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Unique stability of femoral neck fractures treated with the novel biplane double-supported screw fixation method: A biomechanical cadaver study

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ABSTRACT

Osteosynthesis of femoral neck fractures is related to 20–46% complication rate. Filipov's novel method for biplane double-supported screw fixation (BDSF), using three cannulated screws, has demonstrated excellent clinical results since 2007. Its two calcar-buttressed screws are oriented in different coronal inclinations with steeper angles to the diaphyseal axis and intended to provide constant fixation strength under different loading situations.

The aim of this study was to biomechanically evaluate BDSF fixation strength and compare it with the conventional fixation (CFIX) using three parallel cannulated screws.

Methods: Eight fresh-frozen and six embalmed human femoral pairs with simulated AO/OTA31–B2.2 fracture were fixed applying either CFIX or BDSF. Quasistatic tests were performed in anteroposterior (AP) bending, followed by axial quasistatic, cyclic and destructive quasistatic tests run in 10° flexion with 7° or 16° varus specimen inclination.

Results: Initial axial stiffness was significantly higher for BDSF in comparison with CFIX at 7° inclination ($p = 0.02$) and not significantly different between BDSF and CFIX at 16° inclination. Compared with the intact state, it decreased significantly at 7° inclination only for CFIX ($p = 0.01$), but not for BDSF. Interfragmentary displacement during cyclic testing was significantly smaller for BDSF than CFIX at 7° inclination ($p \leq 0.04$) and not significantly different between BDSF and CFIX at 16° inclination. Failure load did not differ significantly between BDSF and CFIX at both inclinations.

Conclusions: Femoral neck fracture stability can be substantially increased applying BDSF due to better cortical screw support and screw orientation. Having two calcar-buttressed screws oriented in different inclinations, BDSF can enhance constant stability during various patient activities. The more unstable the situation, the better BDSF stability is in comparison to CFIX.

Introduction

The rate of femoral neck fractures, one of the most common traumatic injuries in the elderly, increases constantly among the aging population [1,2]. Treatment complications originate from insufficient reduction, unstable fixation, and poor-quality osteoporotic bone [3,4]. Cannulated screws are often used; however, this osteosynthesis is associated with poor results in 21–46% of the clinical cases [5,6]. Screw configuration has been investigated in several biomechanical studies [7–14]. Currently, there is rather a divergence of views and concepts. The majority of authors recommend placement of the distal screw so that it is supported by the distal femoral neck cortex [4,8–10,14–20], which is traditionally called the "calcar", although this is not the true anatomic calcar [21]. Central screw placement on the lateral view is advised in some papers [19], while other authors suggest peripheral placement [8,10,18]. Secured posterior cortical screw support is also recommended [9,18,22]. It is widely accepted that the screws should be placed parallel to each other [4,8,9,17–19,22]. However, the dictum of parallel placement has not been proven [20] and some authors prefer divergent placement on the lateral view [14,20,23]. The inverted triangle configuration is usually favoured because it provides higher stability [7,8], and screw insertion at

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higher angles relative to the diaphyseal axis seems to achieve better fixation strength [12].

The current conventional method for femoral neck fracture fixation uses three parallel cannulated screws, but this does not always provide appropriate fixation strength (Fig. 1). This is especially true if osteoporosis is present, and poor results might subsequently develop. The initial interfragmentary compression of these constructs is frequently insufficient and therefore unable to ensure stability in osteoporotic bone. Moreover, the constructs could be occasionally instable with regard to varus stresses, anteroposterior bending and torsion because of the screws inserted pretty close to each other with entry points localised in the rather thin section of the cortex near to the greater trochanter, lacking appropriate lateral cortical support (Fig. 1).

When cannulated screws are used to fix a femoral neck fracture with osteoporosis, intraoperative interfragmentary compression alone may not ensure adequate stability during the healing process because it could soon be lost on fracture impaction. Construct stability can be considerably increased if cannulated screws with better cortical support in the distal fragment are used, acting more effectively as console beams with overhanging ends.

Filipov's novel method for biplane double-supported screw fixation (BDSF) can increase fixation stability, demonstrates a high degree of reproducibility during its standardized surgical procedure, and has been clinically applied since 2007 [24]. The innovative concept of biplane screw positioning makes it feasible to place three cannulated screws at steeper angles to the diaphyseal axis in order to improve their beam function and cortical support. The three screws are laid in two vertical oblique planes that medially diverge towards the femoral head on lateral view (Fig. 2). The distal screw is placed in the dorsal oblique plane with additional support by the posterior femoral neck cortex. The middle and proximal screws are oriented in the ventral oblique plane.

The entry points of the screws, which are placed with steeper angles relative to the diaphyseal axis, are located much more distally within the thicker cortex of the proximal diaphysis. BDSF uses two calcar-buttressed screws: the distal and the middle ones with different coronal inclinations of 150–165° and 130–140°, respectively. Each of these screws is placed with the following two supporting points (pivots) in the distal fragment: the *medial* supporting point on the distal femoral neck cortex, and the *lateral* supporting point at the screw-entry point into the lateral diaphyseal cortex. The distal screw has an additional third supporting point on the posterior femoral neck cortex. The two calcar-buttressed screws are oriented in different coronal inclinations in order to maintain constant stability during various physical activities. Their medial supporting points are located 10–20 mm apart, thereby distributing the axial load over a larger cortical area. The enhanced cortical support and increased angle improve the beam function of the calcar-buttressed screws when standing, whereas the proximal screw stabilises the upper neck under tensile stress. In addition, the distal screw, with its three supporting points, provides improved beam resistance to AP bending forces (e.g., rising from a chair), while the two anterior screws hold the side under tension.

The aim of this study is to evaluate biomechanically the fixation strength provided with the novel BDSF method in comparison to the conventional fixation (CFIX) for treatment of femoral neck fractures with three parallel cannulated screws.

Hypothesis. From biomechanical point of view, BDSF provides superior stability compared to CFIX.

Materials and methods

Specimens and study groups

Eight fresh-frozen (20 °C; 3 female and 5 male donors; mean age 72.4 years; range 42–76 years) and six embalmed pairs (2 female and 4 male donors; mean age 64.2 years; range 60–71 years) of human cadaveric femora were used in this study. Conventional AP and mediolateral (ML) radiographs were taken to confirm the absence of preexisting pathology in all specimens. Bone mineral density (BMD) was defined using dual-energy X-ray absorptiometry (DEXA) measurements (Lunar Prodigy Primo; GE Lunar, Madison, WI, USA) of the femoral neck and greater trochanter regions.

Each fresh-frozen (FRZ) pair was split and assigned to two study groups, CFIX-FRZ and BDSF-FRZ, to be instrumented applying CFIX

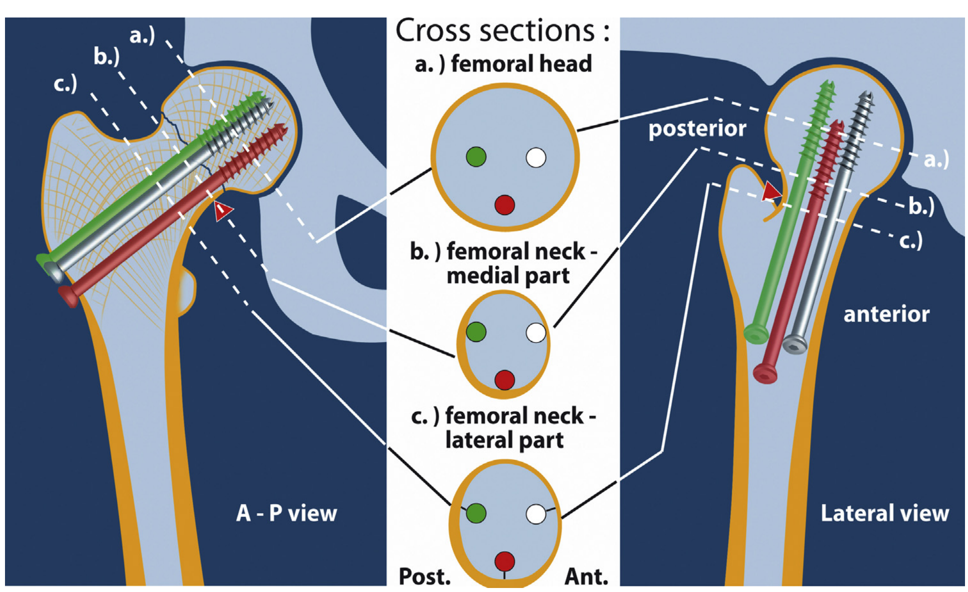

Fig. 1. Schematic of the conventional method with three parallel cannulated screws. Only one distal calcar-buttressed screw is used. Its contact point on the calcar is at the level of the medial part of the femoral neck (cross section b). The parallel screw orientation allows placement angles of 120–130° maximally. The screws are too close to each other.

[25] or BDSF [24], respectively. The same procedure was performed with the embalmed (EMB) femoral pairs which were assigned to another two groups CFIX-EMB and BDSF-EMB for instrumentation with CFIX and BDSF, respectively. The name of each study group reflected the specimen type and fixation method. The two pairs of groups, each one with either fresh-frozen or embalmed specimens, randomised with regard to left and right bone on BMD basis, were then selected for two different biomechanical testing procedures as described below.

The use of fresh-frozen and embalmed femora in the current study was based on literature data demonstrating that both bone types have similar mechanical characteristics and are recommended for biomechanical testing of orthopaedic and trauma devices [26–28]. Consequently, using these two types of femora was not expected to be a confounder for different mechanical behaviour between the study groups.

Specimen preparation

The FRZ specimens were thawed at room temperature for 24 h prior to preparation and biomechanical testing. The soft tissue was stripped off, and all femora were cut to 25 cm and distally embedded to a height of 70 mm in metal cylinders (ø50 × 70 mm) using fast-tightening epoxy (Hilti Hit-Hy200-A, Hilti, Waiblingen, Germany). Prior to fracturing and instrumentation, all specimens underwent two nondestructive biomechanical tests described below.

Subcapital AO/OTA 31-B2.2 fractures (Pawels II) were simulated in all specimens by performing standardised osteotomy perpendicular to the femoral neck axis between the subcapital and transcervical line. A mechanical hand saw was used (Dynagrip; Stanley, New Britain, CT, USA) to ensure identical osteotomy inclinations in each femoral pair [9,29,30].

Femora in the CFIX-FRZ and CFIX-EMB groups were instrumented under radiological control using standard operation technique [25]. After anatomical reduction, three 2.8-mm guidewires were inserted along a parallel guide at 130° with respect to the femoral shaft in an inverted triangle configuration, then 5.0-mm holes were drilled along the guidewires. The lateral cortex was then tapped to 7.3 mm and three partially threaded self-tapping 7.3-mm steel cannulated screws (DePuy Synthes, Solothurn, Switzerland) were inserted parallel to each other with clinically relevant interfragmentary compression at <5 mm subchondrally with <5° deviation so that the distal and posterior screws touched the distal and posterior neck cortices, respectively (Figs. 1 and 3).

The specimens in the BDSF-FRZ and BDSF-EMB groups were fixed after anatomical reduction using three partially threaded self-tapping 7.3-mm steel cannulated screws (DePuy Synthes, Solothurn, Switzerland) according to the BDSF method [24] under radiological guidance. First, a 2.8-mm guidewire for the distal screw was inserted with a lateral entry point at the median line of the lateral cortex at 4–6 cm distal from the lower border of the greater trochanter. The guidewire was directed posteroproximally to the dorsal third of femoral head at 150–165° towards the diaphyseal axis on the coronal plane (AP view), tangentially touching the distal neck cortex (on AP view) and posterior neck cortex (on lateral view) (Figs. 2 and 3). Second, a 2.8-mm guidewire for the middle screw was inserted with an entry point in the posterior third of lateral cortex at 2–4 cm proximal from the distal guidewire, depending on the caput-collum-diaphyseal (CCD) angle. This guidewire was directed anteroproximally to the frontal third of the femoral head (on lateral view) and to its distal third (on AP view) at an angle of 130–140° to the diaphyseal axis, thereby tangentially touching the distal neck cortex. Third, a 2.8-mm guidewire for the proximal screw was inserted with an entry point in the posterior third of the lateral cortex at a distance of 1.5–2.0 cm proximal from the middle guiding wire and parallel to it. This guidewire was directed to the frontal third of the femoral head (on lateral view) and to its proximal third (on AP view) at an angle of 130–140° to the diaphyseal axis. Drilling was then performed using a 5-mm cannulated reamer, followed by overdrilling of the middle screw hole in the lateral cortex using a 7.3-mm cannulated reamer and the insertion of the middle and proximal screws, thereby achieving clinically relevant interfragmentary compression, but without impaction to avoid additional frictional stability. Drilling was then performed using a 5-mm cannulated reamer for the distal screw, followed by 7.3-mm overdrilling of its hole in the lateral cortex. Finally, the distal screw was inserted. All three screws were inserted <5 mm subchondrally. No screw was placed in the central zone of the neck on the

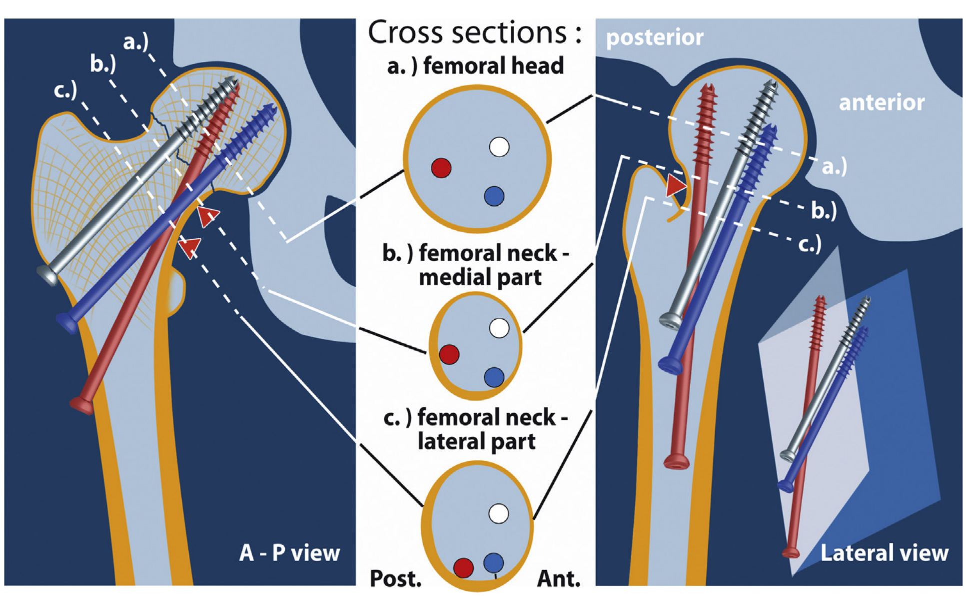

Fig. 2. Schematic of biplane double-supported screw fixation (BDSF). Two calcar-buttressed screws are used. The distal screw (red) touches the calcar in the lateral part of the femoral neck (cross section c). In addition, it is with cortical support on the posterior cortex of the neck (cross section b). The middle screw (blue) also touches the calcar in the middle part of the neck (cross section b). The screws have solid lateral supporting points in the lateral diaphyseal cortex. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

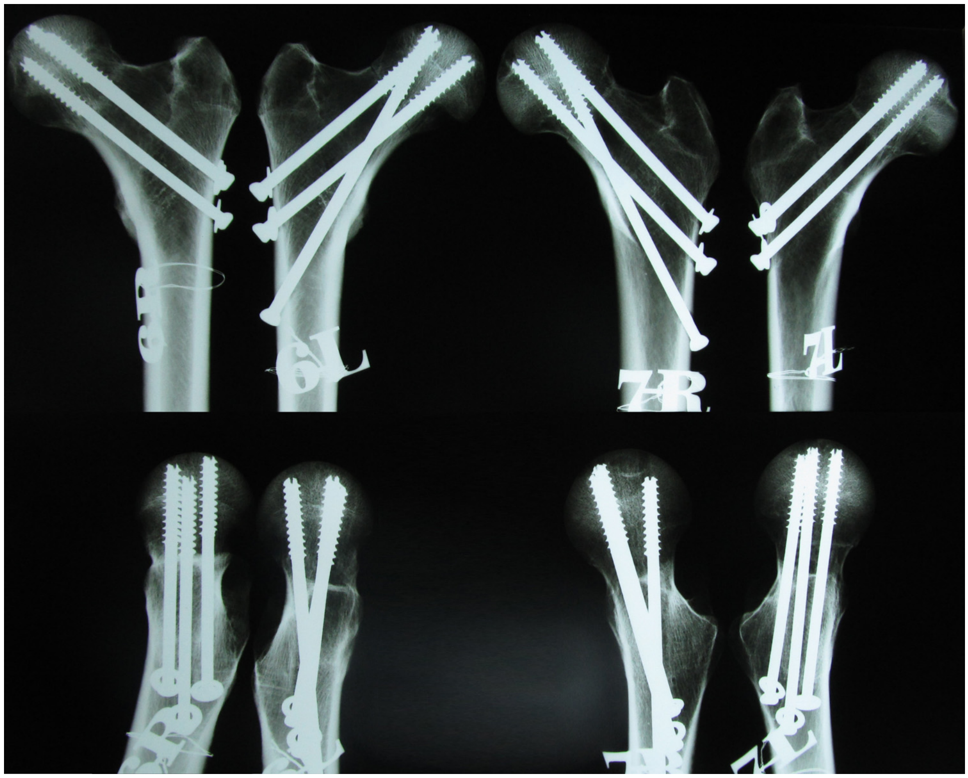

Fig. 3. X-ray radiographs of specimens 6R, 6L, 7R, 7L (FRZ) prior to mechanical testing. AP view (upper) and lateral view (lower).

lateral view. The proximal and middle screws were parallel to each other with $<5^\circ$ deviation.

Biomechanical testing

Biomechanical testing was performed using an Instron 1185 electromechanical test system (Instron, Canton, MA, USA) with a 100-kN load cell. Transducer sensor (WA-50L; Hottinger Baldwin Messtechnik, Darmstadt, Germany) was used to measure overall displacement.

Prior to osteotomy and instrumentation, all intact femora were subjected to nondestructive testing to provide baseline reference values for their stiffness as follows. First, a nondestructive quasistatic AP bending test was performed on all nonfractured (i.e. intact) femora that were horizontally mounted with their distal end in a jig fixed to the test frame [11] (Fig. 4). To avoid unnecessary posterior femoral shaft bending, each specimen was posteriorly supported by the test frame distal to the lesser trochanter at 90 mm from the load application point. The quasistatic ramped load (10–400 N) was applied to the anterior part of the femoral head at a rate of 5 mm/min through a cone-like concavity, simulating the acetabulum, and mounted on the machine actuator with possibility to glide horizontally (with lubrication).

Second, nondestructive quasistatic axial compression testing was performed on all nonfractured femora [13,31]. The FRZ and EMB specimens were tested at 10° flexion with 7° or 16° varus inclination, respectively. The proximal load application hardware and loading protocol were identical to the previous AP bending test. The specimens were distally mounted to a jig that was fixed to the test frame as shown in Fig. 4.

After osteotomy and instrumentation, each specimen underwent the following four biomechanical tests: nondestructive quasistatic AP bending test, nondestructive quasistatic axial compression test, cyclic axial compression test and destructive quasistatic axial compression test. The nondestructive quasistatic tests in AP bending and axial compression were run using the same test setups and loading protocols, as described above for the respective nonfractured specimens. The cyclic axial compression test was performed over 1000 cycles at 1-Hz and 100/1000 N valley/peak loads keeping the test setup from the previous quasistatic test in axial compression. Interfragmentary fracture displacement along the diaphyseal axis during cyclic testing was registered using a laser sensor (OptoNCDT-1401; Micro-Epsilon Messtechnik GmbH & Co. KG, Ortenburg, Germany) that was attached to the diaphysis of the specimen and distal to the lesser trochanter.

Finally, while maintaining the same test setup, destructive quasistatic ramp testing was performed on all femora at a rate of 5 mm/min until catastrophic specimen failure occurred.

Data acquisition and analysis

Machine data on load and displacement were recorded by the load cell, test system transducer, and laser sensor at a rate of 50 Hz. Based on the load- and time-displacement curves, the following parameters of interest were defined and considered for statistical evaluation. AP bending and axial stiffness of both nonfractured and instrumented specimens were calculated from the linear sections of the load-displacement curves during the nondestructive quasistatic AP bending and axial compression tests, respectively. Furthermore, interfragmentary displacement along the diaphyseal axis, as measured by the laser sensor during the cyclic tests, was evaluated at the beginning (cycle 1) and then periodically at intervals of 100 cycles at 100 N (valley) and 1000 N (peak) loads. The initial value at the valley load was considered baseline. Moreover, secondary axial stiffness was evaluated in the instrumented state after the cyclic tests, and failure of the bone-implant

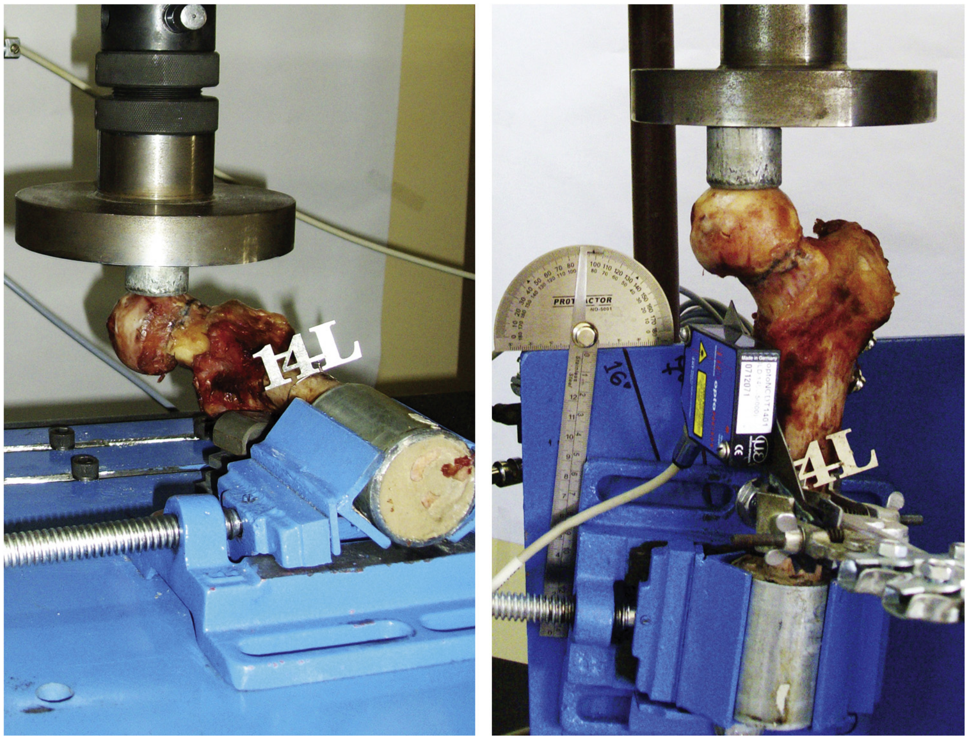

Fig. 4. Setup for biomechanical testing. Specimen 14L instrumented and mounted for anteroposterior bending test (left) and axial compression test at 7° varus inclination (right).

construct (fixation failure) was determined from the load-displacement curves of the destructive quasistatic ramp tests. Whereas secondary axial stiffness was calculated similarly to previous tests, failure load was defined as the absolute maximum load that was followed by a marked decrease in the registered load or fracture. Finally, fixation failure of the bone-implant construct was assessed using radiological data obtained after the destructive tests.

Statistical analyses were performed using SPSS Statistics (Version 19, IBM, Armonk, NY, USA). The normal distribution of each study group was screened with Shapiro -Wilk tests. Paired-samples t tests were used to identify significant differences within each pair of study groups with either FRZ or EMB specimens, as well as significant changes of the parameters between different specimen states in each group. In addition, One-Way ANOVA was used to detect significant differences between the EMB and FRZ specimens. Correlations between BMD and other parameters were screened using Pearson correlation tests. Finally, General Linear Model Repeated Measures was used to identify significant progressive changes in interfragmentary fracture displacement along the diaphyseal axis during the cyclic tests, to screen for significant differences between study groups and determine the influence of BMD as a covariate. The level of significance was set to $p = 0.05$ for all statistical tests.

Results

All parameters of interest were normally distributed in each of the four study groups. BMD values in the femoral neck and greater trochanter regions were not significantly different between the groups, although slightly higher for the embalmed femora (Table 1).

In addition, BMD in the femoral neck was significantly higher than in the greater trochanter and significantly correlated to it ($p < 0.01$).

Initial AP stiffness

The initial AP stiffness of the instrumented specimens did not significantly differ between the groups (Table 1). Compared to the nonfractured specimen state prior to osteotomy, it slightly decreased in each group, as seen in Table 1. The stiffness of the intact and instrumented specimens did not significantly correlate with BMD in the femoral neck or greater trochanter. In addition, intact stiffness demonstrated no significant differences among the four study groups.

Initial axial stiffness

The initial axial stiffness of the instrumented femora was significantly higher following BDSF in comparison with CFIX at 7°

Table 1
BMD (g/cm^2) and AP stiffness (Nm/°) of the specimens in the four study groups (mean ± SEM).

	Fresh-frozen (FRZ)		Embalmed (EMB)	
	CFIX	BDSF	CFIX	BDSF
BMD femoral neck	0.786 ± 0.050	0.776 ± 0.060	0.865 ± 0.041	0.837 ± 0.039
BMD greater trochanter	0.737 ± 0.065	0.743 ± 0.056	0.738 ± 0.053	0.697 ± 0.071
AP stiffness nonfractured	43.5 ± 3.1	42.6 ± 2.5	42.8 ± 2.5	44.8 ± 3.5
AP stiffness instrumented	39.6 ± 2.8	41.0 ± 2.4	39.4 ± 3.4	42.0 ± 3.0

Table 2
Axial stiffness (kN/mm) and failure load (kN) of the specimens in the four study groups (mean ± SEM).

	Fresh-frozen (FRZ) 7° varus inclination		Embalmed (EMB) 16° varus inclination	
	CFIX	BDSF	CFIX	BDSF
Initial axial stiffness nonfractured	1.15 ± 0.11	1.16 ± 0.09	1.28 ± 0.10	1.26 ± 0.09
Initial axial stiffness instrumented	0.53 ± 0.06	0.93 ± 0.10	0.85 ± 0.09	0.82 ± 0.05
Secondary axial stiffness instrumented	1.49 ± 0.06	1.75 ± 0.14	1.43 ± 0.14	1.59 ± 0.33
Failure load	2.68 ± 0.31	3.31 ± 0.36	2.83 ± 0.66	3.02 ± 0.68

varus inclination, $p = 0.02$, and comparable at 16° inclination (Table 2). Whereas for CFIX this stiffness was significantly higher at 16° compared to 7° inclination, $p = 0.01$, for BDSF it remained comparable in both inclinations.

Compared with the nonfractured state at 7° inclination, instrumented initial axial stiffness significantly decreased following only CFIX ($p = 0.01$), but it did not decrease following BDSF (Table 2). Both fixation techniques resulted in a significant drop in the initial axial stiffness at 16° inclination in comparison with nonfractured specimens, $p \leq 0.01$ (Table 2).

In addition, the stiffness of the nonfractured specimens at 7° inclination was lower than 16° inclination. However, this difference was not significant. Statistical significance, in terms of intact stiffness, was not found between the two groups with 7° inclination, nor between the two groups with 16° inclination. Intact axial stiffness at 7° inclination significantly correlated with BMD in the greater trochanter region ($p = 0.03$). No additional significant correlations with BMD were observed in terms of axial stiffness.

Interfragmentary fracture displacement during cyclic testing

Interfragmentary fracture displacement values along the diaphyseal axis during cyclic testing at 100 N (valley) and 1000 N (peak) loads are presented in Tables 3–6. Fracture displacement steadily increased during testing and was smaller in femora fixed with BDSF than with CFIX. The increase at 7° inclination under valley and peak loads was significant between cycles 1, 100, 200, 300, and 400 following both CFIX and BDSF ($p \leq 0.04$; Fig. 5). At 7° inclination, the difference between CFIX and BDSF was significant under both valley and peak loads ($p = 0.04$). In addition, BMD in both femoral neck and greater trochanter significantly affected as covariate the fracture displacement at 7° inclination under valley and peak loads ($p = 0.01$ and 0.02, respectively). Moreover, following each fixation technique separately, interfragmentary fracture displacement at 7° inclination was larger than the displacement observed at 16° inclination during the whole cyclic test. This difference was significant between the two CFIX groups after 100, 200, 300, and 400 cycles ($p \leq 0.04$) but not between the BDSF groups. Regarding interfragmentary fracture displacement in the two groups with 16° varus inclination and BMD's influence as a covariate, no significances were observed.

Secondary axial stiffness

The secondary axial stiffness after cyclic testing, shown in Table 2, was higher following BDSF versus CFIX, with no significant differences between the study groups and no significant correlations to BMD. In addition, the stiffness significantly increased in each group in comparison to initial axial stiffness ($p \leq 0.04$).

Failure load

The failure load was higher for BDSF versus CFIX (Table 2) without statistical significance. In addition, it did not significantly correlate with BMD.

Fixation failure mode

The predominantly observed fixation failure mode for both CFIX and BDSF was longitudinal fracture at the distal femoral neck cortex along the distal screw with varisation (inferior slip) of the proximal fragment and 1–10-mm lateral screw protrusion with or without distal screw bending.

Discussion

In the current study, a novel femoral fracture fixation method was investigated and compared with CFIX, both of which use three partially threaded cannulated screws. Biomechanical tests were performed using different setup and loading protocols to simulate important patient activities. FRZ and EMB human cadaveric femora pairs were used for this purpose. Previous studies concluded that

Table 3
Interfragmentary fracture displacement (mean ± SEM, mm) along the diaphyseal axis at 100-cycle intervals and 100-N loads during cyclic testing of fresh-frozen (FRZ) femora instrumented with CFIX or BDSF and laterally inclined at 7°.

	Displacement at 100-N loads, FRZ at 7° varus Cycle									
	100	200	300	400	500	600	700	800	900	1000
CFIX	0.47 ± 0.16	0.55 ± 0.18	0.59 ± 0.19	0.63 ± 0.20	0.66 ± 0.21	0.69 ± 0.21	0.72 ± 0.22	0.74 ± 0.23	0.75 ± 0.24	0.77 ± 0.24
BDSF	0.14 ± 0.03	0.19 ± 0.04	0.23 ± 0.05	0.25 ± 0.06	0.27 ± 0.07	0.29 ± 0.08	0.30 ± 0.08	0.32 ± 0.09	0.33 ± 0.10	0.34 ± 0.11

Table 4
Interfragmentary fracture displacement (mean ± SEM, mm) along the diaphyseal axis at 100-cycle intervals and 1000-N loads during cyclic testing of FRZ femora instrumented with CFIX or BDSF and laterally inclined at 7°.

	Displacement at 1000-N loads, FRZ at 7° varus Cycle									
	100	200	300	400	500	600	700	800	900	1000
CFIX	0.79 ± 0.18	0.87 ± 0.20	0.91 ± 0.21	0.95 ± 0.23	0.99 ± 0.24	1.01 ± 0.24	1.03 ± 0.25	1.05 ± 0.26	1.07 ± 0.27	1.09 ± 0.28
BDSF	0.38 ± 0.06	0.43 ± 0.07	0.46 ± 0.07	0.48 ± 0.08	0.50 ± 0.09	0.51 ± 0.09	0.53 ± 0.10	0.54 ± 0.11	0.55 ± 0.12	0.56 ± 0.12

Table 5
Interfragmentary fracture displacement (mean ± SEM, mm) along the diaphyseal axis at 100-cycle intervals and 100-N loads during cyclic testing of FRZ femora instrumented with CFIX or BDSF and laterally inclined at 16°.

	Displacement at 100-N loads, EMB at 16° varus Cycle									
	100	200	300	400	500	600	700	800	900	1000
CFIX	0.13 ± 0.03	0.17 ± 0.04	0.21 ± 0.06	0.26 ± 0.11	0.29 ± 0.12	0.31 ± 0.13	0.33 ± 0.14	0.34 ± 0.15	0.35 ± 0.16	0.36 ± 0.16
BDSF	0.11 ± 0.02	0.16 ± 0.03	0.20 ± 0.06	0.23 ± 0.08	0.25 ± 0.10	0.26 ± 0.10	0.28 ± 0.12	0.30 ± 0.13	0.31 ± 0.13	0.32 ± 0.13

Table 6
Interfragmentary fracture displacement (mean ± SEM, mm) along the diaphyseal axis at 100-cycle intervals and 1000-N loads during cyclic testing of FRZ femora instrumented with CFIX or BDSF and laterally inclined at 16°.

	Displacement at 1000-N loads, EMB at 16° varus Cycle									
	100	200	300	400	500	600	700	800	900	1000
CFIX	0.48 ± 0.10	0.52 ± 0.12	0.55 ± 0.13	0.56 ± 0.14	0.57 ± 0.15	0.59 ± 0.15	0.60 ± 0.16	0.62 ± 0.17	0.63 ± 0.18	0.64 ± 0.19
BDSF	0.30 ± 0.06	0.36 ± 0.09	0.40 ± 0.12	0.42 ± 0.13	0.43 ± 0.14	0.45 ± 0.15	0.46 ± 0.15	0.47 ± 0.16	0.48 ± 0.17	0.49 ± 0.18

both these bone types demonstrate similar mechanical characteristics [26–28]. The baseline values for AP bending stiffness, axial stiffness and BMD of all intact femora, obtained in our study, are comparable between the FRZ and EMB specimens, which is consistent with these previous findings [26–28] and justifies the use of both bone types in the current study.

The different levels of axial stiffness between the intact specimens at different inclinations seemed to result from their different levels of tilting.

AP bending and axial compression tests were performed in the current study. The former simulates biomechanical loading situations, such as rising up from a chair, sitting down and climbing stairs with posteriorly directed resultant forces, as described in Davy at al. [32] and Hodge et al. [33]. The axial compression tests at 10° flexion with two different varus inclinations of 7° and 16° resemble the direction of contact forces for most gait activities according to Bergmann et al. [31]. These data contribute to the understanding of the stability achieved with both fixation techniques under different loading situations.

Due to osteotomy in the femoral neck, a general decrease in fixation stability is expected. A slight decrease in AP stiffness after instrumentation was observed in each study group. This was comparable between methods mainly because of similar posterior cortex support.

Compared with the intact state, axial stiffness dropped significantly after both instrumentations at 16° inclination, whereas at 7° inclination the decrease was significant only after CFIX but not after BDSF.

Furthermore, CFIX stability differed significantly between the two inclinations: higher axial stiffness was observed at 16° versus 7°. In contrast, BDSF stability remained similar at both inclinations. Interestingly, axial BDSF stiffness at 7° inclination was even higher than that at 16°. The similar BDSF stability at both inclinations resulted mainly from the specific position and inclination of the distal BDSF screw. When loads are oriented

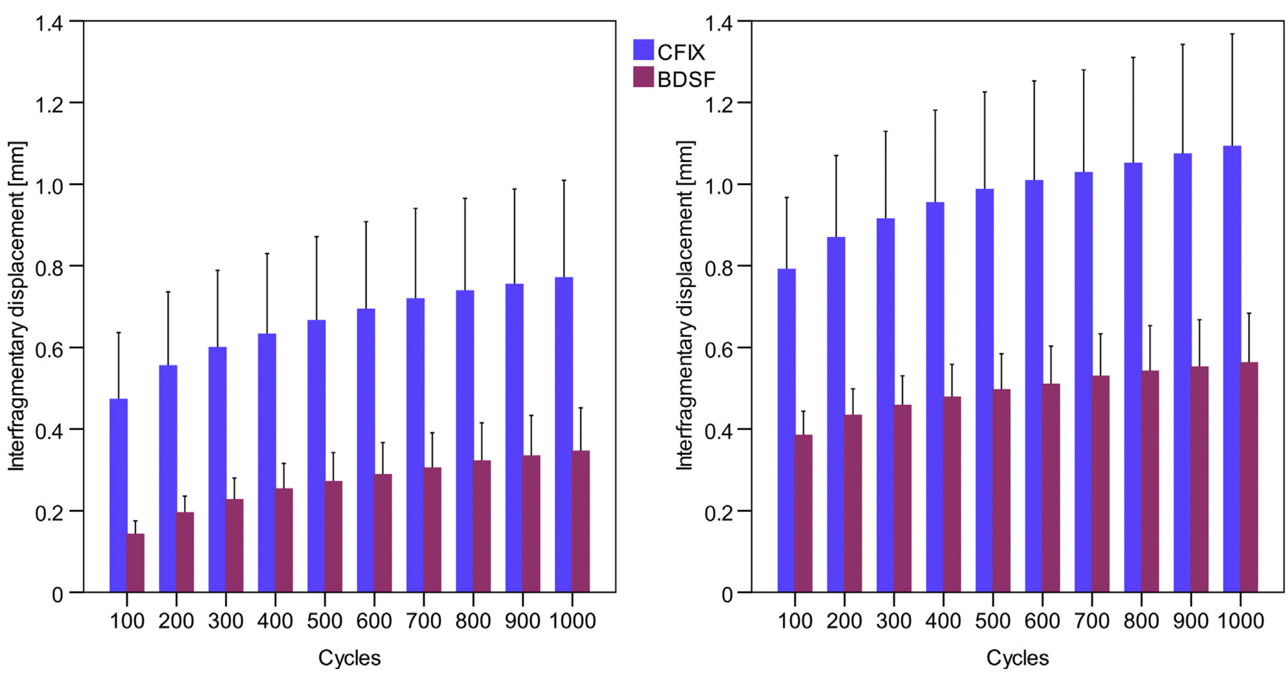

Fig. 5. Interfragmentary fracture displacement along the diaphyseal axis. Mean ± SEM at 100-cycle intervals of 100-N valley (left) and 1000-N peak loads (right). Fresh-frozen (FRZ) femora were instrumented applying CFIX (blue) or BDSF (red) and laterally inclined to 7°. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

more vertically, closer to the diaphyseal axis, construct stability is expected to decrease (as observed following CFIX) due to the increasing transverse component of the load acting on the beam construction, which leads to increasing of the shearing forces. Mechanically, the middle BDSF and distal CFIX screw are fairly equivalent and demonstrate similar entry points, calcar support, lengths, and inclinations. However, in contrast to CFIX, BDSF provides two calcar-buttressed screws that are oriented at different inclinations. If the load is more vertically oriented, the middle screw decreases its bearing capacity, and the obtuse distal BDSF screw comes in optimal orientation for axial weight bearing. Its bearing capacity is added to the middle BDSF screw and helps maintain constant stability across a wide range of inclinations during gait activities, contrary to CFIX. With double support at the inferior and posterior femoral neck cortices, the distal BDSF screw could be especially effective when axially loaded along the diaphyseal axis and when AP bending and torsion are applied. This is an essential advantage of the BDSF method because during diverse patient activities the resultant dynamic forces and moments change their directions, loading the femoral neck in axial compression (e.g., standing on one leg, standing with the feet apart), AP bending and torsion (e.g., rising up from a chair, climbing, running), where the three parallel CFIX screws, all placed at an angle of 120–130° to the diaphyseal axis, can be far less functional.

Secondary axial stiffness after cyclic testing was significantly higher compared with initial axial stiffness regardless of inclination or fixation technique. The reason for this increase could be fracture impaction occurring during the cyclic test.

Higher construct failure loads were noted for BDSF than CFIX. Moreover, the BDSF failure load in the much more unstable situation with 7° varus inclination was similar to that at 16° inclination. This is probably due to the two calcar-buttressed BDSF screws and the specific role of the distal one.

In general, compared with CFIX, the fixation strength of BDSF is considerably higher because of the following factors. (1) Two calcar-buttressed screws are used during BDSF, as opposed to only one screw in CFIX. (2) The two calcar screws are in contact with the distal neck cortex in two different regions, located 1–2 cm apart from each other (depending on the CCD angle), and distribute the applied axial load over a larger surface area. Consequently, in contrast to CFIX, the applied load is spread over approximately 50% of the femoral neck cortex length without concentrating stress in a single spot, thereby resulting in increased bearing capacity. (3) The steeper screw orientation angle to the diaphyseal axis contributes to increased varus resistance, reduced beam sagging, and allows for easier sliding when osteoporotic fracture impaction and shortening occurs during weight bearing, thus avoiding cut-out and maintaining stronger fixation strength. (4) Expected reduced risk of subtrochanteric fracture. The distance between the lateral and medial supporting points of the distal BDSF screw is increased because of its steeper angle to the diaphyseal axis. As a result, the load acting on the lateral and medial cortical-supporting points is reduced [24]. Furthermore, the distance between the distal and medial screw entry points is increased to 20–40 mm, allowing for the tensile forces to spread over a larger area on the lateral cortex. (5) In addition to the posterior cortical neck support, the calcar-supporting point of the distal BDSF screw is located at the lateral part of the inferior neck cortex. Therefore, BDSF can be used for the fixation of more unstable fractures with posterior comminution and/or more vertical fracture lines, whereas CFIX would be inappropriate in these situations. (6) Biologically, BDSF screws are positioned in the ventral and dorsal oblique planes, away from the weight-bearing upper pole of the femoral head, and can thereby avoid the danger of damaging the intraosseous vascularisation.

Biomechanically, no disadvantage of BDSF versus CFIX was found in any sense. This novel fixation method is logical, easy to learn, and repeatable. Anatomical reduction, cortical screw support and intraoperative impaction are the most important steps.

Limitations

The limitations of this study are similar to those inherent to all cadaveric studies. We only included a limited number of specimens, and the use of FRZ and EMB femora limits generalisation to actual patients. In addition, quasistatic destructive testing was performed after cyclic testing.

Conclusion

By providing better cortical support, from a biomechanical point of view the novel BDSF method increases femoral neck fracture fixation strength, improves osteosynthesis outcomes, and extends the indications for internal fixation when osteoporosis is present.

Further clinical studies should be performed to address the question for immediate postoperative full weight bearing without any restriction as recommended in our clinical practice.

Conflict of interest statement

The authors declare that they have no conflict of interest.

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