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Mechanical analysis of the healing of different osteotomies fixed externally

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Abstract Under standard conditions the left tibia was osteotomised in nine sheep. The osteotomy was transverse in six, and oblique in three. The osteotomies were fixated using a strain-gauge-monitored unilateral external fixator. The fixation was rigid in three sheep with a transverse osteotomy (group 1) and also in three sheep with an oblique osteotomy (group 2). The fixation was dynamic in three sheep with a transverse osteotomy (group 3) using a roller-bearing mechanism allowing 1 mm of axial displacement. The bone callus stiffness increased progressively with time. One month after osteotomy the load force through the fixator in groups 1 and 2 was more than 20%, whereas the equivalent load force in group 3 was only about 5%.

Résumé Avec un fixateur monolatéral instrumenté avec des jauges extensiométriques (strain gauges) on a étudié l'évolution des forces qui passent par le fixateur et la rigidité du cal de fracture chez neuf agneaux ayant trois types d'ostéotomies du tibia et des fixations différentes (I: ostéotomie transversale avec fixation rigide; II: ostéotomie oblique avec fixation rigide; III: ostéotomie transversale avec fixation dynamique. La rigidité du cal de fracture augmente progressivement dans les trois groupes alors que les forces qui passent par le fixateur diminuent. Après un mois, chez tous les animaux ayant un traitement rigide (groupe I et II) les forces qui passent par le fixateur sont supérieures à 20%, alors que dans le groupe III, à la même période, les forces moment sont approximativement de 5%.

Introduction

Several authors have shown that measuring deflection under the load force of an external fixation frame can monitor the healing of a bone fracture [2–4, 6], and strain-gauge-monitored external fixators have been used to assess the 'stiffness' of fracture union [4, 6, 14–16, 20].

The callus stiffness increases during the process of fracture healing thus allowing increasing transmission of loads through the callus. Therefore the most physiological external fixator or osteosynthesis device should be able to adapt its rigidity according to the state of bone healing in order to produce faster fracture union [18, 21]. As union progresses, the callus itself contributes to the rigidity of the bone-fixator assembly and part of the load is borne through the fracture so decreasing the load through the fixator. The aim of our research was to study the development of callus axial stiffness and the proportion of forces passing through the fixator in transverse and oblique osteotomies externally fixated under either rigid or dynamic conditions.

Materials and methods

Nine female sheep (*Ovis aries*) weighing 18 (16–19) kg were selected for the study, which was approved by the local Ethical Committee. The animals were divided into three groups of three sheep according to the type of tibial osteotomy (transverse or oblique) and the type of unilateral external fixator (rigid or dynamic; Table 1).

A transverse midshaft tibial osteotomy with a 1-mm gap was made in group 1 and group 3, and a 60° oblique midshaft tibial osteotomy with a 1-mm gap was made in group 2. In group 1 and 2 the fixation was static. In group 3 the roller-bearing mechanism of the Monotube-Triax (Stryker-Howmedica) external fixator was adjusted to allow 1 mm of axial displacement and this allows compression. Axial dynamisation was allowed by the telescoping action of the fixator bar.

Radiographs were taken immediately after surgery and at 7, 15, 21 and 30 days, and each animal was examined daily for signs of pin-tract infection. Measurements lasting 30 s were taken each day for at least 4 weeks. The animals were trained to put the frac-

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Table 1 Overview of experimental conditions

Group	Fracture type	Mechanical fixation	n	Evaluation (days)
1	Transverse	Rigid	3	30
2	Oblique 60°	Rigid	3	30
3	Transverse	Dynamic	3	30

Fig. 1 Effect of loads on inter-fragmentary deformation ($\Delta\delta$) and on the external fixator

tured leg onto the force plate. Both the fixator and the force plate were connected to a computer (Macintosh IIfx; Apple, Cupertino, Calif., USA) with an analogical-digital converter for the acquisition and processing of data (Lab View 2; National Instruments, Austin, Tex., USA). A strain-gauge amplifier (San-ei, N9224; AC Strain Amplifier, Tokyo, Japan) between the fixator and the computer was used to pick up the microvolts transmitted through the bar of the fixator. For each reading the software collected an average mean of 500 data points taken in 1 s and then transformed the information into newtons (N).

With this method we could register at the same time the total axial force (F_z) and the force supported by the fixator (F_f). This allowed us to calculate the axial bone callus stiffness (K_b) as:

$$F_z = (K_f + K_b) * \Delta\delta \quad (1)$$

where $F_b = K_b * \Delta\delta$ (force supported by the bone); $F_f = K_f * \Delta\delta$ (force supported by the fixator); K_f =axial stiffness and K_b =axial bone callus stiffness.

The callus stiffness (K_b) can be expressed as:

$$K_b = K_f \left(\frac{F_z}{F_f} - 1 \right) \quad (2)$$

When axial load F_z is applied to the bone, the force shared between the bone and the fixator device depends exclusively on the rigidity of each of these elements (Fig. 1), although there is three-dimensional movement of the fracture fragments even under axial loads because of the position of the fixator [11].

It is assumed that the total force acting on the bone is F_z , which is the sum of the force supported by the fixator (F_f) and the force supported by the bone callus (F_b). The callus stiffness (K_b) increases as the fracture heals, so this parameter can be used as an index of bone fracture healing [9].

K_f has a known value for the type of external fixator used in this experiment (Monotube; Howmedica) and only varies according to the measurable distance between the frame and the bone. F_z

can be measured with a force plate. To obtain F_f , three unilateral external fixators (Monotube-Triax; Stryker-Howmedica) were fitted with strain-gauge sensors [10, 22], replacing a section of the lengthening bar. This measures the axial force transmitted through the external fixator and has very good linearity. It does not detract from the frame's rigidity [22].

Results

In group 1 the percentage of forces through the fixator decreased progressively to reach about 40% in the 2nd week. In the next 2 weeks they decreased more slowly and were about 20% at the 4th week (Table 2). As the process advanced, a quick falling-off of the forces in the first 10 days, and then later on a more gentle decline was observed. The axial bone callus stiffness (K_b) increased progressively to reach 294 N/mm at the 3rd week and nearly 392 N/mm by the 4th week.

In group 2 a more gentle decrease than in group 1 was seen in the 1st week. There was thereafter a slow decrease from the 2nd to the 3rd week and a very gentle decrease until the 4th week (Table 2); 35% was achieved in the 4th week. The axial bone callus stiffness (K_b) increased very gently to reach 137.2 N/mm by the 4th week.

In group 3 the percentage of forces crossing the fixator the first day and before telescoping was allowed was 100%, but when the dynamisation started these forces were distributed between the bone and the fixator. This distribution had different proportions for each animal, with a mean of 50% for the bone and 50% for the fixator. In the 1st week the forces borne by the fixator were an average of about 55%. In the 2nd week the percentage started to decrease gently and by the 4th week was around 10% (Table 2). The axial bone callus (K_b) increased progressively to reach 490 N/mm at the 3rd

Table 2 Weekly percentage of forces through the external fixator – mean (x) and standard deviation (SD)

Week	Group 1		Group 2		Group 3	
	x	SD	x	SD	x	SD
1	77.26	8.23	85.73	9.12	55.75	7.45
2	47.88	7.18	64.31	6.43	29.26	6.68
3	31.09	6.73	53.70	6.19	16.03	4.13
4	22.42	4.36	34.37	6.32	7.86	2.78

Table 3 Weekly callus stiffness (N/mm) – mean (x) and standard deviation (SD)

Week	Group 1		Group 2		Group 3	
	x	SD	x	SD	x	SD
1	30.57	6.75	9.11	2.57	41.62	11.86
2	98.71	16.05	28.77	7.34	221.60	32.02
3	207.23	26.63	52.61	9.63	478.75	36.28
4	331.05	46.37	96.27	22.66	1078.41	94.76

Fig. 2 Weekly mean callus stiffness evolution in group 1 (stable), group 2 (unstable) and group 3 (dynamic)

week in all animals in this group. After this there was a quick rise to reach 980 N/mm in one animal and to 1960 N/mm in another.

The statistical analysis of axial stiffness as a weekly mean reveals differences among each of the three groups as far as time is concerned ($P < 0.001$), as shown in Table 3 and Fig. 2. When comparing all the measurements the weekly mean stiffness of each group showed no statistical difference in the 1st week ($P = 0.092$). In the 4th week very high statistical differences were found ($P < 0.001$) between group 1 and 2 compared to group 3; but without differences between group 1 and group 2 ($P = 0.081$).

Discussion

The use of engineering techniques to measure strains occurring in various types of implant offers valid information that is useful from both a clinical and a research point of view. External fixation makes it possible to place instruments within the apparatus in order to measure the forces that are transmitted through it. This makes it an extremely useful method for determining the rigidity of the fracture callus [2–6, 9] which can be used to monitor the evolution of fracture healing [16].

From the mechanical point of view the most important aspect of an external fixator is the rigidity of the system against the four different types of stresses that can be found at the fracture site: axial, bending (AP and lateral) and torsion.

When an axial load is applied to an externally fixed fracture, forces are shared between the bone itself and the fixator device, and this depends on the rigidity of these two elements and the placing of the pins. Immediately after fracture the bone fragments are completely separate and nearly all the stress forces are transmitted through the fixator. As union occurs the new bone forming at the fracture site is able to support part of these loads and this has been confirmed in all groups of this study.

The gradual tissue changes which occur during bone healing are characterised by their increased rigidity with an elastic modulus developing, which is more than 200,000 times greater for mature bone than for granulation tissue [19]. According to Perren's interfragmentary

deformation theory each tissue tolerates different amounts of deformation and if this is exceeded then the tissue can 'break' [19]. The tissue changes correlate with the gradual decrease in interfragmentary movement, as granulation tissue is able to tolerate 100% of 'tension'; fibrous tissue 20%; cartilage 10% and bone just 2%. White et al. [23] defined four stages of mechanical fracture healing. Tibial fractures fail in stage 1 through low stiffness at the original fracture site. In stage 2, there is increased rigidity. In stage 3 failure is through the fracture site and through intact bone, and in stage 4 failure is entirely through intact bone which has a high rigidity.

Dynamisation does not necessarily imply movement, and an external fixator is dynamised in order to change its mechanics and thus modify the rigidity but not the stability. This allows axial loads to cross the fracture site [1]. This biocompression produces better contact of the fracture fragments with adaptation to the physiological conditions of the interfragmentary tissue [17]. This increases the stiffness and was observed in group 3. However, as Hart et al. [13] concluded, compression applied through an external fixator system increases the rigidity of fixation but does not benefit union.

Different authors have studied the evolution of forces passing through the external fixator and its relation to the stiffness of the bone callus [1, 2, 4–8, 11]. This has been used for the early detection of pseudarthrosis and also to decide the best time to dynamise the fixator and then to remove it. Other authors have established the level of 15 Nm/° as the minimal stiffness which will allow safe removal of the external fixator in human tibial fractures, and 20 Nm/° in femoral fractures [12, 14, 15, 20]. Nevertheless, the value of stiffness is not the only criteria to allow fixator removal. Others include the time of follow-up, the X-ray image, the patient's weight and the residual angle of the fracture.

Our study demonstrates that bone callus stiffness in healing fractures increases progressively in the three groups as the loads through the fixator decrease, but with a different rate and magnitude in each of the groups. This depends on the mechanical properties of both the fracture and the fixator.

Contrary to other authors [2], we have found that after 1 month of treatment none of the animals under rigid conditions (group 1 and group 2) experienced less than 20% of load forces through the fixator. However, those

in group 3, under biocompression to permit contact between the fragments, had load forces through the fixator of about 5%.

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References

1. Aro HT, Chao EYS (1993) Bone healing patterns affected by loading, fracture fragment stability, fracture type, and fracture site compression. *Clin Orthop* 293:8–17
2. Beaupre GS, Hayes WC, Jofe HH, White AA (1983) Monitoring fracture site properties with external fixation. *J Biomech Eng* 105:120–123
3. Bourgois R, Burny F (1972) Measurement of the stiffness of fracture callus in vivo. A theoretical study. *J Biomech* 5:85–92
4. Burny F (1968) Etude par gauges de deformation de la consolidation des fractures en clinique. *Acta Orthop Belg* 34:917–927
5. Burny F (1979) Strain measurements of fracture healing. In: Edwards AFH (ed) *External fixation. The current state of art*. Williams and Wilkins, London, pp 371–381
6. Burny F, Bourgois R, Donkerwolcke M, Moulart F, Hinsenkamp M, Nicod C, Greer-d'Hollander H (1976) Mesure de la deformation des implants "in vivo". Application a l'étude de la consolidation des fractures traitees par osteotaxis. *Acta Orthop Belg* 42 [Suppl 1]:62–74
7. Chao EYS, Hein TJ (1988) Mechanical performance of the standard Orthofix external fixator. *Orthopedics* 11:1057–1069
8. Cordey J, Schnetzer M, Berwald J, Regazzoni P, Perren SM (1980) Direct "in vivo" measurements of torque and bending in sheep tibiae. In: Uthoff C (ed) *Current concepts of internal fixation of fractures*. Springer, Berlin Heidelberg New York, pp 78–87
9. Cunningham JL, Evans M, Kenwright J (1989) Measurement of fracture movement in patients treated with unilateral external skeletal fixation. *J Biomech Eng* 11:118–122
10. Forriol F, Goenaga I, Mora G, Vinolas J, Canadell J (1997) Measurement of bone lengthening forces; an experimental model in the lamb. *Clin Biomech* 12:17–21
11. Gardner TN, Evans M (1992) Relative stiffness, transverse displacement and dynamisation in comparable external fixators. *Clin Biomech* 7:231–239
12. Hammer R, Edholm P, Lindholm B (1984) Stability of union after tibial shaft fractures: analysis by a non-invasive technique. *J Bone Joint Surg [Br]* 66:529–534
13. Hart MB, Wu JJ, Chao EYS, Kelly PJ (1985) External skeletal fixation of canine tibial osteotomies. *J Bone Joint Surg [Am]* 67:598–605
14. Jernberger A (1970) Measurement of stability of tibial fractures. A mechanical method. *Acta Orthop Scand Suppl* 135
15. Jorgensen TE (1972) Measurements of stability of crural fractures treated with Hoffmann osteotaxis. 2. Measurements of crural fractures. *Acta Orthop Scand* 43:207–218
16. Kristiansen B, Borgwardt A (1992) Fracture healing monitored with strain gauges. External fixation of seven humeral neck fractures. *Acta Orthop Scand* 63:612–614
17. Lazo Zbikowski J, Aguilar F, Mozo F, Gonzalez R, Lazo JM (1986) Biocompression external fixation. Sliding external osteosynthesis. *Clin Orthop* 206:169–184
18. McKibbin B (1978) The biology of fracture healing in long bones. *J Bone Joint Surg [Br]* 60:150–162
19. Perren SM, Rahn BA (1980) Biomechanics of fracture healing. *Can J Surg* 23:228–232
20. Richardson JB, Cunningham JL, Goodship AE, O'Connor BT, Kenwright J (1994) Measuring stiffness can define healing of tibial fractures. *J Bone Joint Surg [Br]* 76:389–394
21. Sarmiento A, Gersten LM, Sobol PA, Shankwiler JA, Vangsness CT (1989) Tibial shaft fractures treated with functional braces: experience with 780 fractures. *J Bone Joint Surg [Br]* 71:602–609
22. Vinolas J, Anora JR, Forriol F, Goenaga I (1993) A transducer for the measurement of forces in external fracture fixator devices. Test results. *Sensors Actuators A* 37–38:301–306
23. White AA, Panjabi MM, Southwick WO (1977) The four biomechanical stages of fracture repair. *J Bone Joint Surg [Am]* 59:188–192