A Finite Element Model to Simulate Femoral Fractures in Calves: Testing Different Polymers for Intramedullary Interlocking Nails

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Objective: To verify if the finite element method can correctly estimate the performance of polyacetal and polyamide 6 intramedullary nails in stabilizing a femoral fracture in calves and to estimate the performance of a polypropylene nail in same conditions.
Study Design: Computational and experimental study.
Sample Population: Finite element models (FEMs).
Methods: Based on a 3-dimensional finite element method (FEM) of the femoral diaphysis, 3 models were constructed to simulate an oblique simple fracture stabilized by an intramedullary nail composed of 1 of 3 distinct polymers. Models were tested under 6 loading conditions that simulated a static calf or a calf in different walking phases. Maximum bone and implant stresses were compared to yield and rupture stresses of specific materials.
Results: Under static conditions, all polymers were resistant to critical deformation and rupture because maximum von Mises stresses were lower than the respective yield and rupture stresses. However, during walking, maximum stresses exceeded the yield and rupture limits of the polymers, in agreement with a previous *in vivo* study, which used polyacetal and polyamide nails.
Conclusions: FEM correctly estimated that polyacetal and polyamide 6 nails would fail to immobilize an oblique femoral diaphyseal fracture in calves that were allowed to walk freely during the early postoperative period. FEM can be useful in the development of new bovine orthopedic devices.

Despite recent developments, long bone fractures in large animals, especially in the humerus and femur, are still considered to be a challenge for veterinary surgeons. Because of their substantial mass, large animals cannot bear weight or move properly using only 3 limbs, even for short periods of time. This commonly leads to complications such as delayed healing, malunions, infections, prolonged recumbency with consequent pressure wounds, and contralateral limb problems (eg, laminitis). In many cases, euthanasia is still considered as a choice to avoid further financial loss and stop suffering.[1]

In human orthopedics, metallic plates, screws, intramedullary rods, and transcortical pins with external fixation have been successfully used to treat most long bone fractures. However, an implant that provides enough strength to support immediate full weight bearing and optimal conditions for fracture healing in large animals has not been developed and may require entirely new technologies and a different approach.[1]

Alternative polymeric materials such polyacetal and polyamide intramedullary nails have been used in a rat fracture model.[2,3] In those studies, 16 weeks after fracture, femora treated with polymeric nails had an increased callus cross-sectional area,[3] and tibiae fixed with polymeric nails were 38% stronger than similar bones fixed with metallic rigid nails.[2] Polyacetal intramedullary interlocking nails have also been successfully used to treat osteoporotic femoral and tibial fractures in people.[4] The medullary canals of osteoporotic bones have a larger diameter and a thinner cortical compared with normal bones. These characteristics are similar to those found in bones of newborn calves, especially the humerus.

The use of polymeric intramedullary interlocking nails in a young calf fracture model has been considered.[5–8]

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Initial studies demonstrated clinical healing in humeral fractures fixed with intramedullary polypropylene nails, indicating their potential for clinical use;[5] however, polyacetal and polyamide nails failed when used to fix femoral fracture in the same model.[7] In addition, an *ex vivo* study using an universal testing machine demonstrated that calf's femora fixed with polypropylene nails were comparable to the intact femora when subjected to compressive forces, but not when subjected to bending forces.[8] Nevertheless, polypropylene nails have not yet tested in femoral fractures using the *in vivo* model.

It is widely accepted that the finite element method (FEM) is one of the most practical and reliable methods in the field of engineering for analyzing mechanical structures. Advances in computer technology in recent years have further promoted the use of this numerical method. FEM was introduced to orthopedic biomechanics in 1972 to evaluate stresses in human bones. Since then, FEM has been applied with increasing frequency in stress analyses of bones and bone-prosthesis structures, fracture fixation devices, and various kinds of tissues other than bone.[9,10]

In veterinary medicine, few studies have used this method, including (but not limited to) models of the metacarpus[11] and laminar hoof junction[12] in horses, studies of deformation[13] and the effect of floor type[14] on bovine claws, and simulations of oral implants in the mandibles of dogs.[15] One of the great advantages of the FEM is that once it is developed and validated, innumerous predictions can be conducted. This avoids the high cost and unnecessary use of live animals, which is common in conventional studies.

With the hypothesis that a computational numerical method is able to correctly estimate the capacity of polymeric nails (polyamide 6 or polyacetal) to stabilize bone diaphysis in a model of femoral fracture in calves and that polypropylene nail is not strong enough to resist the mechanical forces in such conditions, our aim was to use the FEM to evaluate the performance of polymeric nails, and to compare the results obtained with polyacetal and polyamide 6 with previously published *in vivo* studies that used similar procedures.

MATERIALS AND METHODS

Geometric Modeling of a Calf Femur

A femur obtained from a clinically normal 15-day-old Holstein male calf in a slaughterhouse was used to acquire the image data. After removal of the soft tissue by dissection, a computed tomography (CT) scanner (Brilliance16; Philips Medical Systems, the Netherlands) was used to generate 226 images of the bone in digital imaging and communications in medicine (DICOM) format. To obtain the FEM of the diaphysis of the bovine femur, CT images were processed using a computer-aided design (CAD)-based modeling pipeline.[16] Briefly, in the first step, these images were segmented with 3-dimensional active contours,[17] each of which changed in space and time according to a regional competition algorithm. This segmentation method was successfully implemented in ITK-SNAP.[18] To obtain a continuous surface representation from the digital images, a point cloud was extracted from the resulting segmented data using morphological image operators from the Image Processing Toolbox 6.2 (Matlab, Matrix Laboratory, Secunderabad, India). Several spline curves were interpolated from the point cloud, and the bone surfaces were then obtained by sweeping these curves. After closing the endings of the diaphysis, a boundary representation (B-REP) of the femur diaphysis, also designated as a solid model, was generated. The solid model and the corresponding finite element meshes were obtained using CUBIT software (Sandia National Laboratories, Albuquerque, NM).

Finite Element Modeling

Based on the volumetric model of the femoral diaphysis, 3 models were built to simulate an oblique simple fracture (40°, grade A2 by AO/ASIF score system).[19] Bone segments were stabilized by an intramedullary nail made of 1 of 3 distinct polymers: polypropylene, polyamide 6, or polyacetal. The interface of bone segments was modeled with frictionless contact. Nails were constructed with the same length of the diaphysis and were 1 mm shorter in diameter than the medullary cavity. No contact occurred between the bone surface and the nail. Each model of a femoral diaphysis and intramedullary nail complex was equally interlocked by four 4.5-mm cortical screws (stainless steel) that were inserted perpendicularly (2 in the proximal and 2 in the distal bone segments). In each bone segment, the 1st screw was located 10 mm from the fracture line, and the 2nd was located 10 mm from the 1st screw. Both extremities of each screw exceed the cortical tissue by 2 mm.[20] Nails and screws were built using cylinders as primitive geometries. The screws were modeled without heads or threads. Nails and screws were included in the femoral model using Boolean operations, and the contact surfaces were defined using CUBIT software.[21] In the next step, tetrahedral meshes of each model were produced using the TetMesh technique,[21] and routines were developed in the native language of CUBIT (journal files, *.jou) to generate output files with *.inp extensions, which were used as input to the finite element analysis software Abaqus (Dassault Systèmes, Paris, France). Contact interactions among the different materials were established by considering the screws to be totally bonded to the bone and nails. All the materials, cortical bone, nails, and screws, were modeled as homogeneous, isotropic, and linear elastic, with mechanical properties given in Table 1.[22] The finite element mesh was formed by 68,202 tetrahedral 4-node elements with 43,584 elements for the diaphysis.[3] Results were obtained through nonlinear contact analysis and no failure criterion was considered in the finite element simulation.

The 3 different models were subjected to 6 different predetermined loading conditions that simulated a static calf (loading condition 1) and a calf in different phases of a

Table 1 Mechanical Properties of Materials Used to Compose the Models of the Interlocking Nails[21]

Mechanical properties	Cortical bone	Stainless steel	Polypropylene	Polyacetal	Polyamide
Sy – Yield point (MPa)	114 (T)*	205	26.5	68.5	49
Sr – Rupture point (MPa)	133 (T)* 205 (C)*	515	30.5 (T)* 37 (C)*	72.5 (T)* 75 (C)*	67 (T)* 80 (C)*
E –Young modulus (GPa)	21.9	210	1.3	3	2.1
ν – Poisson ratio	0.30	0.35	0.43	0.41	0.40

*T, tensile; C, compression.

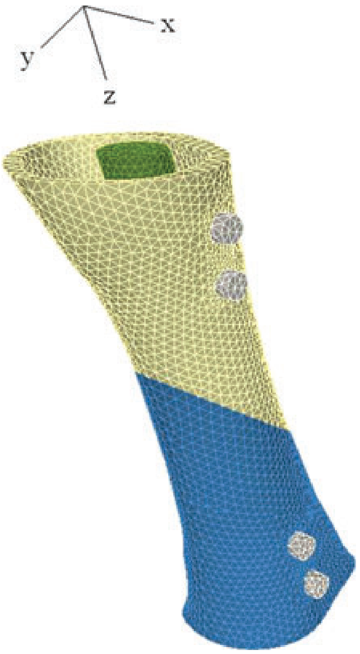

Figure 1 Finite element model of the bovine femoral diaphysis with an oblique fracture stabilized by a polymeric interlocking nail.

walking gait (loading conditions 2 to 6). The loading conditions considered 3 directions in relation to the axis of the bone, lateromedial (x), craniocaudal (y), and proximodistal (z), and each condition had a specific force and moment for each direction (Fig 1; Table 2).

For the establishment of the different load cases, the vertical ground reaction force was previously determined in laboratory considering static and walking conditions.[23] The experiments were performed using calves with weights ranging from 539 N–1333 N. Longitudinal and transversal components were obtained for an average weight of 809.5 N and were based on the measured value of the reaction force and previous results using dairy cattle.[24] A statically equivalent system of loads was then evaluated to be applied to the diaphysis. The displacement boundary condition of the distal end of the diaphysis was fixed. The loads are applied on the proximal diaphysis by means of a rigid plate attached to the proximal femoral extremity. This rigid plate is introduced in the model to simulate realistic loads on the shaft since it allows forces and moments to be applied, and it distributes these loads all over the proximal femoral extremity.

Table 2 Values for the Forces and Moments in Standing (1) and Walking Conditions (2–6) Applied to Bovine Femoral Fracture Models

		Direction		
Condition	Load	X	y	Z
1 Standing	Force (N)	0	70.82	154.09
	Moment (N.mm)	– 5750	0	0
2 Heel strike	Force (N)	11.6	21.08	120.8
	Moment (N.mm)	– 11,580	9150	– 4845.8
3 Maximum braking	Force (N)	21.44	44.67	398.3
	Moment (N.mm)	13,000	16,310	– 2530
4 Midstance	Force (N)	5.73	106.16	324.11
	Moment (N.mm)	– 8570	4100	– 1130
5 Maximum propulsion	Force (N)	8.28	122.3	251.59
	Moment (N.mm)	4160	5570	– 2840
6 Push off	Force (N)	0	13.8	31.24
	Moment (N.mm)	4130	0	0

The maximum von Mises stresses were recorded for each component of the model during each loading condition. The results are presented in Tables 3 and 4, and in Figs 2 and 3. In the latter case, they are presented as relative values compared with the yield and rupture points for each material. The results from each femoral fracture model were obtained to identify any values above the yield and rupture stresses for each of the component materials, which were considered to be indicative of failure.

RESULTS

Values for von Mises stresses in the bone and nails are presented in Table 3. In the 3 models, bone stress values were always below the yield and rupture points for cortical bone. The highest value was registered using polypropylene (113.2 MPa), which was very close to the cortical bone yield point (114 MPa; Fig 2). When polypropylene and polyamide were used, nail stress values surpassed their respective polymeric yield points under loading conditions 2 and 3 and their respective polymeric rupture points under loading condition 3. When polyacetal was used, the polymeric yield and rupture points were overlapped under loading condition 3. The maximum stress reached 244% of the rupture value when polypropylene was used (Fig 3).

Table 3 Maximum von Mises Stresses (MPa) in Bones and Nails from Different Polymeric Interlocking Nails Subjected to Different Loading Conditions in a Model of a Bovine Femoral Fracture

		Loading condition						
Component	Nail material	1	2	3	4	5	6	Mean
Bone	Polypropylene	6.05	39.61	113.20	16.67	48.13	17.81	40.25
	Polyacetal	6.04	38.39	77.56	16.05	46.63	16.41	33.51
	Polyamide	6.05	37.48	89.25	16.87	46.80	17.34	35.63
Nail	Polypropylene	2.16	57.71	90.50	21.97	17.80	0.95	31.85
	Polyacetal	2.33	57.89	89.11	22.58	18.67	1.59	32.03
	Polyamide	2.27	57.71	90.18	22.30	18.08	1.12	31.94

Table 4 Maximum von Mises Stresses (MPa) in the Screws Located in Different Polymeric Interlocking Nails Subjected to Different Loading Conditions in a Model of a Bovine Femoral Fracture

			Loading condition						
Bone segment	Screw	Nail material	1	2	3	4	5	6	Mean
Proximal	Proximal	Polypropylene	4.47	18.28	30.80	11.98	13.97	2.78	13.71
		Polyacetal	5.45	18.72	34.36	14.30	14.92	3.73	15.25
		Polyamide	4.86	18.40	31.70	12.95	13.98	2.99	14.15
	Distal	Polypropylene	11.50	109.80	194.30	54.91	42.10	3.56	69.36
		Polyacetal	10.16	104.80	181.90	50.83	40.20	4.09	65.33
		Polyamide	11.07	106.10	186.90	53.89	41.82	3.69	67.24
Distal	Proximal	Polypropylene	12.14	108.50	191.40	54.13	52.50	6.16	70.80
		Polyacetal	10.28	106.40	179.60	49.39	45.57	6.18	66.24
		Polyamide	11.55	108.60	184.80	53.22	50.69	6.15	69.17
	Distal	Polypropylene	5.33	13.27	43.04	10.97	18.31	4.08	15.83
		Polyacetal	6.64	14.76	44.95	12.77	19.66	4.55	17.22
		Polyamide	5.90	13.52	43.63	18.87	18.59	4.27	17.46

Values for the von Mises stresses in the screws were below the yield stress of stainless steel (205 MPa) under all loading conditions (Table 4). However, under loading condition 3, the von Mises stresses in screws that are closer to the fracture line approached the yield stress limit.

DISCUSSION

In the present study, computational finite element analyses were performed to estimate the effects of the loading weight of a young calf (in static or walking conditions) with a very recent femur break in one of its pelvic limbs (simple oblique diaphyseal fracture), which was stabilized with a polymeric (polypropylene, polyacetal, or polyamide) interlocking nail. In the simulation of the static condition, all of the materials used were resistant to deformation and rupture because maximum von Mises stresses were below each yield and rupture stresses. However, when walking was simulated, the maximum stresses exceeded the yield stress for all polymers, indicating that none of polymeric nails would have enough strength to support normal walking forces in an *in vivo* study.

Our results are in agreement with a similar *in vivo* study[7] in which 6 of 7 polyacetal nails and 2 of 4 polyamide nails broke a few days after intramedullary placements in 3-month-old calves subjected to similar fractures. In the *in vivo* study, the nails failed 3–11 days after placement when polyacetal was used and 12 and 14 days after placement when polyamide was used.[7]

The maximum von Mises stresses in the screws were always lower than the yield and rupture stresses of stainless steel in all models (Table 4). However, in the screws adjacent to the fracture line, these stresses reached values that were very close to the yield point. Moreover, values in these adjacent screws were always higher than the values in distant screws (a mean 4.4-fold increase). These results are in agreement with the *in vivo* study in which all of the nails were broken at the interface with one of the screws that was closer to the fracture line.[7] Retrospective human studies also reported that interlocking nails often failed at the screw hole near the fracture line.[25,26] The 1st screw hole of the distal third of the bone is the site that is thought to be most at risk, especially if it is near the fracture site.[26]

Although the results from this study compare well with what happened *in vivo*,[7] some limitations exist because of modeling assumptions. One limitation of this work is to assume bone as a homogenous isotropic material. However, in this case, this assumption is justified because we are only modeling compact bone. The apparent density of compact bone does not vary too much from point to point, and thus the Young modulus can be assumed to have a single value. Furthermore, in the diaphysis of long bones, compact tissue is transversely isotropic, but for the aim of this work, that is to analyze the behavior of the implant, the assumption of bone isotropy was considered a reasonable approach.

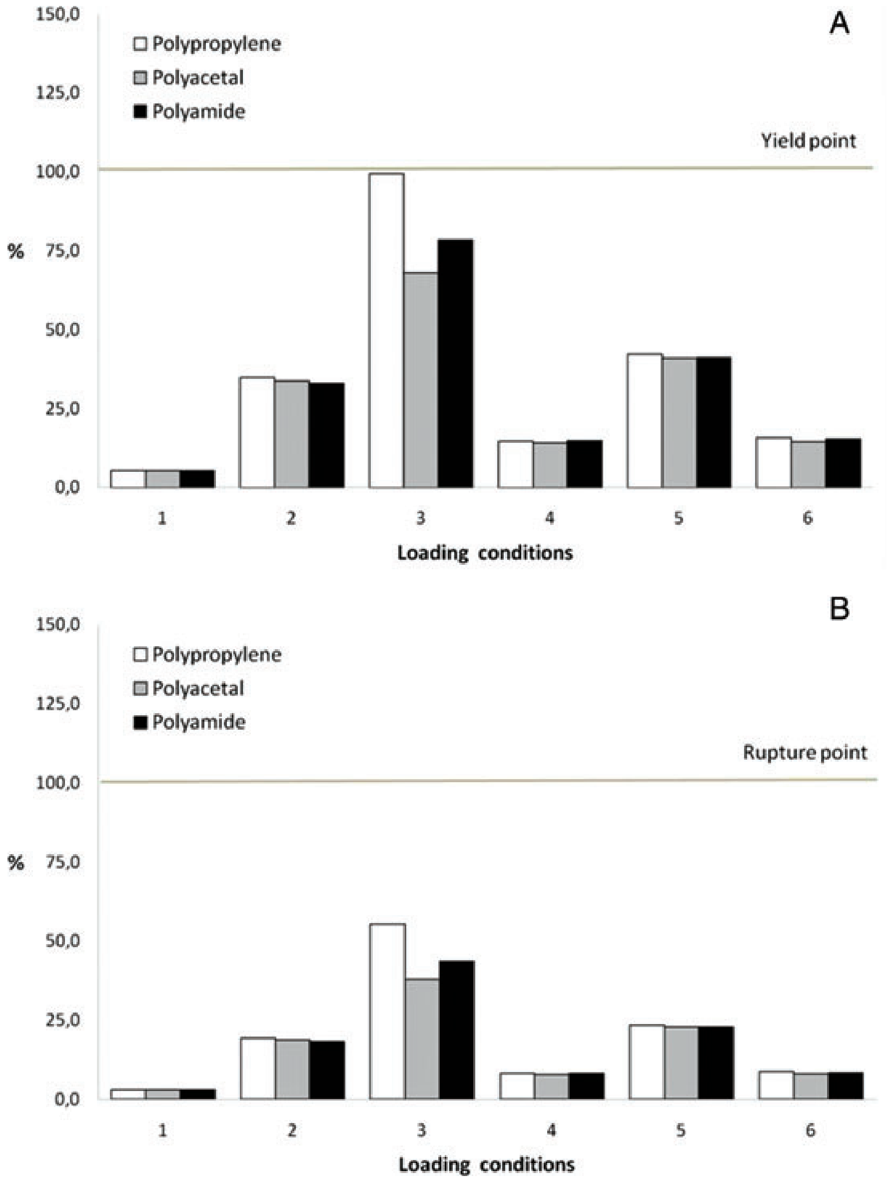

Figure 2 Percentages of maximum von Mises stresses in bones with different polymeric interlocking nails in relation to bone yield point (A) and compressive rupture point (B) when subjected to different loading conditions (1 – standing, 2 – heel strike, 3 – maximum braking, 4 – midstance, 5 – maximum propulsion, 6 – push off) in a model of a bovine femoral fracture.

Moreover, although bone and screws are in fact not bonded, the screw geometry is such that no sliding can be assumed. Thus, for the purpose of our work, the assumption of bonded is reasonable. Although full discretization of the screws and the modeling of contact may be possible to be achieved, this will demand a heavy computational effort.

Another limitation is that healing was not considered. In fact only immediate postoperative conditions were modeled, considering bone segments in contact without friction. To model the healing process and the evolution of the mechanical state, a more complex model should be used.[27] Actually, during the healing process, the fracture stiffness changes, influencing the global performance of the construct. However, for the immediate postoperative situation, contact between bone segments is a reasonable approach since at this stage, the fracture stiffness is negligible.

With respect to the FEM, tetrahedral 4-node elements were used because these elements allow us to model the complex geometry of bone, nail and screws without a significant computational cost. The use of other element type could improve the solution accuracy, but in previous works[6] where different meshes were tested, the results did not show significant differences.

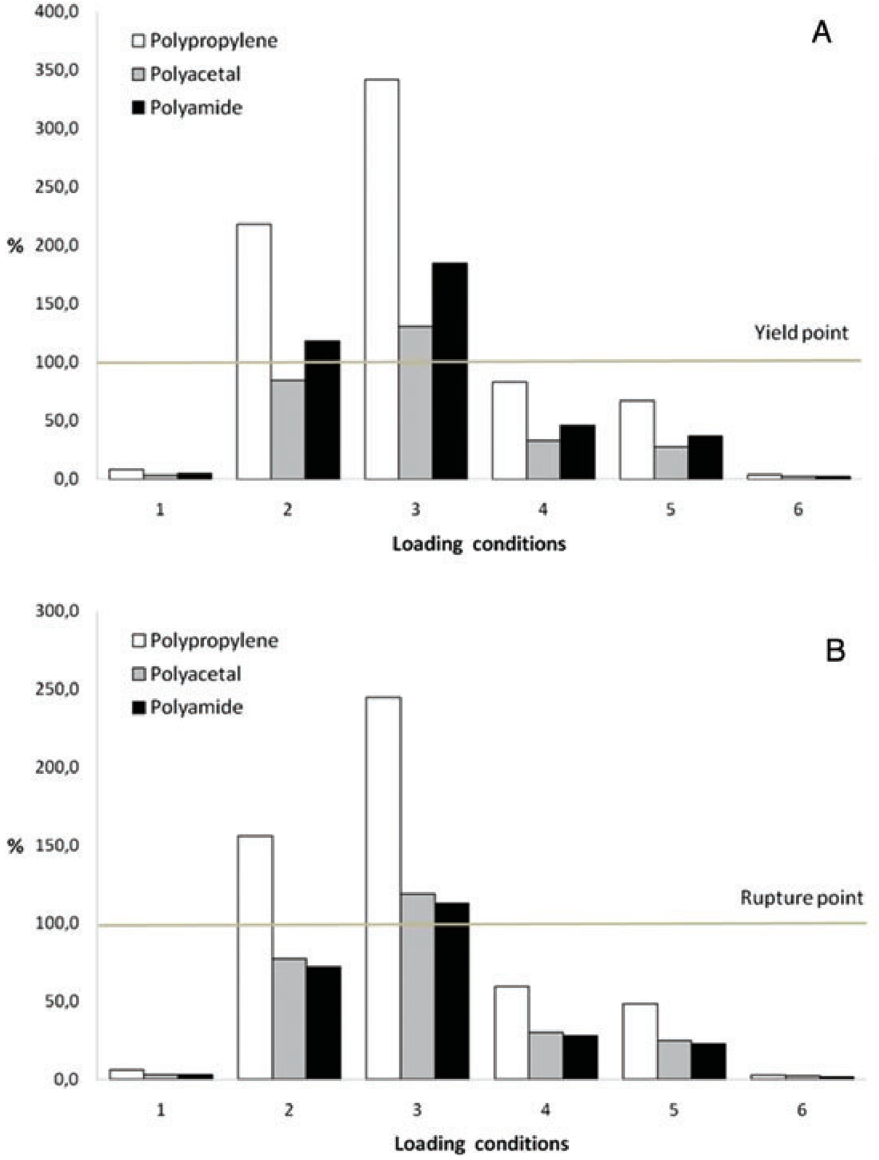

Figure 3 Percentages of maximum von Mises stresses in the different polymeric interlocking nails in relation to their respective yield point (A) and compressive rupture point (B) when subjected to different loading conditions (1 – standing, 2 – heel strike, 3 – maximum braking, 4 – midstance, 5 – maximum propulsion, 6 – push off) in a model of a bovine femoral fracture.

Moreover, further developments should be considered. In an attempt to simulate a real scenario, all of the loading conditions were designed to represent multiple loads, including traction, bending, compression, and torsion efforts, simultaneously. However, weight bearing during movements involved in lying down and getting up was not taken into account in the present model. Under such conditions, the femur is subjected to forces that cause bending, especially when the hind limb has to project the body upwards when the animal gets up. The force generated during this movement may be much higher than the forces used in all of the loading conditions we used. Our model did not consider the cumulative effect of damage caused by sequential loading conditions. New routines including material fatigue, viscoelasticity, damage, etc., could be added to the present model. More specific studies measuring ground force reactions in calves getting up and lying down would be useful.

Polymeric materials were tested in this study because of their biocompatibility, light weight, low cost, and also because of the reduced stress shielding or stress protection atrophy that may occur using rigid metallic nails.[28] Polymeric nails for fracture treatments in cattle have other reported advantages, such as avoiding the use of high-cost image magnifiers (as holes can be drilled during surgery) and avoiding a second surgery to remove the implant.[4,5]

We concluded that the finite element model analyses demonstrated that none of the polymers (polyacetal, polyamide, or polypropylene) were sufficiently resistant to tolerate loading forces imposed on the femur during walking and that the screws closer to the fracture line are important stress areas. These results are in agreement with a previous *in vivo* study.[7] Loading conditions of this finite element analysis could be improved to better simulate forces and create a repertory of calf movements. This computational simulation model was useful in predicting immediate postoperative femoral loading conditions in calves subjected to surgical immobilization of a diaphyseal fracture and should be considered in the development of new devices.

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