthetic femoral fractures, and total hip arthroplasty. All studies from the above-mentioned searches were then reviewed; studies were included if they met the following criteria: (1) English Language; (2) Biomechanical or computational studies of PFF after THA (3) femoral fractures. Additionally, all studies prior to 2010 were excluded as they were reviewed previously (Moazen et al., 2011). In total 39 articles were retrieved, with 30 experimental studies and 9 computational studies. In order to maintain linearity and continuation, this paper will follow the same format as the previous review.

2.1. Experimental methods

A total of 30 experimental studies were reviewed. In many of the present experimental studies, the basic methodology described by Moazen et al. (2011) remained the same. The previous paper highlighted three specific aspects in the experimental methodologies; type of specimen, loading protocol, and methods of measurement. Methodologies in respect to those three aspects typically remained the same, and in-depth details of these can be referred back to the previous review. For most of the studies, mechanical performance is compared by stabilizing a periprosthetic fracture in both a cadaveric or synthetic femur, and different loading protocols are applied to the construct (see Table 1).

2.1.1. Specimen type and repeatability

Despite basic methodology remaining the same, several noteworthy factors have emerged from the reviewed studies; in particular, current studies using cadaveric femora use a higher number of specimens compared to previous studies; where typical sample size ranged from 5 to 16 cadaveric specimens, compared to a sample size range of 10 [5 pairs – (Konstantinidis et al., 2010)] to 24 (Lehmann et al., 2010; Lenz et al., 2014) cadaveric specimens. One exception to this is Lenz et al. (2013) who used 45 cadaveric 70 mm segments of femora. In some studies, authors used the same femur to test different fracture scenarios; Ebrahimi et al. (2012) utilized a single synthetic femur to test experimentally and computationally model and mimic the same femur while intact, after injury, repair, and healing. While most studies used bone mineral density matched cadaveric femora, to ensure no lesions or pre-existing fracture, Lehmann et al. (2010) used an osteoporotic bone model, to represent the group with the highest incidence of PFF. While most cadaveric bones used were fresh frozen, two studies used embalmed femora (Demos et al., 2012; Konstantinidis et al., 2010).

2.1.2. Representation of loads and surrounding conditions

In respect to loading modes and surrounding conditions, higher loading modes have been used by several authors. In previous studies (Moazen et al., 2011), only 500 N could be seen used repeatedly for non-destructive monotonic tests; in present studies, loads of 700 N (Choi et al., 2010; Graham et al., 2015) to 2500 N (Pletka et al., 2011) have been used. A loading mode not seen in previous papers is four-point bending (Lenz et al., 2016a, 2016b; Lever et al., 2010; Lochab et al., 2017) and in one case three-point bending (Choi et al., 2010); examples of these can be seen in Fig. 1. The basic experimental setup seen in most of the experimental studies can be referred back to the previous review (Moazen et al., 2011). There is little consensus seen on loading protocols; loads to failure was also not consistent across the

Table 1
A summary of the specimen preparation and loading protocol in laboratory studies.

Authors	Specimen number and type	Prosthesis	Fracture	Loading	Femur position
(Lehmann et al., 2010)	24 Cadaveric (6 per group)[b]	Cemented, Exeter, Stryker Howmedica Osteonics	Oblique 45° osteotomy, level at tip of hip stem (only for group IV).	Four-point bending – load applied at 0.1 mm/s until fracture.	Horizontal position.
(Lever et al., 2010)	12 matched pairs Cadaveric (5 test modes, 15 test cases.)[b]	Cemented, (Company not mentioned)	Oblique 45° osteotomy	Axial compression – load of 250 N applied (two types tested – abduction, and forward flexion) Torsion – 250 N applied to anterior aspect of femoral head Four-point bending (2 types tested - antero-posterior and medio-lateral forces) – 250 N applied symmetrically on either side of osteotomy site.	Axial - 20° of abduction, and 20° forward flexion Torsion and 4-point bending–Horizontal orientation to simulate 90° of flexion
(Choi et al., 2010)	10 Synthetic	Cemented, Zimmer, Warsaw, IN	Transverse osteotomy, 20 mm fracture gap distal to tip of stem.	Sinusoidal axial loading of 50–700 N for 100 cycles at 2 Hz. Three-point bending – Vertical sinusoidal loads of 50 N – 500 N at 2 Hz for 300 cycles. Torsion – increasing sinusoidal torsional movements 3 N/m – 12 N/m applied at 0.5 Hz for 20 cycles. All tests repeated three times for each construct model.	25° of adduction
(Konstantinidis et al., 2010)	5 pairs Cadaveric[c]	Cemented, Bicontact, Aesculap, Tuttlingen, Germany	Transverse, 10 mm fracture gap distal to stem tip.	Axial and cyclic compression – 1000 N for 10,000 cycles, then progressively loaded to failure at 100 N/2000 cycles.	9° of adduction
(Pletka et al., 2011)	9 matched pairs Cadaveric[b]	Cemented, Ultima, DePuy, Warsaw, IN	Transverse. 10 mm distal to stem tip.	Sinusoidal cyclic loading – up to 10,000 cycles from 0 to 2500 N axial force. 0-15 Nm of torsion at rate of 1 Hz.	Vertical orientation[a]
(Shah et al., 2011)	3 Synthetic	Cemented, King Packaged Materials Co., ON, Canada	Transverse at tip of stem – 5 mm fracture gap near tip of stem	Displacement control, maximum vertical load of 1000 N axial force at rate of 5 mm/min applied.	15° of adduction
(Demos et al., 2012)	24 Cadaveric (6 per group)[c]	Cemented, 100 mm straight metal carriage bolt used instead of hip stem.	Oblique 45°, 20 mm fracture gap distal to hip stem	Axial compression to failure at 5 mm/s.	Vertical orientation, femoral shaft collinear to axis of loading[a]
(Lenz et al., 2012a)	12 Cadaveric[b]	Cemented, Charnley hip endoprosthesis, DePuy IN	45° - 10 mm distal to tip of prosthesis.	Cyclic Axial bending at 2 Hz with synchronal sinusoidal axial loading at 950 N for 10,000 cycles – Axial forces ranged from 50 N - 1000 N. After 10,000 cycles, increased load at rate of 0.1 N/Cycle until catastrophic failure starting from 1000 N.	12° Valgus
(Lenz et al., 2012b)	8 Synthetic (2 groups)	Uncemented, Mathys, Bettlach, Switzerland.	90° transverse osteotomy, 5 mm distal to tip of stem.	Cyclic testing at 3 Hz at constant amplitude of 1800 N for first 5000 cycles. Monotonically increasing sinusoidal load at rate of 60mN/cycle until failure starting from 2000 N	20° Valgus
(Ebrahimi et al., 2012)	1 Synthetic	Cemented, Exeter, Stryker, NJ, USA	Transverse 5 mm gap, 233 mm from top of cement potting cube.	Axial load, at maximum of 1500 N, at rate of 100 N/s	15° of adduction
(Lenz et al., 2013)	45 Cadaveric (segments - 5 per group)[b]	No prosthesis	None- 70 mm length fragments cut from the diaphysis of the femur were used	Axial load to failure at rate of 50 N/s Torsional testing at rate if 2.5 Nm/s until construct failure.	Axial – Vertical orientation Torsional – Horizontal orientation.
(Wähnert et al., 2014)	9 pairs, matched Cadaveric.(9 per group)[b]	Uncemented, Alloclassic, Zimmer, Switzerland.	Proximal horizontal cut and 45° distal cut 5 mm below stem tip.	Cyclic sinusoidal axial loading starting at 750 N, increased at 0.1 N/cycle at 2 Hz until construct failure.	Vertical orientation[a]
(Giesinger et al., 2014)	17 Synthetic (2 groups, 9 in NCB group and 8 in control)	Cemented, CPT, Zimmer, IN	Osteotomy 20 mm distal to tip of stem, 6 mm gap near stem tip.	Axial load of 100-400 N and torsional load of 1-4 Nm applied at 1.5 Hz for 20,000 cycles - osteotomy gap then filled with cement to simulate 'healed' fracture. Then Axial load of 100-1400 N and 1–10.8 Nm torsional load applied for 80,000 cycles	7° Valgus
(Brand et al., 2014)	8 Synthetic	Cemented, Ecofit, Implantcast, Buxtehude, Germany.	15 mm below tip of stem	Axial load to failure – constant increasing load applied with a starting force of 0 N	6° Valgus
(Lenz et al., 2014)	24 matched, Cadaveric[b]	Cemented, Charnley, DePuy, IN			Valgus

(continued on next page)

Table 1 (*continued*)

Authors	Specimen number and type	Prosthesis	Fracture	Loading	Femur position
			60° - 10 mm from stem tip – Distal portion of femur and plate embedded in PMMA.	Axial bending – 50 N to 200 N at rate of 30 N/s. Cyclic testing at rate of 2 Hz, synchronal axial loading with constant valley load of 200 N. 1000 N peak load level increased at rate of 0.1 N/cycle until catastrophic failure	
(Hoffmann et al., 2014)	15 medium Synthetic (5 for each test)	Uncemented VerSys, Zimmer, IN	Oblique 45° to shaft axis at the level of implant tip.	Axial compression - loaded to 500 N at 20 N/s Lateral Bending – loaded to 250 N at 10 N/s Torsion/Sagittal bending – loaded to 200 N at 10 N/s Axial cyclic loading – 50-500 N load applied at 3 Hz for 10,000 cycles. After cyclic loading femurs tested again for all three modalities then loaded to failure or 100 mm displacement in torsional/sagittal bending	10° adduction in frontal plane. Vertically in sagittal plane.
(Sariyilmaz et al., 2014)	15 large, left Synthetic (5 for each test)	Uncemented, Synergy, Smith& Nephew, TN	10 mm fracture gap at level of prosthesis tip – (transverse)	Cyclic rotational loading 10 repeated cylindrical twists at 3 Hz between 0.5 and 10 Nm for 10,000 cycles Cyclic axial loading – force control - 50 N-500 N for 1000 cycle loadings, with 10 repetitions at a 3-Hz. Axial Failure – displacement control – force applied with speed of 15 mm/min until failure.	15° Valgus for cyclic axial loading.
(Griffiths et al., 2015)	12 large, left, synthetic (6 for each test)	Cemented, Exeter femoral stem	45° oblique -25 mm distal to tip of stem, one group had midshaft osteotomy (MO) (anatomically reduced) and the other midshaft gap (MG) (with 5 mm gap)	Axial compression, displacement control – preloaded 100 N to 1000 N, vertical load applied -500 N for MO, 250 N for MG. Lateral bending – 200 N vertical load at 8 mm/min Torsional stiffness – vertical load of 200 N at 8 mm/min Axial load to failure – preload of 100 N at load rate of 8 mm/min till catastrophic failure	Axial - 25° adduction to coronal plane, aligned vertically in sagittal plane. Laterial – horizontal Torsional - Horizontal
(Graham et al., 2015)	5 synthetic	Cemented, Exeter, Stryker SA, Switzerland.	4 fixed as if anatomically reduced. 1 with 10 mm gap	Axial load – displacement control 5 mm/min, max 500 N	0°, 10°, and 20° adduction for no gap model. 10° for gap model
(Gwinner et al., 2015)	20 large, left, synthetic	Uncemented, Alloclassic, Zimmer, Switzerland.	Transverse cut and 45° distal cut at level of implant tip. With 10 mm gap.	Cyclic sinusoidal axial loading starting at 30 N. Increased by 300 N every 1000 cycles.	Vertical orientation[a].
(Lewis et al., 2015)	30 Synthetic	Cemented, Zimmer, Warsaw, IN	Transverse, 25 mm distal to prosthesis tip. Distal part of femur not used to simulate segmental bone loss.	Torsional internal rotation. 20 preconditioning cycles at 100 N/1 Hz. Then loading rate of 8 mm/min until failure. Axial loading – Phase I: load of 4 mm/min until 1200 N. Phase II: 4 mm/min until failure/7500 N.	Torsional - 11° of prosthesis anteversion. Axial - 13° adduction
(Frisch et al., 2015)	24 synthetic	Uncemented, Zimmer, Warsaw, IN	Femoral neck osteotomy 10 mm proximal to lesser trochanter. Longitudinal fracture extending 127 mm distally	Axial load of 50 N pre-load followed by loading rate of 0.8 mm/min and terminated after displacement of 20 mm. Torsion – rotational displacements applied at rate of 2.4°/s, rotated through 40° until failure.	25° adduction, 0° anteversion
(Lenz et al., 2016a)	12 cadaveric, paired, (6 for each test)[b]	Cemented, Charnley, DePuy, IN	10 mm distal to tip of prosthesis. – orthogonal to shaft axis of femur.	Axial loading and displacement at 10 Hz 4-point bending and torsion tested with displacement control at 0.5 mm/min. Up to 250 N applied. Cyclic testing to failure with axial compression from 50 N to load plateau of 200 N at 30 N/s, increased peak load at 500 N at 0.1 N/cycle.	Cyclic testing - 12° valgus and 12° anteversion
(Moazen et al., 2016)	12 large, left synthetic	Cemented, Zimmer, Sulzer, Switzerland	20 mm below tip of stem.	Axial loading – up to 700 N	10° adduction
(Gordon et al., 2016)	20 synthetic (5 for each test)	1. Uncemented, short stem (10), AnaNova Solitär, ImplanTec, Austria	140 mm spiral fracture (100 mm proximal to 40 mm distal of stem)	Sinusoidal cyclic loading - 50 N-500 N at 2 Hz Axial Stiffness – stroke controlled 0.02 mm/s up to 500 N	6° adduction

(*continued on next page*)

Table 1 (*continued*)

Authors	Specimen number and type	Prosthesis	Fracture	Loading	Femur position
		2. Uncemented, long revision stem (10), Modular Plus, Smith and Nephew, Austria		Cyclic sinusoidal fatigue loading- 2000 N max load, increasing by 150 N/500 cycles until failure	
(Lenz et al., 2016b)	12 cadaveric, paired, (6 for each group)[b]	Cemented, Charnley, DePuy, IN	Transverse, 10 mm distal to tip of stem, orthogonal to femur shaft axis	Axial bending to 200 N at 30 N/s Cyclic mechanical testing at 2 Hz with synchronic axial loading increased at 0.1 N/cycle starting from 500 N until catastrophic failure. 4-point bending – 250 N max	12° valgus and 12° anteversion
(Walcher et al., 2016)	38 synthetic	Cemented, Weber standard straight stem, Zimmer.	None to simulate healed periprosthetic fracture situation.	Axial compression at 500 N over 5 s. Then 30 cycles from 400 N–1500 N at 0.25 Hz applied. Torsional testing to 0.6 Nm over 5 s, then 30 cycles of external rotation from 0.6 Nm–50 Nm at 0.25 Hz. Load to failure at constant displacement rate of 100 mm/min in axial loading.	7° valgus
(Wähnert et al., 2017)	10 Synthetic (5 for each group)	Uncemented, Alloclassic, Zimmer GmbH, Switzerland.	45°distal cut and horizontal cut 5 mm distal to stem tip.	Cyclic loading in axial compression at 2 Hz until failure – starting at peak load of 750 N with increment of 0.1 N/cycle.	Vertical orientation[a]
(Konstantinidis et al., 2017)	20 cadaveric	Cemented, Bicontact, Aesculap AG, Germany.	Transverse below tip of stem.	Fluctuating axial load (sinusoidal profile, 0.5 Hz, 2100 N) applied to prosthetic cone, repeated for 20,000 load cycles.	Standard adduction position
(Lochab et al., 2017)	9 pairs of cadaveric[b]	Cemented, DePuy Summit, DePuy Synthes, Warsaw, IN.	45° oblique osteotomy 25 mm distal to tip of stem. 5 mm fracture gap	4-point bending – rate of 8 mm/min with load up to 250 N Torsion and Axial compression – vertical force of 250 N Axial compression to failure or up to maximum vertical displacement of 10 mm.	20° abduction and 20° flexion

[a] Author didn't specify femoral position, but from the images provided, we believe standard adduction vertical positioning was used.
[b] Fresh frozen cadaveric.
[c] Formalin fixed cadaveric femora.

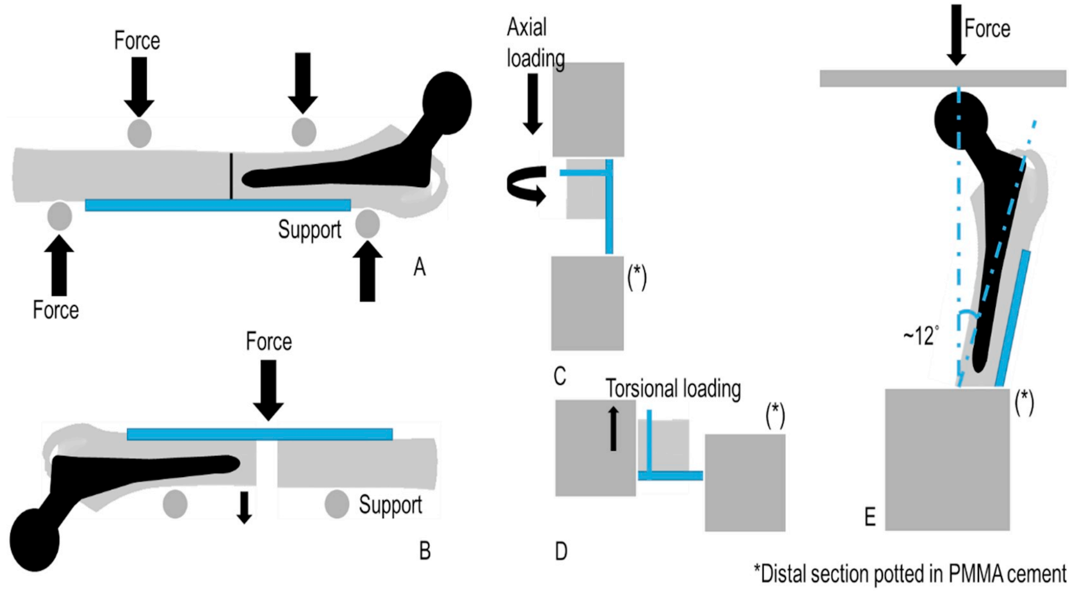

Fig. 1. Schematic diagram of different examples of loading methods used in tests.
A) 4-point bending (Medio-lateral) (Lever et al., 2010).
B) 3-point bending (Choi et al., 2010).
C-D) The embedded femoral shaft bone was connected to the actuator of the testing machine via a xy-table. Setup for axial loading (C) and lateral torsional loading (D) shown. (Lenz et al., 2013).
E) Test set up of specimen positioned in 12° valgus for cyclic testing. Distal part of femur is potted in PMMA cement (Lenz et al., 2012a).

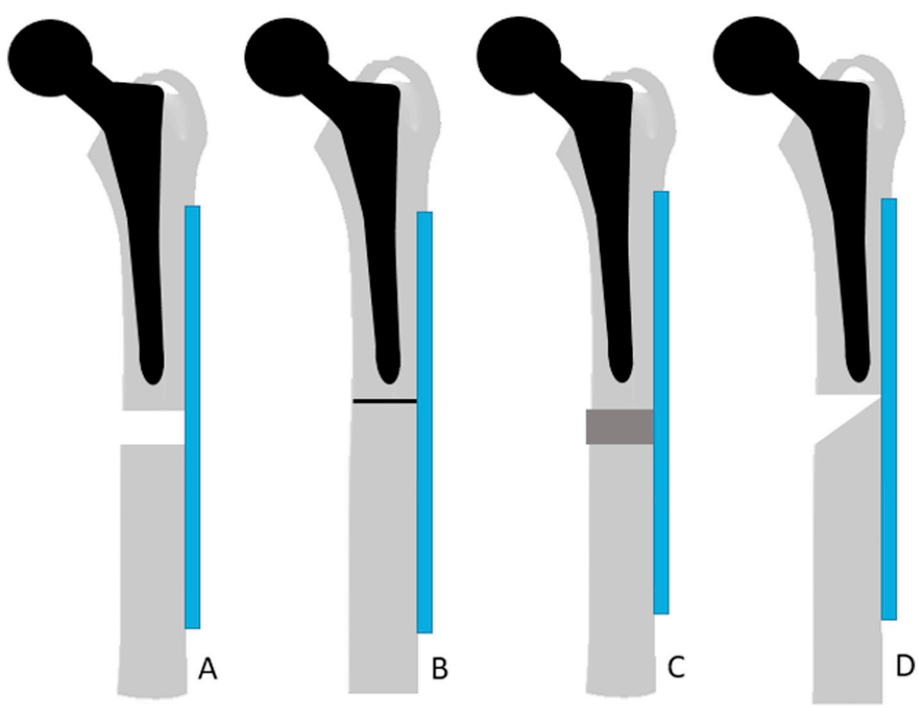

Fig. 2. Schematic diagram of different fracture gap variations used in experimental methods.
A) Fracture Gap (Choi et al., 2010; Giesinger et al., 2014; Graham et al., 2015; Griffiths et al., 2015; Gwinner et al., 2015; Konstantinidis et al., 2010; Lochab et al., 2017; Sariyilmaz et al., 2014; Shah et al., 2011).
B) No gap (Brand et al., 2014; Frisch et al., 2015; Griffiths et al., 2015; Hoffmann et al., 2014; Konstantinidis et al., 2017; Lehmann et al., 2010; Lenz et al., 2012a, 2012b, 2016a; Lever et al., 2010; Pletka et al., 2011).
C) Fracture gap filled with cement (Giesinger et al., 2014).
D) Fracture gap with a wedge-like cut (Gwinner et al., 2015; Wähnert et al., 2014, 2017).

studies, an issue that was raised previously. Boundary conditions, magnitudes, and direction of loads applied varied between authors, seen in Table 1.

The majority of studies reviewed here studied the biomechanical performance of typical variations of an Ogden construct; specifically examining the performance of the plate fixation and its fixation method to the femur via screws, cables, wires or in some cases struts. However, several new trends and parameters may affect the outcome of the fixation method examined across the studies published that was not investigated previously; including fracture gap, type of plate used, and screws and cement mantle integrity. These will be described below with an overview of the materials and methods, and updated parameters used in the studies.

2.2. Overview of recent developments

2.2.1. Fracture configuration

Most studies simulated a Vancouver B1 type fracture in their studies. Introduction of an osteotomy to simulate PFF was most commonly generated using a saw; although fracture position and configuration varies between the studies (Table 1). In many studies, no fracture gap was left after the osteotomy, in order to simulate a stable fracture pattern (Brand et al., 2014; Frisch et al., 2015; Griffiths et al., 2015; Hoffmann et al., 2014; Konstantinidis et al., 2017; Lehmann et al., 2010; Lenz et al., 2012a, 2012b, 2016a; Lever et al., 2010; Pletka et al., 2011). Other studies implemented a fracture gap (where the femur was not fixed as if anatomically reduced, and a gap was left between the fracture), typically below the tip of the hip stem prosthesis; fracture gap implemented ranged from 5 mm to 20 mm (Choi et al., 2010; Giesinger et al., 2014; Graham et al., 2015; Griffiths et al., 2015; Gwinner et al., 2015; Konstantinidis et al., 2010; Lochab et al., 2017; Sariyilmaz et al., 2014; Shah et al., 2011). Fracture gaps were typically used to mimic a fragmented fracture model (Sariyilmaz et al., 2014). Wähnert et al. (2014, 2017) and Gwinner et al. (2015) created a 45° and horizontal cut as the osteotomy gap, and a triangular wedge segment was removed. The fixed fracture with a gap between the proximal and distal fragments eliminates the compressive effect of the fragments, isolating the proximal fixation during testing and simulating a "worst-case" scenario with a comminuted fracture with no cortical apposition (Demos et al., 2012). See Fig. 2 for examples of different fracture gap configurations.

A few studies investigated the effect of fracture gap and no fracture gap (Giesinger et al., 2014; Graham et al., 2015; Griffiths et al., 2015); Giesinger et al. (2014) filled the osteotomy gap with cement after creating a fracture to simulate 'healed' fracture situation. In two studies, no fracture was created to simulate a healed periprosthetic fracture situation (Walcher et al., 2016) or a femur prior to fracture (Ebrahimi et al., 2012). Some studies did not use the distal part of the femur distal to the osteotomy; the femur and plate construct was cut accordingly (Brand et al., 2014; Lenz et al., 2012b, 2013, 2014; Lewis et al., 2015).

2.2.2. Plate type

With the recent interest in advancing strategies for PFF treatment, specialized plates have been developed for PFF, commercialized, and used in recent studies published; these include hook plates, locking compression plates (LCP), Variable Angle Locking plate (VA-LCP), locking attachment plates (LAP), Dall-Miles plates, cable-ready system, and non-contact bridging plate. Currently, the two main periprosthetic systems on the market and most notably studied are the Locking Compression Plate (LCP –Synthes, Solothurn, Switzerland) and Non-Contact Bridging Periprosthetic Proximal Femur Plate (NCB PP-Zimmer GmbH, Winterthur, Switzerland). Most researchers used these systems in their studies, and a few were interested in the direct comparison of different construct systems (Konstantinidis et al., 2010; Lever et al., 2010; Lewis et al., 2015; Wähnert et al., 2014). Some authors investigated the effect of strut allografts in place of a fracture plate or a fracture plate used in conjunction with a strut (Choi et al., 2010; Lochab et al., 2017; Sariyilmaz et al., 2014). The biomechanical performance of using two fracture plates on a single fracture (Fig. 3) was also investigated by several authors (Choi et al., 2010; Lenz et al., 2016a; Wähnert et al., 2017).

Several authors also studied use of bicortical screws for proximal plate fixation; Lochab et al., 2017; Griffiths et al., 2015; Gwinner et al., 2015; Hoffmann et al., 2014; Konstantinidis et al., 2010; Lenz et al., 2012b, 2014, 2016a, 2016b; Lewis et al., 2015; Wähnert et al., 2014, 2017). One recent commercial development and a method used to achieve proximal bicortical fixation was the locking attachment plate (LAP); a clamp-on plate that is compatible and can be used in conjunction with a conventional locking compression plate (LCP) in the treatment of PFF; the lateral arms allows for bicortical offset screw placement laterally to the prosthesis stem (Synthes, Solothurn, Switzerland) (Lenz et al., 2016b). The design of the NCB PP plate (Zimmer GmbH, Winterthur, Switzerland) also allows for proximal bicortical screw fixation. Fig. 3 shows examples of typical variations of the PFF fixation construct used.

2.2.3. Screws and cement mantle

A clinical concern regarding the way that a construct fixation is applied is the potential breach of cement mantle integrity; in particular, cortical screw tips infringing the cement mantle and potentially leading to substantial cement fracture and eventual hip implant loosening (Lever et al., 2010). Two authors (Kampshoff et al., 2010; Konstantinidis et al., 2017) studied the role of cement mantle integrity and screws in PFF. Konstantinidis et al. (2017) deliberately made a more brittle mantle by using hand-mixed rather than the advised vacuum mixed cement, and Kampshoff et al. (2010) forgoed typical plate fixation setup and investigated the effect of different screw implantation techniques by directly drilling different screws in the cement. Brand et al. (2014) proposed and investigated a novel fixation method – intraprosthetic fixation; where screws that fixed the fracture plate to the bone were also drilled and fixed to the cemented hip implant. Another important factor to note is that the risk of fractures is higher around the uncemented compared to the cemented implants (Fleischman and Chen, 2015). This is perhaps due to the higher interaoperative risk of fracture for uncemented implants (Wyatt, 2014). To best of our knowledge eight studies so far have investigated biomechanics of PFF fixation in uncemented hip implants (Frisch et al., 2015; Gordon et al., 2016; Gwinner et al., 2015; Hoffmann et al., 2014; Lenz et al., 2012b; Sariyilmaz et al., 2014; Wähnert et al., 2014, 2017).

2.3. Computational methods

A total of nine computational studies were reviewed in this paper, and the following section will examine the computational method used. Prior to 2010, there were only two computational studies investigating the biomechanics of PFF fixation. The previous review paper (Moazen et al., 2011) highlighted three main aspects in the computational methodologies; 1) representation of the femoral bone and fracture, 2) representation of the loads and surrounding conditions in silico, and 3) simulation predictions and accuracy. In-depth detail of these methodologies can be referred back to the previous paper. Here, developments to these three aspects described above are discussed, with the representation of the femoral construct instead of the femoral bone being highlighted, as well as current trends.

2.3.1. Representation of the femoral construct and accuracy

The increase in present computational capabilities allow for more geometrically accurate modelling of individual parts of the construct. Computational representation ranged from a simplified parametric FE model of a typical construct (Leonidou et al., 2015; Moazen et al., 2012) to more geometrically accurate 3D models. (Avval et al., 2016; Chen et al., 2012; Ebrahimi et al., 2012; Moazen et al., 2013, 2014; Shah et al., 2011; Wang et al., 2016). A clinical case was modelled using a simplified parametric FE model of the PFF fixation construct (Moazen et al., 2012). The bone, hip stem, and cement mantle were modelled as concentric cylinders. A simplified representation of a fracture fixation plate was used, and screws were modelled as cylinders with no screw thread or head. The model was validated against a clinical case study, suggesting that simplified models are sufficient when modelling different construct configurations. Older computational studies generated low resolution meshes [928–2184 elements (Mann et al., 1997; Mihalko et al., 1992)] in comparison to current computational capabilities [61000–400,000 elements (Chen et al., 2012; Ebrahimi et al., 2012; Leonidou et al., 2015; Moazen et al., 2012; Wang et al., 2016)]. All studies used tetrahedral elements to mesh components.

2.3.2. Representation of the loads and surrounding conditions

In almost all studies, FE models assumed the femur had linear, isotropic, and elastic properties. Studies performed by several current authors showed that linear behaviour was a good approximation for femurs when comparisons of FEA, synthetic femurs, and human cadaveric femurs were made (Dubov et al., 2011). However, in many

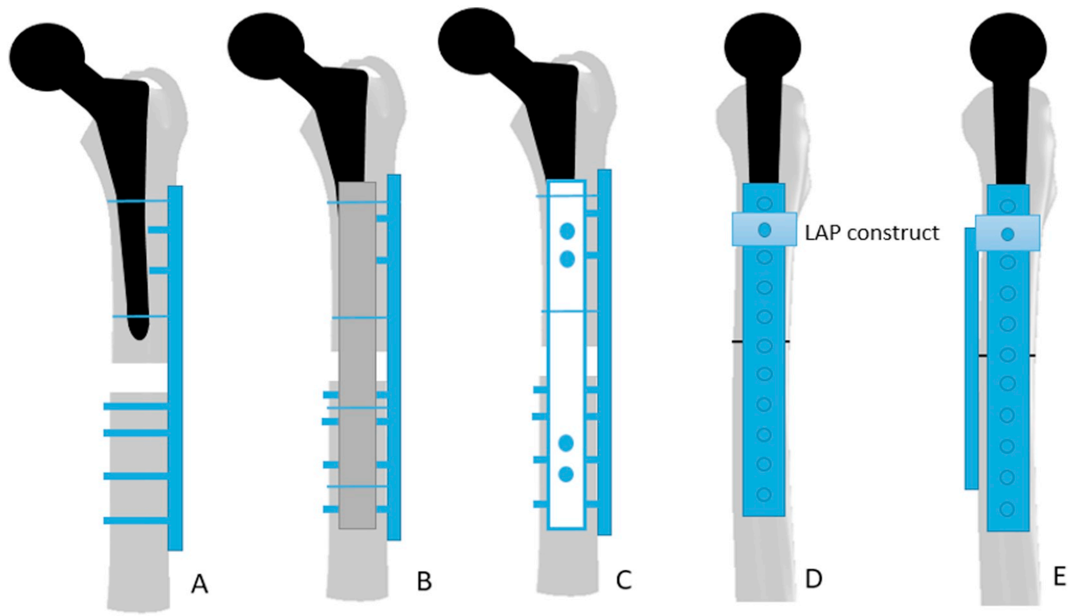

Fig. 3. Schematic diagram of different plate fixation methods onto a femoral construct with a hip stem.
A-C: Schematic diagram of a typical Ogden construct (A) and a construct with an additional plate fixed with wires (B) or with screws (C). (Choi et al., 2010).
D-E: Schematic diagram showing a construct with an additional LAP plate attached proximally to the plate (D), and with an additional LCP plate placed anteriorly (E). (Lenz et al., 2016a).

studies, the bone quality that was simulated experimentally and computationally were considered normal healthy bone stock, and not osteoporotic bone seen in PFF patients. Although Dubov et al. (2011) noted that relative performance of constructs would likely remain the same.

2.4. Overview of recent developments

2.4.1. Fixation methods

Classical computational studies of PFF fixation (Mann et al., 1997; Mihalko et al., 1992) investigated the effects of different stem lengths as treatment methods, although Mihalko et al. (1992) also studied the effect of plate fixation. Recent studies investigated a wider range of different fixation methods, and also the effect of fracture stability, bone quality, and fracture type. Fixation methods in present studies can be divided into two categories. The first category considers the effect of different plate fixations (Avval et al., 2016; Moazen et al., 2012, 2014; Moazen et al., 2013; Wang et al., 2016), typically direct comparisons between two plate types are made; such as rigid vs. flexible plating (Moazen et al., 2012), comparisons between the performance of stainless steel (SS) vs. titanium (Ti) plate fixations and plate thickness (Moazen et al., 2012, 2013), double cable fixation vs. locking plate vs. multi-directional plate (Wang et al., 2016), double plating (Moazen et al., 2014), and lateral vs. anterior plating (Avval et al., 2016). Plate fixation and long stem revision options under partial and full weight bearing conditions were also carried out by one group (Moazen et al., 2014). The second category considers the biomechanical performance of different variations of a typical Ogden construct; typically this involves different configurations of cable, wires, or screws positions (Chen et al., 2012; Dubov et al., 2011). Four studies modelled uncemented hip implants in their studies (Avval et al., 2016; Chen et al., 2012; Moazen et al., 2012; Wang et al., 2016)

2.4.2. Effect of fracture stability, bone quality, and fracture type

While the majority of computational studies focused on Vancouver B1 type fractures; there were several authors did investigate treatment methods for different fracture types (Leonidou et al., 2015; Moazen et al., 2012, 2014), in one instance a Vancouver type C clinical case comparing initially failed fixation vs a successful revision fixation was carried out (Moazen et al., 2012). Femoral fracture stability and bone quality was also computationally modelled by several authors (Avval et al., 2016; Ebrahimi et al., 2012; Leonidou et al., 2015; Moazen et al., 2013); Ebrahimi et al. (2012) investigated the stiffness and peak bone stress of the same femur after injury, repair, and healing with respect to its intact condition. Stable vs unstable fracture on plate fixation performance was also investigated (Moazen et al., 2013).

Avval et al. (2016) studied femoral density changes and bone remodelling in the femur in response to a bone fracture plate and uncemented hip stem implant using a validated mechano-biochemical model. Bone was hypothesized as a thermodynamic system that exchanges energy, matter, and entropy with its surroundings. The model they used assumed that the mechanisms of bone remodelling are executed by bone resorption and bone formation phases through five biochemical reactions (i.e. formation of multinucleated osteoclasts, old bone decomposition, production of osteoblast activator, osteoid production, and calcification.)

One study, by Leonidou et al. (2015) modelled an osteoporotic bone situation by developing three models with different canal thickness ratios (CTR) to represent poor, average, and best bone quality. Further three models were developed with angle fractures varying from unstable transverse (0°), and short oblique (146 °) to the stable long oblique configuration (76 °). Additional three models were developed with fracture at the tip of the stem, 4 mm, and 14 mm below the tip of the stem.

3. Results

Key results of the experimental and computational cases studied are summarised in Table 2. Several studies using computational methods were validated with experimental results (Dubov et al., 2011; Ebrahimi et al., 2012; Lenz et al., 2013; Moazen et al., 2013; Shah et al., 2011). The issue of lack of standardization between tests seen in past studies still exists, making it difficult to make direct comparisons. Most tests, like those seen in previous studies, show that increasing the overall rigidity of the construct increases the stability of the fracture. Rigidity was measured by the overall stiffness of the instrumented femur or by

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	
(Lehmann et al., 2010)	(a)	-	-	-	-	-	-	Two intramedullary implants in femur were associated with decreased fracture strength between these implants. Fracture plate between tip of the stems leads to good stability regardless of presence of osteotomy or retrograde nailing.
	-	-	(b)	-	-	-	-	
	3	-	3 (a) (b)	-	-	-	-	
	3	-	3(a) (b)	-	-	-	-	
(Lever et al., 2010)	-	4C	4	-	-	-	-	Screw-plate systems provided either greater or equal stiffness compared to cable-plates in almost all cases. No statistical differences between the three different plating systems used to compare cable vs screw fixation - Zimmer Cable ready system (Zimmer,IN, USA), AO cable-plate system (Synthes, PA, USA), and Dall-Miles cable grip system (Howmedica, NJ, USA).
	-	4 W	4	-	-	-	-	
	-	4C	4	-	-	-	-	
	3	-	4	-	-	-	-	
	3	-	4	-	-	-	-	
	3	-	4	-	-	-	-	
(Choi et al., 2010)	2	2	4	-	-	-	-	Fixation using double plates show highest stiffness, however results demonstrated that use of additional allograft strut in conjunction with a LCP also provided superior stiffness compared to single locked plate (LCP –Synthes) for Vancouver type B1 femoral fractures.
	2	-	4	Ant	188	2C	2C	
	2/2 (c)	2	4/4[c]	-	-	-	-	
(Konstantinidis et al., 2010)	4 (d - 5)	-	3	-	-	-	-	Bicortical screw placement (NCB plate; Zimmer, IN, USA) showed superior and more stable anchoring compared to unicortical screw fixation (LISS plate; Synthes, Switzerland). Mean force resulting in subsequent model failure similar in both models. Suggesting NCB plate was not superior to the LISS plate; moreover NCB system showed material fatigue under cyclic loading, suggesting increased implant failure rates particularly in cases of delayed bony union.
	4 BC (e-BC)	-	3	-	-	-	-	
(Pletka et al., 2011)	3	2	4	-	-	-	-	Type of plate and working length did not significantly affect failure rate, no significant differences was found between long and short plates for displacement or rotation at fracture site. Lower bone mineral density significantly associated with failure.
	3	2	4	-	-	-	-	
(Shah et al., 2011)	-	4C	4	-	-	-	-	Cables absorbed majority of load, followed by plates and then screws. Optimal mechanical stability can be achieved using cables and screws, then screws – as both had the highest stiffnesses. If only cables are used clinically, a plate without proximal holes recommended.
	4	-	4	-	-	-	-	
	4	4C	4	-	-	-	-	
(Demos et al., 2012)	3 (LS)	-	4	-	-	-	-	Proximal cable fixation provides significantly less axial stability compared to when cables and screws were used. Locking and non-locking screw constructs showed equivalent loads at failure, and superior in load at failure compared to cables.
	3 (LS)	3	4	-	-	-	-	
	3	3	4	-	-	-	-	
	-	3	4	-	-	-	-	
(Lenz et al., 2012a)	5	-	3	-	-	-	-	A-LCP (prototype locking plate) with proximal bicortical and unicortical screw fixation had higher number of cycles to failure compared to conventional LCP using proximal unicortical screw fixation, and showed higher construct stability and strength. Bicortical screw positioning showed less interfragmentary osteotomy movement, suggesting improved osteosynthesis in periprosthetic fractures.
	2 + 3 (e-BC)	-	3	-	-	-	-	
(Lenz et al., 2012b)	3	1 Ce	-	-	-	-	-	LAP-LCP construct group using additional proximal bicortical screw fixation had significantly higher stiffness and number of cycles to failure compared to cerclage- LCP construct. Use of LAP and placing bicortical locking screws laterally at prosthesis stem can improve stability in PFF fixation.
	3 (f- 2 BC)	-	-	-	-	-	-	

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Table 2 (*continued*)

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	
(Lenz et al., 2013) (k)	-	1 Ce	-	-	-	-	-	Both screw fixation types (Unicortical and bicortical) showed significantly higher ultimate strength and stiffness in axial compression and torsion compared to cerclage fixation. Results of mechanical test were visually confirmed by FEA for unicortical and bicortical screws.
	1	-	-	-	-	-	-	
	-	-	1	-	-	-	-	
(Wähnert et al., 2014)	2 (f – 2 BC)	-	3	-	-	-	-	Both fixation systems achieved proximal bicortical screw fixation around the hip stem. LAP-LCP construct found less stable due to less rigid main plate. NCB plate showed significantly higher stiffness and cycles to failure.
	4 (g)	-	3	-	-	-	-	
(Giesinger et al., 2014)	4	–	3	–	–	–	–	No statistically significant differences in axial nor in medial (Varus) stem migration compared to a control group. Locking plate fixation of a PFF with stable cemented prosthesis did not lead to cement mantle failure.
(Brand et al., 2014)	3	-	1	-	-	-	-	Intraprosthetic fixation provided significantly higher failure loads compared to unicortical locked-screw plating. Significant increase in primary stability without weakening the implant-cement-femur-model that could lead to early weight-bearing patient mobilization.
	2 (h)	-	1	-	-	-	-	
(Lenz et al., 2014)	3	1 Ce	2	-	-	-	-	Proximal bicortical fixation using LAP-LCP construct improves stability of proximal plate fixation in Periprosthetic fractures. Cerclage cable -screw combination is valuable alternative, especially in osteoporotic bone. Cerclages should be used in combination with at least one additional screw to achieve stable fixation.
	-	4 Ce	2	-	-	-	-	
	4	-	2	-	-	-	-	
	3(f - 2 BC)	-	2	-	-	-	-	
(Hoffmann et al., 2014)	6 (e - BC)	-	3	-	-	-	-	Proximal bicortical screw placement achieved maximal load to failure and maximal torsional/sagittal bending stiffness. Addition of unicortical screws increased axial stiffness when cable fixation used. Lateral bending not affected by differences in proximal fixation.
	4	1 W	3	-	-	-	-	
	-	3C	3	-	-	-	-	
(Sariyilmaz et al., 2014)	2	2C	4	-	-	-	-	Medial strut allograft with plate fixation showed highest stiffness and failure load values and least displacement at fracture site. Suggesting it is mechanically superior method in B1 type PFF fixation treatment near tip of THA
	2	-	2	Med	150	2C	2C	
	2	-	2	Ant	150	2C	2C	
(Griffiths et al., 2015)	5	2C	5	-	-	-	-	LAP-LCP construct significantly stiffer than cable construct under axial load with bone gap. Offers better axial stiffness compared to cable construct.
	2 (f – 4 BC)	-	5	-	-	-	-	
(Graham et al., 2015)	3	-	3	-	-	-	-	Unicortical screws showed stiffest form of fixation at all loading angles. Suggests that cables result in the proximal screws being pushed into the bone as it is applied, causing screws to loosen their fixation to bone. Fracture gap and no gap model behave differently – degree of fracture reduction affects whole construct stability and bending behaviour of fixation.
	3	3C	3	-	-	-	-	
	-	3C	3	-	-	-	-	
(Gwinner et al., 2015)	4	-	5	-	-	-	-	Failure mode in unicortical group characterised by screw pull-out with no additional bone fracture or fissure. Proximal bicortical screw fixation showed no screw pull-out, and had higher cycles to failure. Bicortical group also showed significant superiority of screw purchase compared to unicortical screws. However, mode of failure resulted in severe commuted fracture patterns compared to the unicortical screws, which only resulted in screw pull-out.
	3 (e - BC)	-	5	-	-	-	-	
(Lewis et al., 2015)	-	3 Ce	-	-	-	-	-	Proximal bicortical screw fixation constructs (LAP + SS LCP, Synthes, PA, USA, and Ti NCB, Zimmer, IN, USA) showed higher maximum forces in torsional loading compared to cable, unicortical locking screws (LS), and
	4 LS	-	-	-	-	-	-	
	4 LS	2C	-	-	-	-	-	

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Table 2 (*continued*)

Authors	Test case: Experimental studies, Plate and strut fixation — Lateral plate fixation: Proximal: Unicortical Screw	Lateral plate fixation: Proximal: Cable/wire	Lateral plate fixation: Distal: Bicortical screw	Strut fixation: Position	Strut fixation: Strut length (mm)	Strut fixation: Proximal: Cable/wire	Strut fixation: Distal: Cable/wire	Results
	4 (2f - 4 BC each) (j) 6 (e)	- -	- -	- -	- -	- -	- -	cable + unicortical LS constructs. Cable constructs showed the lowest maximum forces, in both axial and torsional loading. Bicortical Ti NCB construct showed higher stiffness than the bicortical SS LAP-LCP construct in axial loading.
(Lenz et al., 2016a)	3 (f − 2 BC)/4(i) 3 (2f − 2 BC each) (j)	- -	2 / 2 2	- -	- -	- -	- -	LAP - Double LCP plate (Orthogonal) construct fixation showed significantly higher stiffness, cycles, and load to failure compared to LAP (x2) - single LCP plate construct. Additional locking plate enhances construct stability and increases construct stiffness compared to single plate fixed with two LAP.
(Lenz et al., 2016b)	2(GT), 3 3 (f – 2 BC)	- -	2 2	- -	- -	- -	- -	Hook construct showed significantly lower cycles and load to failure and fixation strength compared to LAP-LCP construct. Plate stiffness between the two groups were comparable in range. Use of subtrochanterical bicortical screw fixation is an effective fixation method in PPF than hook plate, and is less influenced by bone stock quality. Suggests that hook plate is reserved for PPF that requires stabilization of greater trochanter as it is highly BMD dependent.
(Moazen et al., 2016)	6 (e) 6 (e)	- -	4 4	- -	- -	- -	- -	Proximal bicortical screw fixation using far cortical locking screws can reduce overall effective stiffness of locking plates and increase fracture movement while maintaining overall strength of PFF fixation construct compared to bicortical screw fixation using locking screws. In unstable fractures alternative fixation methods such as long stem revision may be better.
(Wähnert et al., 2017)	2 (f-2 BC) 2/2(o)	- -	3 3/2(o)	- -	- -	- -	- -	Construct stiffness and cycles to failure significantly higher in double-plate construct compared to LCP-LAP construct.
(Lochab et al., 2017)	3 4 (2f)(l)	2C -	4, 2C (m) 4	Ant -	200 -	2 Ce (n) -	2C -	LCP-Allograft construct demonstrated higher stiffness values in compressive abduction, torsion, and medial-lateral four-point bending compared to the LAP-LCP construct. No differences identified between the two constructs in compressive flexion, anterior-posterior bending or load to failure tests.
Long stem vs short stem (Gordon et al., 2016)	Comparison of 4 groups, short stems versus long stems for their effectiveness, and locking plate fixation versus cerclage system: 1 – Long stem/Cerclage– (4 titanium cerclage bands and 2 stabilizers) 2 – Long stem/Plate –(NCB, 5 proximal unicortical screws and 4 distal bicortical screws) 3 – Short stem/Cerclage – (4 titanium cerclage bands and 2 stabilizers) 4 – Short stem/Plate–(NCB, 5 proximal unicortical screws and 4 distal bicortical screws)							Results indicate that for Vancouver B1 fractures, osteosynthesis with plate fixation has no biomechanical advantages over use of simple cerclage system – cerclage constructs demonstrated larger stiffness, larger strength, and more cycles to failure compared to plate construct. Revision with a long stem provides superior mechanical stability regardless of type of osteosynthesis fixation, thus suitable for Vancouver B1 fracture treatment. In short stem increased subsidence is seen in cerclage system compared to plating.
Plate and stem distance (Walcher et al., 2016)	Biomechanical performance to establish safe distance of plate from tip of femoral prosthesis. – Amount of plate to stem overlap or whether there is a safe gap between the stem and the plate end to reduce risk of future fractures. All NCB distal plates were attached to the femur at a defined distance from the stem to the plate at varying gaps from 80 mm gap to 60 mm overlap, in 20 mm increments. 40 mm gap – 40 mm overlap considered close group, and greater that 40 mm overlap or distance considered far group.							Strain increased with the decreased overlap or gap. All early failures occurred between 20 mm overlap and gap. Significantly less strain in the far group in both axial and torsional loading. Suggests that results can aid orthopaedic surgeons in plate positioning in Vancouver type-C PFF fixation. Reduction in post-operative complications

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Table 2 (*continued*)

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	
	Distal plates were fixed using 2 bicortical screws at the most proximal screw holes and 2 bicortical screws in the fourth and fifth holes. No femoral fracture was applied to simulate situation of healed periprosthetic fracture with implants still in situ.							by positioning the plate in a manner that may reduce stress risers that could lead to future fractures.
Cables								
(Frisch et al., 2015)	Biomechanical response of cerclage systems in fixation of intraoperative PFF in cementless THA. Four constructs compared, 1) CoCr (Cobalt-Chrome) cable 2) Hose clamp 3) Monofilament wire 4) Synthetic cable. No plate fixation used.							Metallic constructs with positive locking system performed best, supporting highest loads with minimal implant subsidence (both axial and angular) after loading. CoCr cable and hose clamp had highest construct stiffness and least reduction in stiffness with increased loading.
Screws								
(Kampshoff et al., 2010)	Effect of different screw implantation techniques on the integrity of local cement mantle and fixation strength of the screw. Using different kinds of locking screws. LCS (Synthes, Oberdorf, Switzerland) and NCB (Zimmer, Warsaw, IN, USA), with or without a flattened tip, implanted in unicortical and bicortical configurations using different core drill sizes							No unicortical screw induced cracks – unicortical screws can be implanted without damaging cement mantle. Screws with shortened tip, smaller flutes and double threads were significantly better for pull-out resistance. Bicortical screws have significantly higher pull-out resistance, but increase risk of local cement mantle damage. By increasing drill diameter, onset of cracks decreased, but so does pull out resistance.
(Konstantinidis et al., 2017)	Damage analysis of cement mantle after revision screw insertion; Influence of three variables on the incidence of crack formation in cement layer was studied; screw type. Cement mantle thickness, and position of screw relative to cement mantle and prosthetic stem. LCP plate shortened to have four holes only, and applied laterally to the femur at the level of the prosthetic stem (proximal). Three types of screw fixation methods investigated 1) four bicortical non-locking screws. 2) Four unicortical locking screws. 3) Four bicortical locking screws.							Crack formation found to be homogeneous for all three screw types. Screw positions relative to cement and prosthetic stem was divided into four categories: 1) No contact between screw and cement mantle. 2) Screw touches cement mantle and is partially within it. 3) Screw is entirely within the cement mantle. 4) Screw is in direct contact with periprosthetic stem. Significant association between screw position and incidence of cracks in the mantle. Probability of damage to the cement mantle increases significantly as the distance to the implanted prosthesis decreases. Screws in direct contact with the periprosthetic stem showed high incidence of crack damage to the cement mantle. Suggests that only the position of the screws relative to the cement mantle and/or periprosthetic stem exerts significant influence on the crack damage to the cement mantle.
Computational studies								
(Dubov et al., 2011)	Biomechanical performance of cable-screw position in repairing periprosthetic femur fractures near tip of a total hip implant (Computational study of the experimental study carried out by the same group Shah et al., 2011). Three different fixation methods described 1) Construct A- Cable-screw pairs in positions 1 and 2 proximal to fracture 2) Construct B-Cable-screw pairs in positions 1 and 3. 3) Construct C– Cable-screw pairs in positions 1 and 4. All three constructs had four bicortical screws distally.							Numerical computation at 1000 N axial load and 15° adduction of the femur showed higher axial stiffness and higher surface stress transfer to the femoral bone for Construct C (cable-screw pairs in extreme positions). Suggesting in the case of good bone stock, optimal fixation can be achieved by Construct C, and could potentially reduce bone re-fracture compared with A and B – as it is expected that the highest stiffness may achieve optimal mechanical stability. FEA showed excellent correlation with experimental results.
(Moazen et al., 2012)	Development and analysis of an FE model of a Vancouver type C clinical case comparing mechanical effects between two implemented fixation methods; wherein the initial fixation failed and replaced by a second fixation that led to healing. Initial plate fixation following THA used a femoral polyaxial plate (rigid fixation - POLYAX, DePuy, IN, USA), Refracture revision used a condylar blade plate (flexible fixation - Angled Blade Plate, Synthes, PA)							Rigid fracture fixation (polyaxial plate) with short bridging length in the case of PFF can suppress fracture movement that can prevent healing and may ultimately fail. In contrast use of a flexible fixation non-locking plate with larger bridging length promoted healing. Changing bridging length made a more substantial difference to stiffness and fracture movement than other parameters. Results suggest that a computational approach to compare stiffness and fracture movement of different fixation constructs can help determine optimum fixation method for PFF.

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Table 2 (*continued*)

Authors	Test case Experimental studies Plate and strut fixation Lateral plate fixation: Proximal (Unicortical Screw, Cable/wire); Distal (Bicortical screw) Strut fixation: Position; Strut length (mm); Proximal (Cable/wire); Distal (Cable/wire)	Results
(Chen et al., 2012)	Finite element analysis performed to study internal biomechanical forces during fixation of Vancouver type B1 periprosthetic fracture with an Ogden construct and four variations of this construct. 1) A - Proximally 3 wires, distally 2 bicortical screws. 2) B- Proximally 3 wires plus 2 unicortical screws, distally 2 bicortical screws 3) C-Proximally 3 wires, distally 2 bicortical screws plus 3 wires 4) D-Proximally 3 wires plus 2 unicortical screws, distally 2 bicortical screws plus 3 wires.	Results show that original basic Ogden construct (A) fixation has inferior outcome compared to other fixation methods. Addition of two screws above the fracture site (Construct B) visibly decreased displacement and stress. Additional wires fixed below fracture site do not noticeably decrease either von Mises stress or fracture displacement (Construct C). Better fixation power is achieved by using both proximal and distal screws in treating Vancouver B1 periprosthetic fractures after THA.
(Ebrahimi et al., 2012)	Experimental and computational study to predict overall stiffness and peak bone stress in the same femur after injury, repair, and healing, with respect to its intact condition. Four stages were described. 1) Stage 1 – intact femur 2) Stage 2 – mimicked femur with a hip stem 3) Stage 3 – mimicked 5 mm fracture gap repaired with plate and screws 4) Stage 4 – represented complete fracture union. FE model validated against experiments and re-analysed using clinical-level force of 3000 N	Stage 3 (immediate post-surgical scenario of periprosthetic femoral fracture fixation) showed least stable situation compared to stage 1, being the most vulnerable to re-injury; yielding the lowest stiffness and highest bone stress compared to stage 1 (Intact femur). Stage 4 (healed femur) showed rise in stiffness surpassing stage 1 and re-distribution of stresses back to femur itself compared to stage 3. Study highlights the potential adverse effects of stress-shielding and high stresses throughout the surgical process and even after fracture healing. Suggests a stiffness-matching strategy in future design of implants relative to the intact femur. At 1500 N, FE vs Experimental strains had excellent linear agreement.
(Moazen et al., 2013)	FE model of Vancouver type B1 PFF fixation within a stable stem with good bone quality developed. Effect of fracture stability on locking plate fixation performance quantified, and comparison of stainless steel (SS) and titanium (Ti) plate in stable and unstable fracture under two weight-baring conditions - 500 N and 2300 N, analysis carried out.	Stress and strain on the plate was higher in the unstable compared to the stable fixation. In the case of unstable fractures, it is possible for a single locking plate fixation to provide the required mechanical environment for callus formation without significant risk of plate fracture, provided partial weight bearing is followed. In cases where partial weight bearing is unlikely, additional biological fixation could be considered.
(Moazen et al., 2014)	Biomechanical performance of six different fixation methods for Vancouver B1 and B2 type fractures studied. 1) Use of eight-hole locking plate: fixed laterally using 3 unicortical screws proximally and 4 bicortical screws distally. 2) Ten-hole locking plate: 4 unicortical screws proximally and 4 bicortical screws distally. 3) Double locking plates: as with method 1 plus an additional anterior eight-hole locking plate fixed using three unicortical screws proximally and three bicortical screws distally. 4) Revision stem (201 mm): short stem used in method 1–3 replaced by a 201 mm long stem; the cement mantel was expanded medio-laterally to fit. 5) Revision stem (201 mm) and eight-hole plate: as with method 4 plus an additional eight hole locking plate fixed proximally with three unicortical screws and distally with one unicortical and two bi-cortical screws.6) Revision stem (241 mm): as with method 4 except stem extended by 40 mm. 1–3 represent PFF fixation methods for Vancouver B1 fractures, and 4–6 for Vancouver B2.	Indicate that in treatment of B1 fractures, a single locking plate is sufficient provided partial weight baring is followed. In B2 fractures, long stem revision and bypassing fracture gap by two femoral diameters are recommended. Long stem revision could be considered in all comminuted B1 and B2 fractures when considering risk of single plate fracture.
(Leonidou et al., 2015)	Comparing treatment methods for different bone quality – three models with different canal thickness ratio (CTR), representing poor, average, and good bone quality. Further three models were developed with angle fractures varying from unstable transverse (0°), short oblique (146°), and stable long oblique configuration (76°). Comparisons were also made on three different models with the fracture at the tip of the stem, 4 mm, and 14 mm below the tip of the stem.	Results suggest that in good bone quality and acute fracture angles, single locking plate fixation can be considered as an appropriate management method. Conversely poor bone quality and obtuse fracture angles alternative methods may be required as fixation might be under higher risk of failure. Suggests that orthopaedic surgeon should take into consideration the PFF topography and bone quality and not entirely rely on Vancouver classification to formulate a treatment plan.
(Avval et al., 2016)	Investigation into femoral density changes in response to bone fracture plate and hip implant; long-term behaviour of a femur in response to these implants and fixations were simulated. Bone mineral density changes evaluated for lateral plating and anterior plating (3 unicortical screws and 5 bicortical screws used).	Results showed that areas directly under the plate experienced severe bone loss (Up to $\sim -70\%$). Some level of bone formation ($\sim +110\%$) was observed in the vicinity of the most proximal and distal screw holes in both lateral and anterior plated femurs. In respect to bone remodelling response, anterior plating is not superior to lateral plating.

(*continued on next page*)

Table 2 (*continued*)

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	
(Wang et al., 2016) (p)	Biomechanical performance of three different PFF fixation methods for Vancouver type B1 fractures in normal and osteoporotic bone was examined via FE model. 1) Double circle cable 2) Traditional locking titanium plate (LCP) 3) Multidirectional locking plate. Stress distribution, stiffness, maximum stress and relative displacement were compared under same axial and torsional loading using FEA.							Multidirectional locking plate system showed higher stability and stiffness, more even stress distribution under the same axial and torsional loading in both normal and osteoporotic bone than double circle-cable and traditional locking titanium plate for Vancouver B1 periprosthetic fractures.

Key to content: C, cable; Ce, cerclage; W, wire; BC, Bicortical screws; LS, Locking screws; GT, greater trochanter; Ant, anterior; Lat, lateral; Med, medial.
(a) Only hip stem inserted.
(b) Distal retrograde femoral nail used, two locking screws used proximally, and one distal.
(c) Additional 188 mm anterior locking plate was used.
(d) In this instance 5 fixed angle self-cutting locking screws used to affix plate to lateral femoral condoyle.
(e) In this instance, bicortical screws were used for proximal fixation instead of unicortical screws.
(f) Locking attachment plate (LAP) was used, additional bicortical screws used for proximal fixation.
(g) In this instance, proximal fixation involved two 4 mm variable angle screws (VAS) passing the prosthesis and two 5 mm VAS passing the stem, to achieve bicortical fixation.
(h) In this instance, two intraprosthetic screws were implanted in place of three unicortical screws.
(i) In this instance, orthogonal plating was implemented – two plates were used, one placed laterally (with LAP) and one anterior.
(j) In this instance two LAP was used.
(k) In this instance, LCP length was made short, and 70 mm long segments of cadaveric femoral diaphysis was used.
(l) In this instance, two LAP were applied using 2 transcortical locking screws in the outer most holes.
(m) In this instance, distal fixation also included 2 cables in conjunction with the bicortical locking screws.
(n) Author didn't specify number of cerclage cables with crimps system was used, but based on data available and from images we think 2 cables for proximal and 2 for distal fixation were used in allograft strut fixation.
(o) In this instance, an additional 5 hole-LCP plate was fixed anteriorly to the femur in conjunction with a lateral LCP plate.
(p) Study didn't specify specific brand of plate, but we hypothesize that the traditional locking titanium plate is the LCP plate by DePuy Synthes and Multidirectional locking plate is the NCB plate by Zimmer.

motion across the fracture. However, the recent literature has indicated that biomechanically, better plate fixation is not dependent on the rigidity of a structure alone (Lujan et al., 2010; Moazen et al., 2012).

Results indicate that better plate fixation can be achieved by:

1) Fixation with screws, or screws with cables, in preference to cables and wires. (Chen et al., 2012; Graham et al., 2015; Lenz et al., 2013; Lever et al., 2010; Shah et al., 2011; Wang et al., 2016)
2) Proximal fixation using bicortical screws instead of unicortical (Gwinner et al., 2015; Hoffmann et al., 2014; Konstantinidis et al., 2010; Lenz et al., 2014; Lewis et al., 2015); or addition of a LAP or LAP-like construct (Griffiths et al., 2015; Lenz et al., 2012b, 2016a)
3) Double plating (Use of additional plate in fixation) (Choi et al., 2010; Lenz et al., 2016a; Wähnert et al., 2017); or strut (Lochab et al., 2017; Sariyilmaz et al., 2014)
4) Intraprosthetic fixation (Brand et al., 2014)
5) Use of long stem revision. (Gordon et al., 2016; Moazen et al., 2014)
6) Larger bridging length (Moazen et al., 2012; Walcher et al., 2016)
7) Application of far cortical locking technology (Moazen et al., 2016)
8) Positioning of Screws or Cable-screws (Dubov et al., 2011; Konstantinidis et al., 2017)

Many authors reported that in cases of good bone stock (typically Vancouver B1 type fractures), fixation with plate and screws provided most stability. Shah et al. (2011) showed that plate-screws with additional proximal cable fixation were the best choice for healthy bone; in cases of osteoporotic bone, a plate without proximal holes and proximal fixation with only cables was supported. A similar result to Shah et al. (2011) was reported by Demos et al. (2012). However, Graham et al. (2015); found that when unicortical screws are used in conjunction with cables, results in proximal screws being pushed into the bone as it is applied, causing screw loosening fixation to the bone. Furthermore, Gordon et al. (2016) showed that osteosynthesis using plate fixation offered no biomechanical advantages over the use of a simple cerclage system. They suggested that revision with a longer stem would provide superior mechanical stability regardless of the type of osteosynthesis fixation. A similar result could be seen in the computational study by Moazen et al. (2014) who also suggested long stem revision in both B1 and B2 fractures when considering the risk of single plate fracture. However, Lewis et al. (2015) found that cable constructs failed in torsion by the femur rotating and loosening within the cables. The constructs also had significantly less maximum force compared to all other constructs in both torsional and axial loading. They found that unicortical, and unicortical with cable specimens tended to fail by catastrophic fracture of the femur due to cracks typically stemming from insertion sites of the screws. Clinically, many studies have reported that cerclage wiring alone has a high failure rate, and proximal unicortical screws in dynamic compression plates, while more stable than cerclage wiring along, are also inadequate (Schwarzkopf et al., 2013).

In regards to the likehood of cement mantle failure when using screws; Giesinger et al. (2014) found that plate fixation of PFF using proximal screws with a stable cemented prosthesis didn't lead to cement mantle failure. In contradiction, Kampshoff et al. (2010) found that use of screws with shortened tip, smaller flutes and double threads, showed better pull out resistance, but increase the risk of cement mantle failure. Bicortical screws had significantly superior construct stability and pull-out resistance when compared to unicortical screws; however bicortical screws also increased the risk of local cement mantle failure. Additionally, Gwinner et al. (2015) also showed that the mode of failure was more catastrophic in proximal bicortical screw fixation; with severe comminuted fracture patterns occurring, compared to screw pull-out with less bone damage seen in the unicortical screws group. Konstantinidis et al. (2017) showed that the probability of cement mantle damage increases significantly the closer it is to the implanted prosthesis. Direct contact of screws with cement mantle resulted in higher incidence of cement mantle crack damage. Lever et al. (2010) also noted that in a clinical situation; cortical screw tips could nick the lateral surface of the femoral stem, resulting in metallic wear debris forming during daily activities. Furthermore, some of the mechanical stiffness measured may be due to screw impingement into cement; thus slightly overestimating stiffness levels that could be achieved in vivo.

Demos et al. (2012) found that there was no difference between locking screws and non-locking screws. Many studies using bicortical screws or a LAP construct for proximal fixation showed higher rigidity compared to unicortical screws. However, there were contradictions; Wähnert et al. (2014) found that use of LAP did not provide the most stability as it caused a less rigid plate. Moazen et al. (2016) found that distal far cortical locking screws can reduce the overall effective stiffness of locking plates and increase fracture movement. They also found that the overall strength of the PFF fixation construct was maintained when compared to bicortical fixation with distal locking screws. However, in unstable fractures, alternative fixation methods may be a better treatment option.

Fracture gap and bridging length were also found to influence the stability of a fixation construct; Graham et al. (2015) found that fracture gap model behaves differently to the no gap model and that the degree of fracture reduction affects whole construct stability and bending behaviour of bone. Walcher et al. (2016) showed increased strain with decreased over-lap or gap of the plate to stem. An FE analysis of a clinical case carried out by Moazen et al. (2012) suggested that implementing a fracture plate with a larger bridging length may promote healing compared to a plate with shorter bridging length, displaying the importance of plate positioning in Vancouver type C PFF fixation.

4. Discussion

A total of 30 experimental and 9 computational studies published since 2010 relating to PFF were reviewed in this paper. Several advancements and differences were summarised compared to past studies; however, some issues still remain. Four main issues that were highlighted in the previous review (Moazen et al., 2011), remain important; briefly, they are as follows;

1) Lack of standardization in methods used.
2) Variation in the level of sophistication in both experimental and computational models; in experimental studies, there is typically a trade-off between accuracy and consistency. In computational studies, the balance is between realism and time for development and processing.
3) Biomechanical studies are primarily concentrated on Vancouver type B1 fractures. With less focus on type A and C.
4) The relationship between results presented and the clinical situation needs to be better defined. Two main issues that are clinically important are, firstly the fracture heals, and secondly, the construct doesn't fail.

Table 1 shows that there is still a lack of standardization for testing PFF. Current experimental studies still show a lack of consistency in both testing procedures and measurements. This makes it difficult to make direct and conclusive comparisons between findings. Biomechanical testing comparing the two main plates for PFF fixation (The LCP by DePuy Synthes, and NCP by Zimmer) typically use the same NCB plates but different DePuy Synthes plates, or plates of different lengths, making it difficult to make direct comparisons between the different studies and plates used (Konstantinidis et al., 2010; Lever et al., 2010; Lewis et al., 2015; Wähnert et al., 2014).

Modelling of the clinical problem is not easily done because each PFFs case is different. The best approximation to the clinical challenge in either experimental or computational studies is made by taking into account all different parameters that affect the clinical result. Thus modelling appropriate anatomic region and the stability of the fracture, bone stock and the stability of the implant, and patients' characteristics as demographics are the important basic requirements that we have to consider when making the best experimental or computational study. Most of the biomechanical studies still concentrate on Vancouver type B1 fractures, with no studies conducted on Vancouver type A; and only one experimental and one computational study (Walcher et al., 2016; Moazen et al., 2012;) on Vancouver type C fractures. This may be due to the fact they are clinically less prevalent, and more easily treated (Brand et al., 2015; Capone et al., 2017; Fleischman and Chen, 2015; Lever et al., 2010). Vancouver type B2 and B3 fractures are more challenging to conduct experimentally, with some studies using a fracture gap to mimic an unstable fracture (Choi et al., 2010; Giesinger et al., 2014; Graham et al., 2015; Griffiths et al., 2015; Konstantinidis et al., 2010; Lochab et al., 2017; Sariyilmaz et al., 2014; Shah et al., 2011; Wähnert et al., 2014, 2017). However, it is important to note that clinically, type B2 and B3 fractures are not only unstable fractures, but the stem itself is unstable, meaning the stem has lost the connection with the surrounding bone and requires additional revision or treatment, typically with a longer stem (Schwarzkopf et al., 2013). In addition, there are still several contradictions to which treatment method is the 'optimum'. The lack of standardization may be attributed to inadequate understanding of treatment and differentiation between stable and unstable prosthesis; as failure to identify an unstable implant may lead to treatment failure if osteosynthesis rather than revision surgery is performed (Schwarzkopf et al., 2013). Thus it is important to also have biomechanical models that differentiate between stable and unstable prosthesis.

A distinct difference seen in present studies compared to older ones is the reduced use of struts and increased use of the LAP and double plating in the experimental studies. Of the 30 experimental studies, only three cases used struts in their biomechanical experiments. This is a stark contrast in comparison to the previous review, where of the 14 experimental cases reviewed, eight studies used struts. This is in place of the introduction and increase in testing the biomechanical performance of double plating and the use of a LAP or similar construct. Clinically, there is not much data regarding the use of the LAP, however, there have been some reports of acceptable outcomes from using an LAP to manage PFF with a well-fixed stem (Kim et al., 2017a, 2017b) or when stability of plate is insufficient (Kammerlander et al., 2013). Despite the significant decrease in the use of struts in biomechanical testing; clinically struts in conjunction with plate fixation are still widely used for PFF fixation treatment, with some studies showing positive clinical outcomes (Barden et al., 2003; Khashan et al., 2013; Kim et al., 2017a, 2017b).

Another interesting and perhaps important development is the increased use of computational modelling in simulating PFF and its fixation methods; possibly because researchers have realised the added value of using this approach. The review of Moazen et al. (2011) reported only two computational studies; here nine cases were reviewed, ranging from simple models to more complex situations such as investigating femoral density changes in response to bone fracture plate and hip implant (Avval et al., 2016), or modelling clinical cases (Moazen et al., 2012). While experimental studies remain the key component of these biomechanical studies, there is no doubt in the value that computational studies bring to testing and evaluating effective fixation methods in a greater range of fracture scenarios and more complex situations. Several computational studies were corroborated against experimental results (Dubov et al., 2011; Ebrahimi et al., 2012; Lenz et al., 2013; Shah et al., 2011) demonstrating their validity. However, whether clinicians or researchers on the whole have confidence in the outcome of computational results over experimental is still a matter of debate.

From a clinical point of view, the crucial outcome is that the fracture heals, return to pre-injury function, and the construct itself doesn't fail. Much of the research has hence focused on construct stiffness; and this is still the case in many of the present studies which highlight the higher construct overall stiffness as the "better" fixation; this is despite studies shown by several groups that locking plates (depending on how they are applied) lead to overly rigid fixations that can supress callus formation (Lujan et al., 2010; Moazen et al., 2012). This can be partial since we still do not know the overall stiffness of PFF fixations in situ, and that can be widely different to the way that they have been tested biomechanically. An interesting development in response to this has been the introduction of far cortical locking technology (Bottlang et al., 2009; Bottlang and Feist, 2011); commercially named *Motion*Loc, and can be used in Zimmer NCB plates. The screws lock into the plate and bypasses the near cortex, reducing the effective stiffness of locking plates compared to standard locking screws that are secured in both near and far cortices, limiting the rigidity of the fixation and supporting callus formation. While there are some clinical data available that show some positive results in the use of far cortical locking screws, particularly in distal periprosthetic femoral fractures (Bottlang et al., 2010; Ries et al., 2013; Wang et al., 2018); none, to the best of our knowledge have reported any clinical data regarding PFF after THA specifically.

Thus more clinical data regarding the use of these new plating methods or technologies needs to be reported to better translate and validate experimental and computational data. In this review, only one study (Moazen et al., 2016) focused on far cortical locking screws; again demonstrating the importance of computational studies in testing more complex scenarios. There has been evidence of experimental and computational studies being translated into clinical practise; studies by Gordon et al. (2016) and Moazen et al. (2014) advocated long stem revision in cases of B1 and B2 fracture treatment; this aligns with clinical data of patients with failed B1 fracture osteosynthesis showed that revision to a long stem provided good results (Cassidy et al., 2018; Randelli et al., 2018). Cassidy et al. (2018) suggested that revision rather than repeat fixation, regardless of how well fixed the stem appears would be optimum.

Present biomechanical studies used either cemented or uncemented hip stems; however, no studies made a no direct comparison between the two and its effects on the biomechanical performance of the fixation construct. Thus it is difficult to say whether or not the literature for one prosthesis implantation method can be applied to the other; consequently whether subsequent treatment methods derived from biomechanical studies where most studies used cemented prosthesis (22 out of 30 experimental, and 4 out of 9 computational), can be used for uncemented and vice versa. Thus the relationship between cemented versus uncemented hip prostheses and its fixation methods needs further research in order to provide more clinically relevant data, this is particularly paramount as the use of uncemented stems is increasing for THA (Kim et al., 2015; Philippe et al., 2015).

It is also important to consider that clinically, there is different behaviour between cemented and uncemented THA. Failure is more likely to occur in patients who underwent uncemented THA (Wyatt, 2014). However, Wyatt (2014) noted that a 13-year long follow up of THA cases showed that there was no significant difference in revision between implantation methods; suggesting the higher early revision rate may be due to intraoperative events from an inexperienced surgical team. However, this contradicted the Swedish registry results, which show that uncemented stems are revised twice as frequently as cemented stems during the first five years, and that cemented stems were ten times less likely to require revision for periprosthetic fracture.

The Vancouver classification system for treating PFF was originally developed for THAs with cemented femoral components (Duncan and Masri, 1995), and does not differentiate treatment between cemented and uncemented hip stems; thus raising the question of can direct comparisons for treatment of PFF to be made between cemented and uncemented prostheses. While the Vancouver classification system is reported to be reliable and valid, it is difficult to strictly apply rules for treatment in some cases as there is no objective standard to assess the bone quality or prosthetic stability, and is an arguable drawback of the Vancouver classification system (Park et al., 2011). Another caveat of this system is that it cannot differentiate between stable and unstable prosthesis easily, which is one of the most crucial parts of treatment. Thus it would be useful if different types of PFF models that are not easily recognised in the clinical setting could be simulated experimentally and computationally.

Another critical issue that needs to be discussed is the lack of osteoporotic bone models; in most studies, the bone quality that was simulated experimentally and computationally could be considered normal healthy bone stock, and not osteoporotic bone seen in patients with high risk of PFF; with only 3 studies using osteoporotic bone models, two of which did compare bone quality (Lehmann et al., 2010; Leonidou et al., 2015; Wang et al., 2016). The same issues can be raised as to whether or not results from current biomechanical studies can be translated into clinical cases, and thus further studies using osteoporotic bone models is required.

While these issues still exist, it is important to recognize the improved strides made towards understanding the underlying issues of PFF and its treatment methods. With the increased interest in PFF, many of the current studies show a higher level of sophistication in their methods used. This is reflected in many of the studies showing more consideration and highlighting parameters that may affect PFF that were not previously tested in earlier studies (Moazen et al., 2011); such as fracture gap (simulating unstable fracture), cement mantle integrity, bridging length, and plate type used. Comparison of biomechanical performance between constructs in different situations was also studied (e.g. before fracture, fracture with a plate, healed fracture gap - Giesinger et al., 2014; Graham et al., 2015; Griffiths et al., 2015). The interest in improving PFF fixation has also seen the development of new commercially available plates specifically designed for PFF. Several studies have made comparisons on the two major plates used for PFF; the LCP and NCL plate, as well the LAP (Griffiths et al., 2015; Lenz et al., 2012b, 2016b).

5. Conclusion

This review follows our earlier review of experimental and computational modelling of PFF fixation (Moazen et al., 2011). While there have been improvements in the way biomechanical testing of PFF fixation is carried out, the lack of literature to address the situations described above hinders its translation into clinical practise. In particular, the optimal treatment for Vancouver type B fractures remains controversial with experimental data not always reflecting actions occurring in situ. This is primarily due to available literature; which mainly consists of small to medium-sized heterogeneous case studies that offer little comparative evidence (Fleischman and Chen, 2015). With the incidence of PFF expected to rise, a consensus on biomechanical testing methods, and subsequent optimum treatment methods need to be achieved. The effect of cemented versus uncemented prosthesis on fixation methods needs further research, as well as the development of more osteoporotic bone models. An effective method can be seen in using experimental methods in conjunction with computational methods to help bridge this gap and develop more clinically relevant models.

Conflict of interest

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

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