Does unicondylar knee arthroplasty affect tibial bone strain? A paired cadaveric comparison of fixed- and mobile-bearing designs

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Effect of unicondylar knee arthroplasty on tibial bone strain: a paired cadaveric comparison of fixed- and mobile-bearing designs

Running Title: Bone Strain in UKA

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Each author certifies that his or her institution approved the human protocol for this investigation and that all investigations were conducted in conformity with ethical principles of research.

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Abstract

Background Unexplained pain in the medial proximal tibia frequently leads to revision after unicondylar knee arthroplasty (UKA). As one of the most important factors for osteogenic adaptive response, increased bone strain following UKA has been suggested as a possible cause.

Questions/purposes (1) perform an *in vitro* kinematic analysis on paired cadaveric specimens before and after mobile-bearing and fixed-bearing UKA and (2) simultaneously characterize the strain distribution in the anterior and posterior proximal tibia during squatting.

Methods Five pairs of fresh, frozen full-leg cadaver specimens (4 male, 1 female, 64–87 years) were subjected to a dynamic squatting motion on a kinematic rig to simulate joint loading for a large range of motion. Forces were applied to the quadriceps and hamstrings during the simulation, while an infrared camera system tracked the location of reflective markers attached to the tibia and femur. Tibial cortical bone strain was measured with stacked strain gauge rosettes attached at predefined anterior and posterior positions on the medial cortex. Pairwise implantation of mobile-bearing (UKA_{MB}) and fixed-bearing implants (UKA_{FB}) allowed a direct comparison of right and left knees from the same donor through a linear mixed model.

Results UKA_{MB} more closely replicated native kinematics in terms of tibial rotation as well as in anteroposterior and mediolateral translation. Bone strain values consistently increased compared to native with both designs in the anteromedial and posterior region. However, in the anterolateral region of the medial tibial bone, UKA_{FB} demonstrated the overall biggest

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increase in strain (average peak strain: 1010με±787, $p<0.05$), while UKA_{MB} (613με±395) closely replicated values of the native knee (563με±234).

Conclusion Both UKA_{MB} and UKA_{FB} lead to a significant but comparable increase of anteromedial and posterior tibial strain in comparison with the native knee. In the anterolateral region of the medial tibial plateau UKA, proximal tibial bone strain was significantly closer to native after UKA_{MB} than after UKA_{FB}.

Clinical Relevance Clinical studies will have to show whether the differences in strain increase between both designs translates into a higher rate of pain problems with UKA_{FB}.

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Introduction

Unicondylar knee arthroplasty (UKA) has been advocated as an alternative to total knee arthroplasty (TKA) when osteoarthritis is limited to a single compartment of the knee [9,18,27,49], owing to advantages such as more functional anatomy, and improved post-operative kinematics [9,13,23,27,31,49].

The mobile-bearing UKA (UKA_{MB}) implant, developed as an alternative to the traditional fixed-bearing UKA (UKA_{FB}), allows the polyethylene insert to rotate and translate on the metal tibia tray, thereby contributing to wear reduction [5,6,22,44,47]. However, conflicting findings have been reported on potential differences between these designs in terms of survivorship and mechanical performance [2,5,6,18,44].

Unexplained post-operative pain in medial UKA – particularly in the medial side of the anterior proximal tibia – accounts for 23% of all UKA revisions [36,40]. Data from UK and Australian registries suggests “pain” as the primary reason for revision UKA in 40% and 10% of the cases, respectively. Experimental studies have suggested abnormal distribution of cortical bone strain as a possible cause of post-operative pain [41,42]. While strain gauges have been the gold standard to measure tibial strain [1,41,42], fibre Bragg grating sensors, digital image correlation and finite element methods have also been used to quantify bone strain [1,11,14,35,50].

Few experimental studies have investigated bone strain *in vivo* owing to the invasiveness of the measurement protocol, lack of experimental control and consequent ethical issues [4,29]. Alternatively, *in vitro* studies reporting tibial bone strains are usually conducted under static

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non-physiological loading conditions in isolated bone structures, thereby neglecting soft-tissue forces and the effect of UKA-induced kinematic changes [1,11,41,42,43].

To the best of our knowledge, no *in vitro* study has compared the restoration of strain in the tibial cortex following UKA_{MB} and UKA_{FB} to the native condition, especially for dynamic physiological joint loading. Moreover, the effect of implant design on increased post-operative strain still remains unclear. Therefore, the purpose of this *in vitro* study was (1) to compare kinematics following UKA_{MB} and UKA_{FB} on paired specimens during squatting and (2) to simultaneously characterize the strain distribution in the anterior and posterior regions of the proximal tibia.

The rate of revision surgery following UKA_{FB} is reportedly higher than that for UKA_{MB} [26], which may be linked to abnormally high bone tibial strain as it is the most important factor for osteogenic adaptive response [48]. Hence, it was hypothesized that a UKA_{MB} would result in lower changes in strain compared to native than UKA_{FB} in the anterior region of the tibial cortical bone surface.

Materials and Methods

Paired lower limbs from five fresh-frozen cadavers were disarticulated at the hip (Table 1). All specimens were screened for trauma sequelae, implant material and severe bone deformities before inclusion in the study. Ethical approval was obtained from the regional ethical committee (NH019 2015-11-03). During testing, all specimens were kept moist using a phosphate-buffered saline solution.

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Pre-processing

5mm bicortical bone pins were affixed to the tibia and femur at a distance of 14cm and 17cm from the joint line, respectively, to rigidly mount motion tracking markers. Computed tomography (slice thickness: 0.6mm; Siemens Definition Flash, Siemens, Erlangen, Germany) was performed. To minimize inter-rater variability [45], a single researcher (OT) generated 3D models of the knee and identified anatomical landmarks required to define specimen-specific joint coordinate systems [16] (Mimics 20.0, Materialise, Leuven, Belgium) according to the Grood and Suntay convention [16,19].

Preparation of the Specimens and Sensor Positioning

Specimens were thawed 24 hours before testing. The soft tissue surrounding the knee was removed while carefully preserving the joint capsule, ligaments, and tendons. The femur, tibia and fibula were resected using an oscillating saw 320mm proximally and 280mm distally from the joint line. The femoral head was preserved for later instrumentation with dummy reference strain gauges. The lateral and medial hamstrings were sutured, and the quadriceps tendon was affixed to a custom-made clamp. Each bone was embedded in metal containers with a cold-cure acrylic resin (VersoCit2, Struers, Ballerup, Denmark). Care was taken to maintain the physiological tibiofemoral alignment.

To allow for measurement of tibial cortical bone strain, three stacked strain gauge rosettes (diameter: 10mm, grid length: 3mm; KFG-3-120-D17-11L1M2S, Kyowa, Tokyo, Japan)

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[8,38,42] were attached using a previously described method [15] at predefined anterior and posterior positions on the tibial cortex, 34.4mm ± 5.28mm distally from the tibial plateau (Fig. 1A). An anteromedial sensor was placed 18.18mm ± 5.4mm medially from the tibial mechanical axis. An anterolateral sensor was placed 10.55mm ± 6.9mm medially from the tibial mechanical axis. A posterior sensor was placed 6.77mm ± 4.3mm medially from the tibial mechanical axis. Additionally, three 'dummy' sensors were placed on the separate piece of femoral head, and connected to the measurement strain gauges using a Wheatstone bridge configuration, to prevent environmental factors, such as temperature change, from confounding the measurements. Sensor outputs were recorded at 2000 Hz and synchronized with the other devices using Labview (National Instruments, Austin, TX, USA).

The position of the strain gauges on the tibia were digitized by two operators (OT, JS) using a digitizing wand tracked by a motion capture system (Vicon, Oxford, UK), and normalized as per the length and proximal mediolateral width of the tibia. There were no differences between specimen sides (Wilcoxon signed rank test, $p > 0.05$), except for the posterior sensor's normalized mediolateral position ($p=0.043$); the sensor in the left legs were on average positioned 12(±7)% closer to the tibial mechanical axis.

Test Set-up and Protocols

Specimens were mounted in a previously validated cadaveric knee joint simulator [46] that applied a dynamic squatting motion to simulate load through a full range of motion. Clusters with four spherical retro-reflective markers each were mounted on the bone pin-mounted

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holders (Fig. 1B). An electromechanical actuator was used to apply dynamic load to the quadriceps, while the hamstrings were loaded at 50 N using constant-force springs [46]. Six cycles of pre-loading for squatting were applied, with a resting time of 1 minute between cycles to precondition the specimen and minimize hysteresis [17,39]. During squatting, the quadriceps load was programmed to maintain a constant vertical ankle load of 110 N [19,21,33].

A six-camera motion capture system (MX40+, Vicon, Oxford, UK) was used to track marker clusters on the bones. Tibiofemoral kinematics during squatting were analysed using dedicated motion capture software (Nexus 1.8.5, Vicon, Oxford, UK) and custom-programmes in Matlab (R2017b, Mathworks Inc., Natick, MA, USA) [16,21].

Implantation

Following native testing, specimens were implanted with a medial UKA by a single surgeon (GP) following the manufacturer guidelines. UKA_{MB} (right leg: Oxford, Microplasty; Zimmer Biomet, Warsaw, IN, USA) was performed on all right knees and UKA_{FB} (left leg: Vanguard M; Zimmer Biomet, Warsaw, IN, USA) on the left using a minimally-invasive medial parapatellar approach with controlled under-correction of the overall mechanical alignment [3,7,24]. All motion trials were repeated for each specimen as above. Furthermore, each tibial tray position was digitized by a wand to compare varus and posterior tilt angles [30,37] between both specimen sides, i.e. design groups, (Wilcoxon signed rank test). No differences were found ($p>0.05$).

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Data Processing

Data obtained from the motion capture system were downsampled and interpolated at intervals of 1° of flexion and within a common range of knee flexion (40°-99°) for all specimens. Kinematics were reported as mean ± standard deviation for each condition (native, UKA_{FB}, and UKA_{MB}). Maximum and minimum principal strains were recorded by the rosette sensors as a function of knee flexion [28]

$$\varepsilon_{max} = \frac{1}{2}(\varepsilon_1 + \varepsilon_2) + \frac{1}{\sqrt{2}}\sqrt{((\varepsilon_1 - \varepsilon_3)^2 + (\varepsilon_3 - \varepsilon_2)^2)}$$

$$\varepsilon_{min} = \frac{1}{2}(\varepsilon_1 + \varepsilon_2) - \frac{1}{\sqrt{2}}\sqrt{((\varepsilon_1 - \varepsilon_3)^2 + (\varepsilon_3 - \varepsilon_2)^2)}$$

$\varepsilon_{1,2,3}$ express the normal strains from each of the rosette strain gages, ε_{max} and ε_{min} express the maximum and minimum principal strains, respectively (Fig. 1C).

Statistical Analysis

All kinematic and strain data were expressed as differences between post-operative and native condition, i.e. $Post_{(UKA_{MB})} - Pre_{Right\ (Native)}$ and $Post_{(UKA_{FB})} - Pre_{Left\ (Native)}$. This allowed us to consider repeated measurements comparing pre-UKA and post-UKA conditions, and left and right legs, without sacrificing statistical power.

A linear mixed model was used to test for significant differences between implant designs ($p<0.05$), using the “nlme” package (R-Studio 1.0.136, Boston, MA, USA), with implant side

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as a function of knee flexion angle (fixed effect) and the donor as the second repeated measurement (random effect) (dataset ~ implant type*flexion angle, random=~1|Donor). Furthermore, for both designs the difference in peak strain with respect to native over the flexion range were analysed in terms of effect size (Cohen's d).

Results

Kinematics During Squatting

The UKA_{MB} and UKA_{FB} designs were not different in terms of valgus orientation throughout the range of flexion (Table 2). With respect to the native condition, UKA_{MB} and UKA_{FB} both demonstrated a shift towards increased valgus (Table 2, Fig. 2A).

In contrast, UKA_{FB} demonstrated more external rotation than UKA_{MB}; the latter more-closely replicating the native condition between 79 and 99° of flexion (Fig. 2B).

In terms of translational kinematics, the UKA_{MB} design more closely replicated the native condition for the inferior-superior position of the medial and lateral femoral condyle centers (Table 2); however, UKA_{FB} consistently demonstrated a more superior position throughout the flexion range (Fig. 2C-D). This introduced a difference in the inferior-superior position between the UKA designs for the medial (86°-92° and 94°-99°) and lateral femoral condyle (82°-99°). Finally, UKA_{MB} also more closely replicated native behavior in terms of anterior-posterior translation of both condyles between 65° and 94° and between 55° and 99° of flexion for the medial and lateral femoral condyle centers, respectively (Fig. 2E-F, Table 2A).

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Native knees demonstrated rollback patterns during squatting in which the medial and lateral femoral condyle centers translated anteriorly and posteriorly on the tibial plateau, respectively (Fig. 3). Associated with the above findings in terms of anterior-posterior translations; the medial and lateral femoral condyle centers qualitatively demonstrated similar patterns during squatting in the UKA_{MB} condition, whereas the UKA_{FB} condition demonstrated posterior translation for the medial and lateral femoral condyle centers as the flexion angle increased.

Tibial Cortical Bone Strain

Bone strain values consistently increased compared to native with both designs in the anteromedial and posterior region (Fig.4), with the increase in UKA_{MB} being larger from 96° and 94° onwards, respectfully (Table 2B), and an increased peak strain of 92% (effect size 1.5) and 97% (effect size 2.1 - Table 3). However, in the anterolateral region of the medial tibial bone, UKA_{FB} demonstrated consistently increased strain, while UKA_{MB} closely replicated strain values of the native knee in this particular area (Fig.4). This region also showed the overall highest maximal principal strain values following UKA_{FB} (1010 $\mu\varepsilon$ (±787) or 79% increase, effect size 1.9 - Table 3) compared with all regions in both the native and UKA_{MB} conditions.

In terms of the minimum principal strain, the anteromedial region showed no difference with numbers available between designs throughout the range of knee flexion (Table 2), associated with small effect sizes (Table 3). In contrast, UKA_{MB} and UKA_{FB} showed differences in the minimal anterolateral tibial strain between 97° and 99° of flexion (Table 2B), with the UKA_{FB} design demonstrating consistently increased, as well as the overall highest, strain

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(effect size 0.8 - Table 3), similar to the maximal principal strain. The UKA_{MB} design's values closely replicated those of the native strain (Fig. 4), resulting in an effect size of 0 (Table 3).

Discussion

It has been suggested that UKA_{MB} replicates native tibiofemoral kinematics better than UKA_{FB} does [10,19-21,33]. Although previous studies have compared the kinematics of the native knee with those of UKA_{MB} or UKA_{FB} designs individually, this study is, to our knowledge, the first cadaver-based study directly comparing UKA_{MB} and UKA_{FB} using matched pairs and integrated assessment of tibial bone strain.

Kinematics

In terms of kinematics, this study confirms the results of previous studies evaluating these UKA designs individually; there were design-specific changes in terms of kinematics [20,21]. We found that UKA_{MB} replicates native tibiofemoral kinematics better than UKA_{FB} does. More specifically, both UKA_{MB} and UKA_{FB} demonstrated increased valgus as compared to the native condition (Fig. 2A) which has been associated with stiffness mismatch induced by these implants at the medial side [21,24]. However, UKA_{MB} allowed for better preservation of tibial rotation and better AP stability of the medial femoral condyle through the flexion cycle than UKA_{FB} did. UKA_{MB} also more closely approximated the native inferior-superior translation than UKA_{FB} did, although with both designs, the medial femoral condyle center had a more superior position throughout the range of flexion. Additionally, although tibiae

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moved into more external rotation with the UKA_{FB}, UKA_{MB} preserved internal rotation of the tibia and the associated screw home mechanism towards full extension. It has been suggested that these differences are because of increased conformity of the UKA_{MB} design, which mimics the concavity of the anatomy of a native knee [19,33]. In most UKA_{FB} designs, including the one tested here, the concave anatomy is typically replaced with a flat polyethylene articulating surface, resulting in less AP constraint. Similarly, there were differences between the UKA_{FB} and UKA_{MB} designs; with UKA_{FB}, the lateral femoral condyle paradoxically slid anteriorly, while it remained relatively stable with a UKA_{MB}.

Tibial Cortical Bone Strain

Both implants demonstrated increased bone strain in the posterior and anteromedial tibial bone in deep flexion in comparison with the native knee. The biggest difference between both designs was noted in the anterolateral part of the medial compartment where UKA_{MB} demonstrated bone strain close to native, in contrast with a 79% increase in peak strain in UKA_{FB}. The results confirmed our initial hypothesis that UKA_{MB} generates lower strain than UKA_{FB}, for the anterolateral region only. Although there were differences in deep flexion between UKA_{MB} and UKA_{FB} in terms of the peak strain at the far medial side of the anterior tibia as well as in the posterior region, the UKAs demonstrated similar patterns and had only small differences in strain magnitude. Based on the outcome of anterior strains and previous studies [12,32], our results thus suggest that cortical strain values for UKA_{MB} in the anterolateral region might be associated with a lower risk of post-UKA symptoms such as pain, implant loosening, or fracture.

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In an attempt to link kinematics and strain behavior of these designs there seemed to be no obvious relation with the numbers available. One might expect that the consistently more posterior position of the medial femoral condyle center on tibia in UKA_{FB} would lead to loads being transferred through the posterior aspect of the tibia and consequently lead to lower compressive strains on the medial tibia's anterior side. This is the opposite of our most pronounced finding in terms of strain; that is, increased strain on the anterolateral aspect of the medial tibia in UKA_{FB}. Instead, the observed strain differences might be associated with the orientation and magnitude of the quadriceps muscle force that is transferred to the tibia through the patellar tendon [25]. While the knee goes into flexion during squatting, the force the patellar tendon exerts on the tibia increases, along with a decreasing angle between the patellar tendon and the tibial axis in the sagittal plane (Fig. 5) [25]. Because of this decreased angle, the vertical component of the force vector increases, which further increases compressive strains on the tibia. Pegg et al. [34] reported that after UKA, strains around the patellar tendon's insertion were increased, and these authors highlighted the impact of muscle forces on tibial strain. Regarding our kinematic findings, the more posterior position of the medial femoral condyle center in UKA_{FB} might have been associated with a further increased vertical component of the patellar tendon's force vector, perhaps explaining the increased strain at the anterolateral side of the medial tibia. However, because we unfortunately did not measure the position of the patella in our study, no other supporting data are available.

Limitations

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This study had several limitations. First, there are inherent limitations to cadaver studies, including limited and high-cost access to specimens, leading to typical sample sizes of 5 to 10 [21]. Despite careful selection (see Table 1) and preservation, cadaver bone may have behaved different from the targeted clinical population. Nevertheless, for the native condition, a peak maximum anteromedial principal strain of 311 με ± 190 με and a peak maximum anterolateral strain of 563 με ±234 με anterolateraly were found during squatting which compare well to the results of prior *in vivo* studies: Lanyon et al. [29] reported principal strains of 850 με during running and 400 με during walking. Nevertheless, our findings only apply to one motor task and others may be associated with different strains. As such, Burr et al. [4] focused on more-demanding motor tasks and noticed a dramatic increase especially during zigzag running with maximal compressive strains of 1226 με ±168 με and tensile strains of 743 με ±77 με were recorded; indeed higher than the native strain ranges in our study. Measurements were still subject to the exact location of strain sensors. Given the absence of systematic differences between both design groups we do not expect them to have impacted our design-related findings. Nevertheless, we plan to further investigate the full field proximal tibial strain, as well as the effect of other possible contributing factors such as ligament tensioning, motor tasks and implant distortion, through a detailed finite element analysis based on the data of this study. Additionally, for safety and inter-specimen variations in tibiofemoral alignment, squatting motions were limited to 40 - 99° of flexion, with a relatively low vertical load of 110N. Scott et al. [41] performed an *in vitro* digital image correlation-based strain analysis of medium composite tibial sawbones to investigate different UKA designs under a higher, but static vertical load of 2500 N directly applied to the medial tibial compartment. They reported that the fixed-bearing design generated lower maximal

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vertical strain in the medial aspect of the tibia (1301 με ± 328 με) than the UKA_{MB} did (1662 με ± 32 με) [41], which complies with our findings (anteromedial sensor) (Fig. 4A). The findings made here may not be applicable to other UKA of other manufacturers and designs. Last, many of the observed differences remain small and it remains unclear to which extent they clinically matter.

Still, to the best of our knowledge, this is the first study comparing cortical strain behavior under dynamic, loaded conditions between UKA_{MB} and UKA_{FB}. Overall, the strength of the study is the simultaneous comparison of kinematics and bone strain between UKA_{MB} and UKA_{FB} in both knees of the same donors in an established setup by a single surgeon using two closely related UKA designs.

Clinical Relevance

*C*linical studies will have to show whether the observed small differences in strain increase between both designs translate into a higher rate of pain problems with UKA_{FB}.

Conclusions

In this *in vitro* cadaver study both, UKA_{MB} and UKA_{FB} lead to a significant increase of bone strain in comparison with the native knee. In the anterolateral region of the medial tibial plateau, proximal tibial bone strain was lower after UKA_{MB} than after UKA_{FB}.

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Legends

Fig. 1 Experimental setup: (A) preparation of the tibial surface and attachment of anteromedial (AM) and anterolateral (AL) strain gauges (B) dynamic knee simulator replicating loaded squatting (C) representation of strain calculation ($\varepsilon_{1,2,3}$ = normal strains from each rosette strain gauge; ε_{max} and ε_{min} = principal strains).

Fig. 2 Kinematics of the knee in the native condition (black) and following unicondylar knee arthroplasty using fixed-bearing (FB) implants (green) and mobile-bearing (MB) implants throughout the range of flexion for (A) valgus orientation (B) tibial internal rotation (C) inferosuperior position of the medial femoral condyle center (FMCC IS) and (D) lateral femoral condyle center (FLCC IS) (E) anteroposterior position of the medial femoral condyle center (FMCC AP) and (F) lateral femoral condyle center (FLCC AP). Data is represented as mean (solid) ± standard deviation (shaded).

Fig. 3 Mean femoral rollback pattern across the specimens in (A) the native condition and following unicondylar knee arthroplasty using (B) fixed-bearing (FB) implants and (C) mobile-bearing (MB) implants throughout the range of flexion during squatting. Solid dots on the tibial plateau represent the centers of the medial and lateral femoral condyles.

Fig. 4 Maximum and minimum principal strains in the native condition (black) and following unicondylar knee arthroplasty using fixed-bearing (FB) implants (green) and mobile-bearing (MB) implants throughout the range of flexion for (A) anteromedial (AM) (B) anterolateral (AL) and (C) posterior sensors (P) attached on the proximal tibial cortex. Data is represented as mean (solid) ± standard deviation (shaded).

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Fig. 5 Schematic of the patellar tendon (PT) force, quadriceps tendon (QT) force and medial femoral condyle center (FMCC) in (A) full extension and (B) flexion at 90° following unicondylar knee arthroplasty using fixed-bearing (FB) implants (green) and mobile-bearing (MB) implants (red).

Queries on figure legends

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