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1 **Should we recommend occipital plate fixation using bicortical screws**
2 **or inverted occipital hooks to optimize occipito-cervical junction fusion?**
3 **A biomechanical study combining an experimental and analytical approach.**

4
5 A. Germaneau^{1,2} (PhD), T. Vendevre^{1,2} (MD, PhD), A. Delmotte² (MD), S. D'Houtaud² (MD),
6 C. Brèque^{1,3} (PhD), L. Petureau¹ (PhD), P. Doumalin¹ (PhD), J.-C. Dupré¹ (PhD), F. Brémand¹ (PhD),
7 P. Maxy⁴, J.P. Richer³ (MD, PhD), P. Rigoard^{1,2} (MD, PhD)

8
9 ¹ Institut Pprime UPR 3346, CNRS – Université de Poitiers – ISAE-ENSMA, France.

10 ² Spine & neuromodulation functional unit, Department of neurosurgery, CHU Poitiers, PRISMATICS Lab, Poitiers, France

11 ³ ABS Lab, Université de Poitiers, France.

12 ⁴ Medtronic, Medtronic International Trading Sarl, Tolochenaz, Switzerland

13
14 Corresponding author: A. Germaneau, arnaud.germaneau@univ-poitiers.fr

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19

1

2 **Abstract**

3 *Background:* Occipito-cervical fusion can be necessary in case of cranio-cervical junction instability. Proximal
4 stabilisation is usually ensured by bi-cortical occipital screws implanted through one median or two lateral
5 occipital plate(s). Bone thickness variability as well as the proximity of vasculo-nervous elements can induce
6 substantial morbidity. The choice of site and implant type remains difficult for surgeons and is often empirically
7 based. Given this challenge, implants with smaller pitch to increase bone interfacing are being developed, as is a
8 surgical technique consisting in inverted occipital hook clamps, a potential alternative to plate/screws
9 association. We present here a biomechanical comparison of the different occipito-cervical fusion devices.

10 *Methods:* We have developed a 3D mark tracking technique to measure experimental mechanical data on
11 implants and occipital bone. Biomechanical tests were performed to study the mechanical stiffness of the
12 occipito-cervical instrumentation on human skulls. Four occipital implant systems were analysed: lateral
13 plates+large pitch screws, lateral plates+hooks, lateral plates+small pitch screws and median plate+small pitch
14 screws. Mechanical responses were analysed using 3D displacement field measurements from optical methods
15 and compared with an analytical model.

16 *Findings:* Paradoxical mechanical responses were observed among the four types of fixations. Lateral
17 plates+small pitch screws appear to show the best accordance of displacement field between
18 bone/implant/system interface providing higher stiffness and an average maximum moment around 50N.m
19 before fracture.

20 *Interpretation:* Stability of occipito-cervical fixation depends not only on the site of screws implantation and
21 occipital bone thickness but is also directly influenced by the type of occipital implant.

22

23 **Keywords:** occipital bone, occipito-cervical fusion, mark tracking, 3D measurements, biomechanics

24

1 **1. Introduction**

2 Surgical occipito-cervical fusion is required when cranio-cervical junction stability has been compromised
3 (Winegar et al., 2010) characterized by excessive laxity or mobility. This condition can result from various
4 degenerative (Goel et al., 2010), tumoral (George et al., 2006; Zimmermann et al., 2002), traumatic (Vaccaro et
5 al., 2005), inflammatory (Grob et al., 1999), infectious (Shukla et al., 2005) or congenital diseases (Giussani et
6 al., 2009).

7 Stabilization can be achieved by performing a posterior instrumentation between the horizontal part of the
8 occipital bone and the cervical spine to bridge the instability area (Vaccaro et al., 2005). The choice of multi-
9 level articular screws at the cervical level has rapidly reached a consensus, as a compelling technique has been
10 developed to implant screws safely at the facet level, thereby ensuring distal stabilisation anchorage (Sasso et al.,
11 1994).

12 By contrast, at the occipital level, the choice of site and implant type since first indications has been
13 controversial. The anatomical particularities of this region can partially explain why so many types of implants
14 have been developed by industry to provide adequate anchorage at the proximal occipital level. The median
15 external occipital protuberance (EOP) is the thickest part of the occiput. From this anatomical landmark and the
16 underlying median external crest joining EOP to the foramen magnum, occipital bone thickness decreases
17 laterally and inferiorly in a radial fashion (Papagelopoulos et al., 2000). EOP underlying projection is frequently
18 associated with the convergence of major venous sinuses (Roberts et al., 1998; Wolfla, 2006; Zipnick et al.,
19 1996). A recent study reported that a contact between venous structures and the internal occipital crest has been
20 found for 37% of the study population (Kobayashi et al., 2006).

21 From a mechanical point of view, proximal occipital stability of the fixation is usually ensured by bi-cortical
22 occipital screws implanted through one median or two lateral occipital plate(s) (Heywood et al., 1988; Roberts et
23 al., 1998; Wolfla, 2006). The variability of occipital bone local thickness combined with the anatomy of dural
24 sinuses, should considerably influence the perimeter of a safe working area and the length of occipital screws.
25 This can potentially affect fixation strength and stiffness and entailing but also operative risks (massive
26 haemorrhage, air embolism, Cerebro-Spinal Fluid - CSF) leak and direct cerebellar tissue injury (Ebraheim et al.,
27 1996; Heywood et al., 1988; Korovessis et al., 2001; Wolfla, 2006). For that, it seems important to be able to
28 study the mechanical behavior at the interface between bone and implant by identifying stiffness of fixations
29 (ratio between loading moment to impose movement and range of angular variation).

30 Given this challenge, surgical implant design has evolved in three new directions:

1 (i) A surgical technique consisting in implanting an inverted occipital hook clamp on each side, inserted through
2 a burr hole was developed in 1998 (Faure et al., 2002, 1998) and recently used for occipito-cervical arthrodesis
3 for young and small children (Odent et al., 2015), as a potential alternative to optimize occipito-cervical fixation.
4 Using this technique, dura mater, venous and parenchyma structures seem to be better controlled during surgical
5 intervention. Concerning the mechanical behavior, the hooks would propose a significant contact surface to
6 transmit forces involved by neck movements to the occipital bone.

7 (ii) In another direction, industry has provided more screws with smaller pitch to increase bone interfacing and to
8 decrease contact pressure between bone and implant. Historically, screws have progressively replaced hooks
9 because of their higher resistance in pull-out (Hafer et al., 1999) and lower iatrogenic positioning. The
10 association of using screws and lateral plates could provide a better stiffness (Anderson et al., 2006) by
11 decreasing load arm because of screw alignment with cervical implants. The risk of lesioning a venous sinus
12 appears lower but dural lesion and cerebellar wounds have been described and associated lower bone thickness
13 at this level (Kobayashi et al., 2006; Zipnick et al., 1996).

14 Conceptually, the use of bi-cortical screws should offer higher mechanical strength but according to some
15 authors (Hafer et al., 1999; Papagelopoulos et al., 2000), simple cortical screws have produced good results in
16 terms of mechanical response and mitigated risks of lesioning adjacent structures (venous sinus, dura mater,
17 cerebellum).

18 (iii) Some authors tend to favour a median positioning of one single occipital plate (Antar and Turk, 2018)
19 instead of two laterally. Indeed a correlation exists between bone thickness, screw length and pull-out strength
20 (Hafer et al., 1999; Hurlbert et al., 1999; Wolfla, 2006). A proper anchorage in this region would require a
21 significant length of screws to provide optimal stability. However, this positioning could lead to a higher risk of
22 venous bleeding related to median occipital venous sinus (Kobayashi et al., 2006; Zipnick et al., 1996). A frontal
23 shift between median occipital implants and cervical implants could decrease stiffness by increasing load arm, in
24 particular for rotation movements (Anderson et al., 2006). Therefore, some clinicians do recommend this medial
25 occipital zone and the insertion of screws as long as possible to maximize stability and ensure a durable stable
26 mechanical response. This strategy is however linked to a higher risk of median occipital venous sinus damage
27 for these anatomical reasons.

28 Overall, the choice of occipital implant and implant site can therefore be difficult for the surgeon. To determine
29 the best compromise in terms of clinical benefits/surgical risks balance and surgical stability, the goal of this
30 study was to provide a biomechanical analysis and a comparison of the four most used devices dedicated to

1 occipito-cervical junction fusion. A two-step combined methodology was envisaged: An initial analytical
2 mechanical model was developed as a pre-requisite to define contact pressure, corresponding to mechanical
3 stress transmitted to the bone, involved by implants at the screw site or plate interface. This theoretical approach,
4 presented in this article, describes the determination of contact pressure for various implants in order to analyse
5 the mechanical effects according to the interface and the bone thickness and to predict the risk of bone damage
6 or fracture for imposed loadings. As a second step, we designed an experimental study to analyse the
7 biomechanical behaviour of occipital implants on human cadaveric specimens. We developed a specific
8 experimental loading setup coupled with a non-contact analysis by using an optical method to measure
9 displacement fields and to analyse mechanical behaviour. The experimental approach describes a complete
10 mechanical analysis of occipital fixations by evaluating their stiffness from the relationships between mobility
11 (displacement / angle variations) and loadings (imposed moment for flexion-extension movements) and by
12 identifying their mechanical strength from pull-out tests.

13

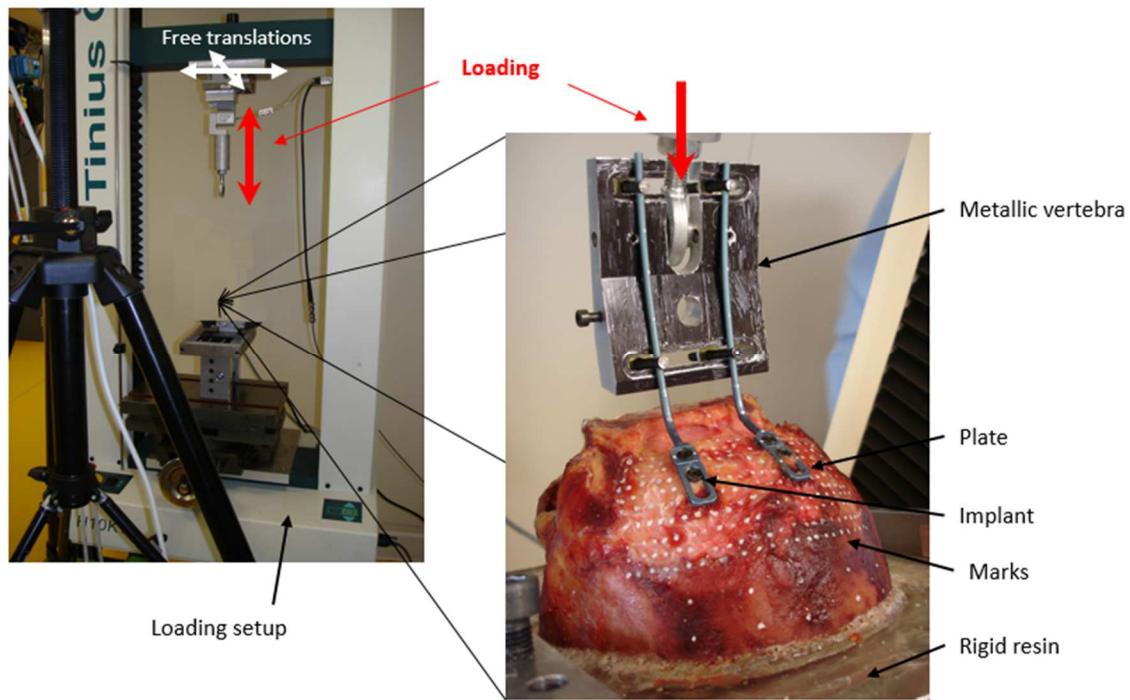
14 **2. Methods**

15 **2.1. Specimen preparation**

16 Twenty-four anatomical fresh frozen skulls were prepared (Laboratory of anatomy, Université de Poitiers -
17 n°DC-2019-3704). After thawing, skulls were extracted and cleared of soft tissue. In each one, the calvaria was
18 sawed and the brain and soft tissues were removed. Anterior skull structures including the mandible were
19 removed too. Then cervical vertebrae were substituted by a metallic plate (Figure 1) on which load was applied.
20 For each skull, superior sawed part was potted in rigid polyurethane resin. Fixations were achieved using 3mm-
21 diameter rod. The straight side of the rod was fixed to cervical spine and the other extremity was fixed to the
22 occipital bone with implants and a plate. Different implants were studied: screws directly implanted in the bone
23 or hooks (Figure 2). The specimens were distributed among four groups, each one corresponding to one type of
24 implant (Figure 2): lateral plates + large pitch screw (LP+LPS), lateral plates + hooks (LP+H), lateral plates +
25 small pitch screws (LP+SPS) and median plate + small pitch screws (MP+SPS). Diameter of screws implanted
26 in the skull was 4mm and the length was adapted to the local thickness of the bone. To achieve the fixation using
27 hooks, a burr hole of 10mm diameter was drilled on both sides of the external occipital crest. The specimens
28 were distributed among the four groups with homogeneity in sex and age (Table 1). Bone thickness was also
29 measured with a specific gauge just after hole drilling. We can observe a higher value of standard deviation of

1 bone thickness in the group MP+SPS due to implantation of 2 screws in the occipital crest (high bone thickness)
 2 and 2 screws on both sides.

3

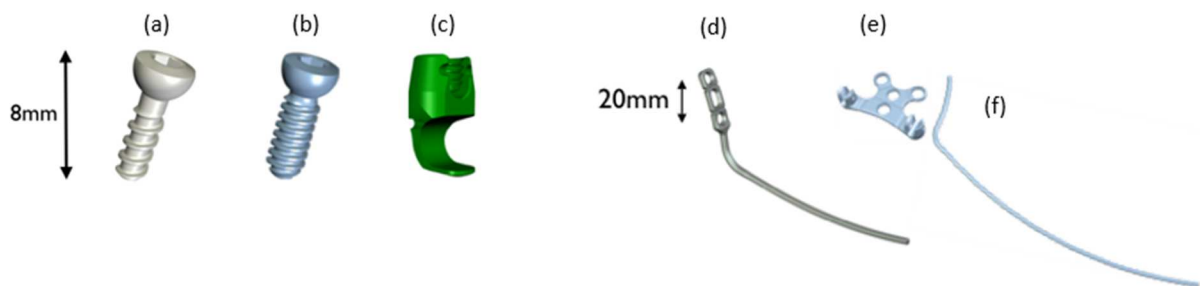


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5 Figure 1: Experimental setup and specimen preparation

6

7



8

9 Figure 2: Occipital implants: (a) Large Pitch Screw (LPS); (b) Small Pitch Screw (SPS); (c) Hook (H); (d)

10 Lateral Plate (LP); (e) Median Plate; (f) Rod

11

Fixation	Mean age (years)	Sex-ratio	Bone thickness (mm) Mean (SD)
LP+LPS	71.66	3H/3F	5.3 (0.9)
LP+H	71.33	3H/3F	4.0 (0.9)
LP+SPS	74.33	4H/2F	6.9 (1.4)

MP+SPS	73	4H/2F	9.9 (2.5)
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1 Table 1: Specimen distribution according to sex and age

2

3 **2.2. Analytical model**



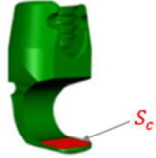
4 An analytical approach was performed to establish theoretical mechanical response of implants and to determine
5 contact pressure between bone and implants by considering similar loading and boundary conditions than the
6 ones imposed by the experimental setup presented in Figure 1. Determination of contact pressure provides an
7 evaluation of mechanical stress transmitted between implants and bone during a movement of the occipito-
8 cervical junction. Mechanical loadings (force F and moment M) and pressures (p) are supposed to be
9 homogeneous distributed in the contact areas of fixations. Relationships allowing the estimation of contact
10 pressure are presented in the Table 2. Fixations by screws are essentially ensured by the contact between
11 implants and cortical bone (highlighted in red in the figure presented in the Table 2). In the lateral area, occipital
12 bone is principally constituted by cortical bone and mechanical loadings are supposed to be homogeneously
13 distributed along the thread on the whole thickness of the bone. Concerning median fixation, there is a
14 significant part of cancellous bone and pressure involved by loading applied on the fixation cannot be supposed
15 homogeneously distributed along all the length of the thread. For this latter case and according to the observation
16 of cadaveric specimens, a coefficient from the ratio between thicknesses of cortical bone and cancellous bone
17 was considered to estimate contact pressure involved by load transmission in screws.

18 Concerning fixations by hooks, two situations were considered to determine contact pressure:

19 - If the hook size exactly corresponds to the thickness bone (Table 2 (b1)): the pressure is supposed to be
20 uniformly distributed on the contact surface between hook and bone.

21 - If there the hook size does not exactly correspond to bone thickness (Table 2 (b2)): a slightly larger hook is
22 chosen and contact area between bone and hook is localized and corresponds to a line. In this case, Hertz theory
23 (Johnson, 1987) can be used to determine contact pressure, we assumed an elastic behaviour for materials,
24 mechanical characteristics are given in Table 2.

25

Implant	(a) Screws (LPS and SPS)		(b) Hook (H)	
		(a1) LPS Diameter = 4.5mm Thread depth = 0.75mm Pitch = 1.8mm 	(a2) SPS Diameter = 4.5mm Thread depth = 0.5mm Pitch = 1mm 	(b1) Hook size = bone thickness Width = 2mm Depth = 3mm 
Contact pressure	$p_s = \frac{F}{S_c} + \frac{M \cdot d_i}{I_c}$		$p_h = \frac{F}{S_c}$	$p_h = \left(\frac{F}{\pi \cdot r_i \cdot c \cdot \left(\frac{1 - \nu_i^2}{E_i} + \frac{1 - \nu_b^2}{E_b} \right)} \right)^{1/2}$
Notations	F and M : loading components (force and moment) S_c and I_c : contact surface and second moment of area d_i : dimension of the screw section		c : contact length (hook width) r_i : local radius of the hook at the contact with the bone	
Mechanical characteristics			E_i : Elastic modulus of Titanium implant, $E_i = 110GPa$ ν_i : Poisson ratio of Titanium implant, $\nu_i = 0.3$ E_b : Elastic modulus of cortical bone, $E_b = 12GPa$ (Boruah et al., 2017) ν_b : Poisson ratio of cortical bone, $\nu_b = 0.3$	

1

2 Table 2: Evaluation of contact pressure according to the type of implant and contact areas

3

4 **2.3. Experiments**

5 Load was applied directly on the metallic vertebra with a classical tensile test setup (Figure 1) to study
6 mechanical effects linked to flexion-extension movements. Moments (\vec{M}) were imposed by a force (\vec{F}) with a
7 level arm between the load axis to the point of force application (distance between noted \vec{d}). Free linear rail
8 guides were employed to maintain direction of the force during the loading (Figure 1) and to make free the
9 displacements along the other directions. The imposed moment can be deduced from the force controlled by the
10 loading setup: $\vec{M} = \vec{d} \wedge \vec{F}$ (values of forces and moment are given by norms of the vectors, noted F and M).

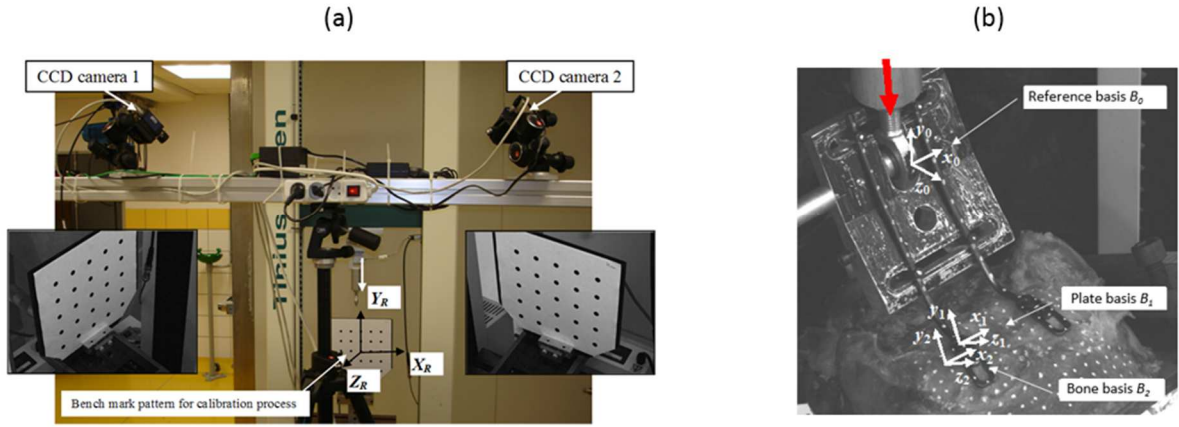
11 A 3D mark tracking method (Germaneau et al., 2016) was employed to measure displacement components
12 during the mechanical loading on the full surface of the posterior, on the fixations and on the rods and also on
13 the metallic vertebra to compute the imposed moment value (M). This method enables to measure 3D
14 displacements of marks (natural or artificial) deposited on surfaces from a stereoscopic vision and the use of two
15 CCD cameras (Figure 3-a) to obtain evolution of mark positions in the space.

16 To determine the range of motion of the bone according to the metallic vertebra and the applied loading, three
17 orthogonal local bases were defined (Figure 3-b): a reference one on the metallic vertebra (noted B0), on second
18 one on the plate fixed on the bone (B1) and the last one on the bone close to the plate (B2). Thus, angle
19 variations (θ) between each base were measured according to the imposed load allowing determination of

1 stiffness values (k) of the fixations (variations between B0 and B2) and the implants (variations between B1 and
2 B2) from identification of a linear relationship ($M = k \cdot \theta$).

3 For biomechanical tests on the cranio-cervical junction, alternating sequences of flexion/extension moments of
4 $\pm 2.5\text{N.m}$ can be usually applied (Richter et al., 2000; Wolfla, 2006) but imposed moment values can be much
5 higher in case of dynamic loading during sport practice or collision (Cazzola et al., 2017). To test mechanical
6 behavior of fixations in severe conditions, each specimen was cyclically loaded ramping from -6N.m to $+6\text{N.m}$
7 for flexion-extension movement. To impose these values, loading was in real-time controlled and imposed with a
8 speed of 15mm/min . Motion fields were measured by 3D mark tracking technique with a frequency of 0.5Hz and
9 with an accuracy lower than 0.05mm for displacements and 0.1° for angles.

10 After cyclic loadings, pull-out tests were performed by imposing a ramp extension loading up to bone fracture in
11 order to identify the limit values of moments and angles according to the type of implant and the bone thickness.
12



13
14 Figure 3: Measurements of displacements and angles: (a) 3D mark tracking setup; (b) Definition of reference
15 bases for measurements of angular mobility.

16
17 **2.4. Statistical analysis**

18 Comparison of mechanical responses were performed from variance analyses to evaluate differences in the
19 experimental measurements. Non-parametric ANOVA (Kruskall-Wallis) and multiple comparison testing
20 (Dunn) (Altman and Bland, 2009) to test statistical differences between specimens about stiffness, contact
21 pressure or moment value for pull-out. All statistical analyses were done at a significance level (p -value) equal to
22 0.05 .

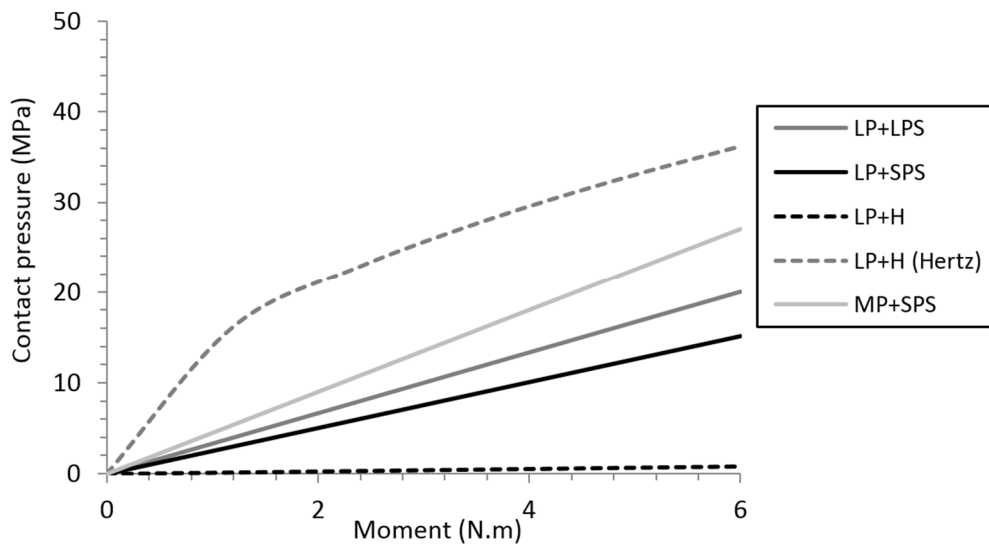
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1 **3. Results**

2 **3.1. Analytical model**

3 From analytical approach, pressures were computed from relationships given in Table 2 and by considering a
4 bone thickness of 4mm for lateral implants and 10mm for median positioning. Imposed loading went up to
5 100N, involving a maximal moment of 6N.m. Figure 4 shows theoretical evolution of contact pressure between
6 implants and bone for the four types of fixations. Among fixations by screws, solution with median plate
7 involves the highest contact pressure (25MPa for a moment of 6N.m). Concerning lateral plates, pressure with
8 SPS is lower than pressure involved with LPS (respectively 16 and 20MPa). For fixations by H, contact pressure
9 is very low if an exact contact is considered (if bone thickness and hook dimension are equal, corresponding to
10 an optimized hook). If bone thickness and hook dimension are not similar, Hertz theory shows a significant
11 increasing of the contact pressure, up to 35MPa for a moment of 6N.m.

12



13

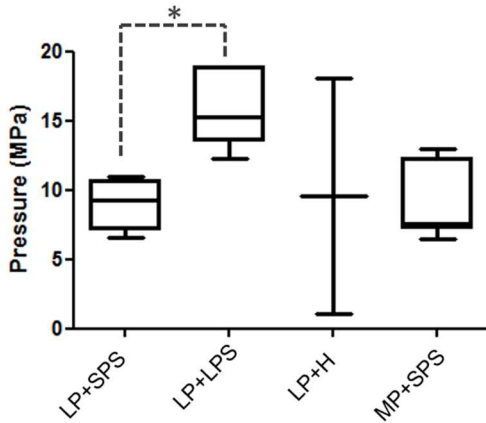
14 Figure 4: Evolution of contact pressure according to imposed moment for the four types of implants

15

16 To complete this preliminary observation, specific contact pressures for each specimen were computed by
17 considering their real values of bone thickness and implant dimensions to determine contact pressure evolution
18 from the relationships and parameters given in Table 2. Figure 5 shows distribution of contact pressures between
19 implants and bone for all the specimens of the four types of fixations and for a moment equal to 6N.m. For
20 fixations by IH, only two values were obtained corresponding to optimized or non-optimized contacts which lead
21 to a large range of pressure between 1 and 18MPa. Contact pressures observed SPS were similar for median or

1 lateral positioning (7-12MPa). Fixations by LPS involved the highest values (12-18MPa). Comparison between
2 the four groups from a statistical analysis highlighted one significant difference between fixations by LP+LPS
3 and fixations by LP+SPS.

4



5

6 Figure 5: Distribution of theoretical contact pressure involved on implants according to the type of fixation

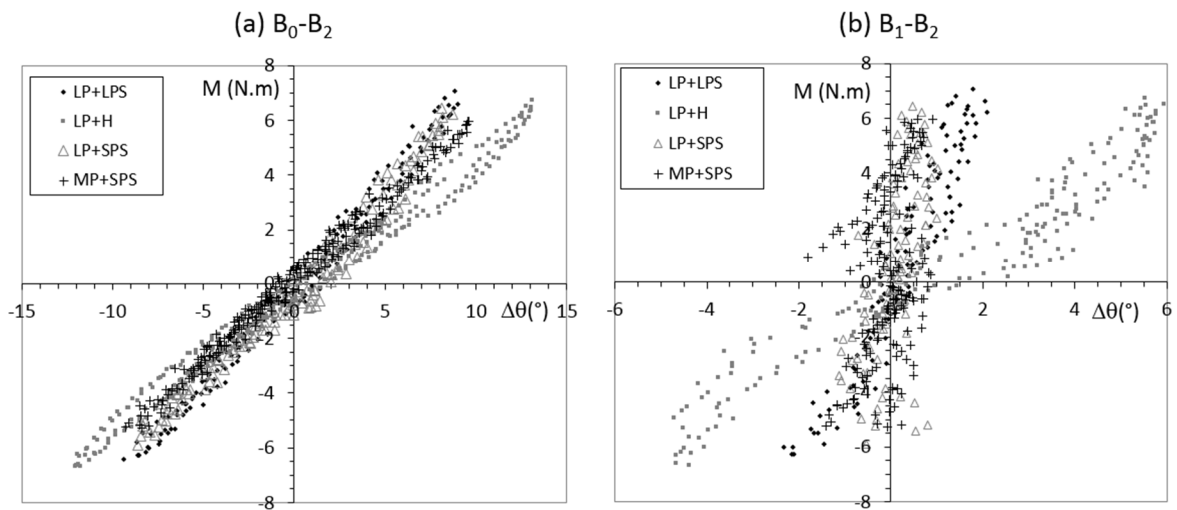
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8 3.2. Experiments: stiffness analysis

9 Figure 6 shows an example of the evolution of angular mobility according to imposed moment for one specimen
10 per group. Global behaviours of fixations (Figure 6-a) were similar even if fixation by IH seemed to be more
11 flexible. This tendency was confirmed with the analysis of angular mobility of implant (Figure 6-b) where
12 LP+IH gave a lower stiffness. These observations were performed for all the specimens of the four groups.
13 Flexibility curves were approached by linear approximation to identify stiffness coefficients for each specimen.
14 Figure 7-a shows distribution of global stiffness values measured between bases B₀ and B₂. Fixation system by
15 MP+SPS seemed to be less rigid than LP whatever the implant. This can be explained by the geometry of MP
16 involving the deformation of fixations with rods. Statistical analysis did not show any significant difference for
17 the global stiffness of fixation.

18 Figure 7-b shows distribution of implant stiffness values. Fixations by LP+SPS gave stiffness with a wide range
19 but values were largely higher than the other fixations. Range values of stiffness were more limited for the others
20 implants. Fixations by H were slightly less rigid than implants by screws. Mechanical responses of LPS in lateral
21 position and SPS with median plate were similar. One significant difference was revealed by the statistical
22 analysis between SPS and H. Differences were also observed between LP+SPS and the two other implants but
23 the significant threshold value was not reached.

1



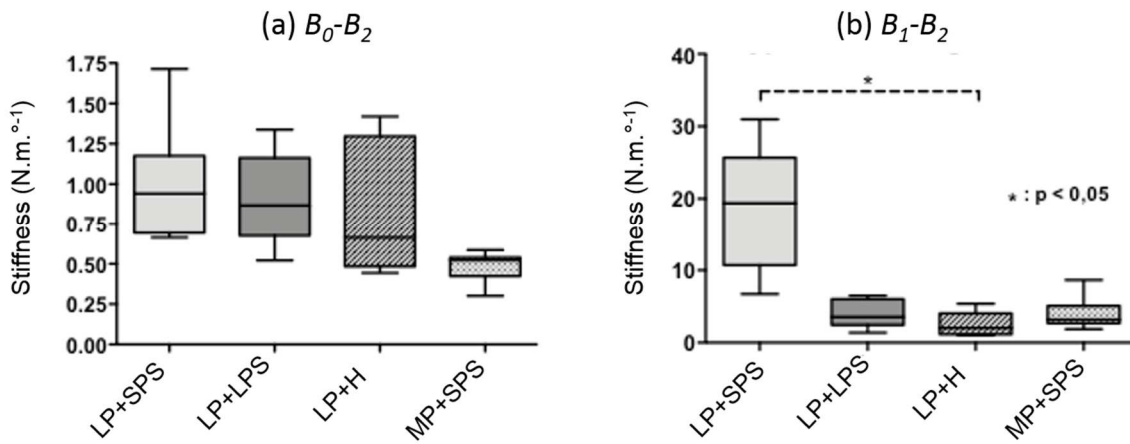
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3 Figure 6: Evolutions of angular mobility for one specimen of each group between metallic vertebra and bone (a)

4 and between plates and bone (b)

5

6



7

8 Figure 7: Stiffness distribution for each group between metallic vertebra and bone (a) and between plates and

9 bone (b)

10

11 3.3. Pull-out tests

12 Figure 8-a shows distribution of moment values observed for pull-out tests. For all the specimens, mechanical

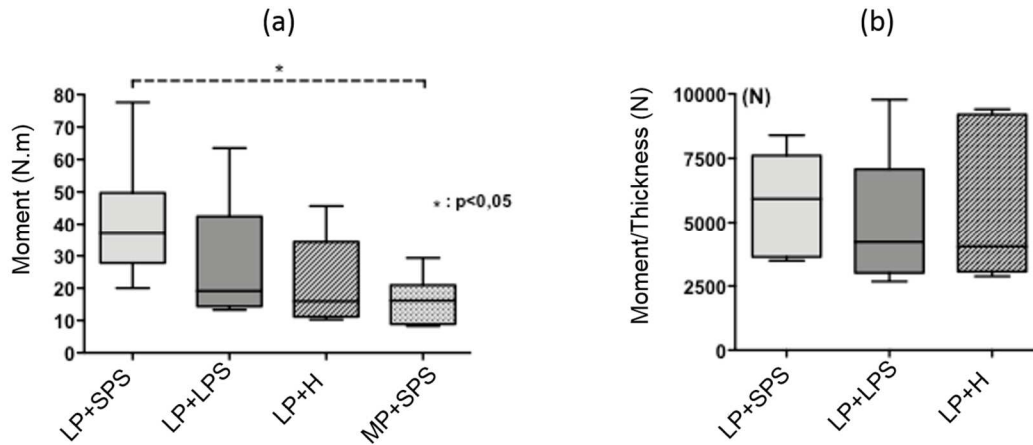
13 loading led to bone fracture at the fixations between implant and bone. Fixations by LP+SPS screws gave the

14 maximum value of moment with a mean value of 35N.m. Values of pull out moments were lower for the other

15 fixations (between 15 and 20N.m). Statistical analysis highlighted one significant difference between fixations

1 by LP+SPS and MP+SPS. These values of moments for pull-out of implants were dependant to the thickness of
 2 the bone. For each specimen, the ratio between moment value for pull out and thickness was computed. The
 3 distribution of these ratio values is shown on the Figure 8-b. This analysis gave similar range of values and no
 4 significant difference was observed.

5



6

7 Figure 8: Distribution of moment values for pull-out (a) and ratio between moment value for pull-out and bone
 8 thickness (b)

9

10 4. Discussion

11 4.1. Flexion-extension tests

12 The study of stiffness values measured between cervical vertebra and occipital bone (B0 vs B2) did not reveal
 13 any significant statistical difference among the four groups. From a clinical point of view, these results do not
 14 reveal any difference about mechanical behaviour of the junction constituted by rods and implants. This global
 15 stiffness did not characterize stiffness implants but is essentially linked to the mechanical response of the rods
 16 and the fixation between rods and implants. This can explain the response given by fixation by MP+SPS. MP
 17 was linked on the occipital bone by screws and rods were assembled on the plate with other screws on small
 18 butterfly extremities, which could be deformed under loading. This double system of fixation and deformation of
 19 plate involved globally a more flexible fixation.

20 Concerning analysis of local stiffness of implants (B1 vs B2), a difference was observed between fixation by
 21 screws and fixation by hooks. A statistical significant difference was observed on fixations with lateral plates,
 22 between SPS and H. This observation of less stiffness on hooks was linked to the thickness differences of
 23 occipital bone and hook size. This was confirmed by the analytical model using theoretical approach of contact

1 pressure. If the hook dimension is equal to the bone thickness, the contact pressure is low and the stiffness
2 should be high. This was observed on some specimens among the fixations by hooks in the tested specimens.
3 Concerning surgery complication ratio, a retrospective study on the comparison between rigid (e.g. screws with
4 plate or rods) and non-rigid (e.g. sublaminar wiring or bone grafting) osteosynthesis systems of the occipito-
5 cervical junction showed superiority of rigid systems from a clinical point of view (Garrido et al., 2011).
6 However, complications appearing among non-rigid systems were less important. Non-rigid systems were
7 different to the ones studied in the present work and were constituted by rods fixed to the bone by threads.

8

9 **4.2. Analytical model**

10 An analytical approach was proposed in this study to determine contact pressure involved in each type of
11 fixation. This analysis revealed a significant difference for LP between LPS and SPS. This difference was due to
12 the pitch dimension of each type of screw. SPS with a smaller pitch (1mm vs 1.8mm) offered a larger surface for
13 load distribution involving a lower contact pressure. Thus, stress distribution in the bone remains lower with this
14 type of implant offering a better ability to allow higher loadings due to movements of the head.

15 According to this analytical model, the best solution for a low contact pressure and a low stress distribution in
16 the bone would be fixation by IH. For that, the necessary condition requests the same dimension between hook
17 and bone thickness. If dimensions are not exactly the same, loading are transferred from a contact line involving
18 mobility and a significant increasing of contact stress which can lead to local damages in the bone.

19 Analytical approach constitutes an original solution to give the mechanical response of occipital implants. The
20 literature does not reveal any study similar with this model. The comparison with the experimental data showed
21 great correlation of the results. The bone dimensions and thickness can be considering from the analytical
22 approach that constitutes a great predictive model of mechanical behaviour of implant to determine contact
23 pressure and stress distribution generated in bone tissues. Thus, it could be possible to adapt the best stabilisation
24 solution in accordance between local bone density / thickness and pressure generated by implant.

25

26 **4.3. Pull-out tests**

27 Pull-out tests revealed a significant difference between LP+SPS and MP+SPS. For both fixations, implant was
28 similar, only nature of the bone was different: bone thickness is higher in the median region but is constituted by
29 cancellous bone whereas lateral region is thicker but with compact bone. The choice of median plate is based on
30 the higher bone thickness but results obtained in the present case are not in accordance with previous

1 observations (Richter et al., 2000; Winegar et al., 2010). Anatomical studies (Hafer et al., 1999) have shown that
2 thickness of cortical bone of external table is larger close to the external occipital protuberance but also that
3 cancellous bone is proportionally present in this area. SPS with a small pitch are well adapted for screwing in
4 compact bone but not optimized for cancellous bone. Thus, implanting cannot benefit of the whole bone
5 thickness, only cortical bone is really solicited by this type of screw.

6 Furthermore, during pull-out tests, moment imposed by cervical vertebra involved a rotation loading on the
7 occipital fixation corresponding to a tensile loading only on the screw closer to vertebra. In the case of lateral
8 plates, this loading is distributed on both sides and so on the two implants contrary to only one implant for
9 median plate. Finally, the choice to distribute implant areas at each side of the protuberance in compact bone
10 with adapted SPS would provide a stronger implanting in the occipital bone.

11 Pull-out tests on fixations by hooks gave results in accordance with analytical approach. This solution should
12 give the most robust anchorage but measured moment values for bone fracture were lower compared to results
13 obtained with SPS (Figure 8-a) but similar if the thickness was considered (Figure 8-b). As commented in the
14 analytical study, this can be due to the difference between hook dimension and bone thickness. Implanting
15 technique of IH requires a burr hole with a sufficient diameter to allow hook positioning in a region where bone
16 thickness decreases. This technique could weaken occipital bone involving locally a lower strength, which could
17 be amplified if hook dimension and bone thickness are different.

18

19 **4.4. Limitations**

20 The main limitation of this work is that experiments were performed on cadaveric specimens with a mean age of
21 72.5 years. Even if the protocol of this study does already include a larger sample size than the other studies
22 available in the literature, the number of samples could be a limitation and it might be great to increase it. It was
23 verified that the tested specimens were pathology-free and this mean age corresponds to the mean ages of
24 specimens tested in the few other available studies.

25 This type of experimental approach is necessary to validate numerical or analytical models. Another limitation of
26 the present study was the measurement resolution of the contact pressure between bone and implant. The bone
27 thickness which was measured with a specific surgical gauge with a resolution of 0.1mm giving a sensitivity of
28 0.3MPa for contact pressure evaluation. This approach was very interesting because it gave relevant results for
29 contact stress analysis but it did not allow evaluation of micro-contact pressure. However, the advantage of this

1 in vitro design is that the stabilization devices are directly used adequately with anatomical considerations.
2 Furthermore, biomechanical observations and conclusions can be directly clinically transposable by surgeons.

3

4 **6. Conclusion**

5 This paper, by focusing on a biomechanical analysis of occipito-cervical fusion, compares four types of occipital
6 implants using an analytical approach combined with experiments on anatomical specimens lead with a specific
7 loading setup and 3D optical measurement method. Proximal stability of occipito-cervical fixation is not only
8 insured by the choice of anatomical site of implantation of occipital screws and plate (conditioned by occipital
9 bone thickness) but is directly influenced by the type of implants. The association of two lateral plates fixed to
10 the occipital bone with small pitch screws appears to ensure an optimal compromise in terms of biomechanical
11 stability and bone/implant/device interfacing. The analytical approach proposed in this work can be considered
12 as an interesting tool to predict mechanical behaviour and stress distribution generated by implant at the level of
13 occipital bone.

14

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