Finite element analysis of stress in the equine proximal phalanx

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Summary

Reasons for performing study: To improve understanding of the internal structure of the proximal phalanx (P1), response of the bone to load and possible relation to the pathogenesis of fractures in P1.

Objectives: To model the P1 and replicate the loads experienced by the bone in stance, walk, trot and gallop using finite element analysis.

Methods: The geometry of the P1 was captured using micro-computed tomography (µCT) and was reconstructed in 3 dimensions. Values for material properties and forces experienced at stance, walk, trot and gallop were taken from the literature and were applied to the reconstructed model. Using the same load total across the proximal articular surface, the model was solved with and without loading of the sagittal groove. Biomechanical performance was then simulated with finite element analysis and evaluated in terms of von Mises stress maps.

Results: Compared with the lowest force simulation equivalent to stance, the effects of the gallop force showed higher levels of stress along the sagittal groove and on the palmar surface just distal to the sagittal groove in both models, with and without the sagittal groove loaded. The results highlighted an area of bone on the dorsal aspect of P1 that experiences lower stress compared with the rest of the dorsal surface, an effect that was much more apparent when the sagittal groove was not loaded. Qualitative comparison of the models revealed minimal difference in the pattern of von Mises stress between the loaded and unloaded groove models.

Conclusions: The study demonstrates a finite element model of P1 that produces results consistent with clinical observation. The simulated high-stress levels associated with the sagittal groove correspond to the most common site for fractures in the equine P1.

Potential relevance: With refinement of the model and further investigation, it may be possible to improve understanding of the behaviour of P1 under loading conditions that more closely simulate those experienced in the living animal, leading to a more solid understanding of fractures of P1.

Keywords: horse; proximal phalanx; fracture; finite element analysis; stress

Introduction

Injury in the equine athlete has a major impact on both the economics of the racing industry and welfare of the racehorse [1]. Musculoskeletal injury accounts for around 80% of all equine injuries [2] with fractures along the sagittal groove of the proximal phalanx (P1) accounting for 17% of total fatal limb fractures in racehorses over a 3-year period in the UK [3,4]. The most common configuration of P1 fractures in Thoroughbred racehorses is a sagittal fracture through the central groove [5,6]. Prognosis for these fractures varied from good to fatal with 32.8% of horses unable to race again, 8.4% being destroyed and 2% suffering another fracture at 3 years of age [7]. Clearly, a better appreciation of the biomechanics of the P1 bone would be of considerable benefit in improving our biomechanical understanding of these relatively common injuries.

Finite element analysis is a technique initially developed for the engineering industry to enable the prediction of stress and strain distributions across a complex geometrical object subjected to a load [8]. Stress is a measure of the force per unit area experienced in a body, compared with strain, which is a measure of deformation. Finite element analysis involves creating an accurate 3 dimensional (3D) model of an object, which is then assigned material properties, such as Young’s modulus, which is a measure of the elasticity of the material and Poisson’s ratio, which is the ratio of transverse to longitudinal strain. The model must also be constrained at a number of points to prevent it from moving in space when a load is applied. These properties and boundary conditions can then be used to simulate the behaviour of the model under certain loading conditions [8]. The results are frequently presented as a contour map of von Mises stresses, which give an indication of the likelihood of failure of the structure at any given point by calculating the differences between the 3 principal stresses at that point.

Finite element analysis is a recognised technique for studying bone biomechanics and fracture repair in man [9] and has been used previously to model the equine metacarpus [10]. However, owing to advances in computing power, the model of P1 presented here is an improvement on the metacarpal model in terms of resolution and accuracy of the bone geometry. More recently, finite element analysis has been used to study parts of the equine anatomy, such as the pulmonary artery [11], digit [12] and hoof [13,14].

The aim of this study was to build a finite element model of the equine P1 that was as biologically realistic as possible and to utilise the model to simulate loading conditions experienced by the proximal portion of the bone under a variety of gaits from stance to gallop. The aim is to gain insights into the response of the P1 bone to different loading conditions and thereby improve our understanding of the biology of this bone, particularly with respect to the biomechanical aetiology of acute and chronic fractures. The objective of this study was to create stress maps of the bone under different loading conditions, including a loaded and unloaded sagittal groove that could be analysed to determine changes in the pattern of stress with changes in load.

Materials and methods

One P1 bone from the forelimb of a fit Thoroughbred racehorse, which died from nonmusculoskeletal causes, was used. The first stage of creating the model was to capture the surface geometry and internal structures of the equine P1. This was done using micro-computed tomography (µCT) scanning on the X-Tek S2000 system housed in the Henry Moseley Imaging Facility, University of Manchester. Data were collected with 95 kV, 110 µA, and 1400 views, generating isotropic voxels of 0.057 mm. These voxels were subsequently down-sampled to isotropic voxels of 0.066 mm to reduce noise further and improve data handling in silico. The µCT images were imported into Amira 5.3.3 in which a 3D surface reconstruction of the bone was created. At this point, cortical and trabecular bone were differentiated.

The 3D surface was imported into HyperMesh 10.0 where the surface was converted to a 3D volume mesh consisting of over 6 million linear tetrahedral elements, representing 3 components: cortical bone, trabecular bone and a platform on which the bone sat. The bone marrow was left as a space for the purpose of this model. The platform was added in silico, after µCT scanning and was attached to the distal end of the bone (Fig 1). The 4 distal corners of the platform were constrained against
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Translations and rotations in all 3 dimensions (Fig 1). This prevented free body motion of the model without placing constraints directly on to the bone itself, which would have distorted the stress patterns around the distal end of the bone. The platform is not an attempt to replicate the proximal interphalangeal joint – it is simply a method that avoids over-constraining of the distal end of P1.

Material properties were assigned to each of the 3 model components based on previously published values. Cortical bone was assigned a Young’s modulus of 18,000 MPa and a Poisson’s ratio of 0.3 [15,16]. Trabecular bone was assigned a Young’s modulus of 1500 MPa and a Poisson’s ratio of 0.3 [17]. The distal platform was assigned the properties of Young’s modulus of 3 MPa and a Poisson’s ratio of 0.1 [18]. These properties, which most closely simulated cartilage, were assigned to the platform as they gave the most realistic results with the model compared with the properties of bone or concrete. The presence of the platform meant that it was not possible to assess the von Mises stresses experienced by the distal end of P1; however, this did not affect the results of this study, which focused on the proximal end of P1.

Loads were created by applying forces to a number of nodes on the proximal articulating surface of P1, representing the forces transmitted from the third metacarpal bone (McIII). The orientation of the force was 90° to the floor of the platform. The number and location of nodes was selected according to the increase of contact area between P1 and McIII from stance to gallop reported by Brama et al. [19]. Nodes were dispersed across the whole contact area, with each simulated increase in gait. The number of nodes, the force per node and total forces are summarised in Table 1.

Because the loads were applied to discrete points rather than surfaces, elements in the immediate vicinity of loaded nodes showed unrealistically high stresses in our models. As the sagittal groove on the proximal surface of the bone was of particular interest, the model was created with direct loads placed on the articulating surfaces including on the groove. Subsequently, the model was re-run with the direct loads on the proximal articulating surface but not in the sagittal groove, for comparison.

The models were solved using the computer software Abaqus 6.7.1®, producing colour maps of von Mises stresses across the bone (Fig 2). The maps show the range of stress experienced by the bone surface, as well as the areas of highest and lowest stress response of P1.

Results

The model shows that as the applied force increases, the stress experienced by the bone begins to centre on several areas, when compared with forces simulating stance (Fig 2). At stance, the stress experienced by the bone is minimal (Figs 2a,b,c). As load increases to walk, the increase in stress experienced by the bone model is located almost exclusively on the dorsal surface of P1. There is also an increase in stress along and distal to the transverse ridge on the palmar aspect of the sagittal groove, which is best observed in Figure 2e. A further increase in load to the level experienced at trot shows a more obvious increase in stress experienced by P1. In the dorsal view of trot, there is an increase in stress on both medial and lateral aspects at the proximal end of P1, in the metaphyseal region (Fig 2g). Along and distal to the transverse ridge on the palmar aspect of the groove, a much higher level of stress is present compared with the rest of the proximal articulating surface (Fig 2h). As the force is increased to simulate gallop, the stress experienced by P1 intensifies. The proximal end of the body of P1 experiences high levels of stress, particularly along the lateral aspect (Fig 2j). The degree of stress in the sagittal groove alters from relatively low at trot to very high at gallop. The palmar view of the model at gallop also shows that stress is beginning to propagate distally from the sagittal groove on the palmar aspect of the shaft (Fig 2k, bright green triangular area, highlighted within the red square). The von Mises stresses are high along the sagittal groove with a steady increase in the stress progressing towards the dorsal margin. There is a rim of high stress (red line) at the most palmar extent of the sagittal groove, the transverse ridge (Fig 2l). The model was solved again at loads simulating forces experienced at gallop; however, this time the nodes were placed to exclude loading of the sagittal ridge (Figs 2 m–o). In this scenario, one area of the dorsal aspect, near the proximal end of the bone, was shown to experience lower stress compared with the rest of the dorsal surface (darker blue) as seen in Figures 2m,n,o. This area was present in the model of gallop with the groove loaded, but to a lesser degree. Without loading the groove, the triangular area of increased stress and the rim of high stress remain on the palmar aspect of the sagittal groove with a focal area of high von Mises stress present at the dorsal most aspect of the sagittal groove (Figs 2m,n,o).

Discussion

This preliminary study has constructed a finite element model of P1, which focuses on the proximal part of P1 and may assist our understanding of pathology seen in P1 in the clinical setting. At gallop, the model showed an area of very low stress on the dorsoproximal aspect of P1. This area is present at gallop when the sagittal groove is both loaded and unloaded; however, to a lesser degree, when the sagittal groove is loaded. It is not entirely clear why this region experiences relatively low stresses in our model but it may be because, compared with the palmar surface, the dorsal cortex has slightly thicker cortical bone (Fig 3a). In addition, the region of low stress could be sitting above and in front of the intersections of stresses that descend from the proximal articular facets via the trabeculae towards the thick collar of cortical bone around the P1 shaft (Fig 3b). The directing of the stresses away from the dorsoproximal area would account for the lower stress that descend from the proximal articular facets via the trabeculae towards the thick collar of cortical bone around the P1 shaft (Fig 3b). The directing of the stresses away from the dorsoproximal area would account for the lower

<table>
<thead>
<tr>
<th>Gait</th>
<th>Total force (N)</th>
<th>Number of nodes</th>
<th>Force per node (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stance</td>
<td>1800</td>
<td>71</td>
<td>25.35</td>
</tr>
<tr>
<td>Walk</td>
<td>3600</td>
<td>115</td>
<td>31.30</td>
</tr>
<tr>
<td>Trot</td>
<td>5400</td>
<td>124</td>
<td>43.55</td>
</tr>
<tr>
<td>Gallop (groove loaded)</td>
<td>10,000</td>
<td>152</td>
<td>65.79</td>
</tr>
<tr>
<td>Gallop (groove not loaded)</td>
<td>10,000</td>
<td>100</td>
<td>100.00</td>
</tr>
</tbody>
</table>

Table 1: A summary of the forces used to simulate the various gaits and how they were applied to the model. The total force applied to the models of gallop (both loaded and unloaded groove) were the same, but the number of nodes varied and therefore the force per node varies.

Fig 1: Three-dimensional model of horse P1 bone comprising 6 million linear tetrahedral elements. a) Whole model including constrained nodes (red triangles) and loads (red arrows); b) Cross-section through model showing constraint platform (yellow), cortical bone (blue), trabecular bone (red) and marrow space.
Fig 2: von Mises stress patterns produced on the dorsal, palmar and proximal articulating surfaces of the proximal phalanx (P1) from the right forelimb of a horse, under conditions simulating stance, walk, trot and gallop. Column 1 = dorsal view, column 2 = palmar view, column 3 = proximal articulating surface. Stress patterns for stance (a,b,c), walk (d,e,f), trot (g,h,i), gallop (j,k,l) when the groove is loaded and gallop (m,n,o) when the groove is unloaded are shown. The points on the proximal articular surface surrounded by red, green and grey, represent the nodes at which the virtual force was applied. Lateral is on the left hand side for the dorsal and proximal articulating surface views. Lateral is on the right hand side for palmar views. The red square in k) represents the area of high stress on the palmar surface of P1.
The model shows that as load increases, the sagittal groove and medial and lateral metaphyses undergo the highest levels of stress. The increased stress at the palmar and dorsal edges of the sagittal groove is relevant as these areas of highest stress correlate with the common fracture configuration in the living horse [3–5]. It is currently not known if sagittal fractures originate on the palmar or dorsal aspect of the sagittal groove of P1; however, the model indicates that the entire sagittal groove experiences high stress when compared with the rest of the bone, with the exception of the lateral proximal metaphysis.

The palmar transverse ridge and the area immediately distal are consistently areas of high stress across all the models (Figs 2f, k, l, n, o). The loaded groove seems to only experience higher stress when at gallop and relatively little stress in stance, walk and trot, in a nonlinear pattern, which is not consistent with the increase in stress along the sagittal groove as described by Den Hartog et al. [21].

The comparison of the unloaded and loaded groove models at loads simulating gallop show that in both scenarios, the palmar aspect of P1 is a consistent area of high stress (Figs 2k, n). However, when the sagittal groove is loaded, the palmar aspect appears to experience lower stress than when the groove is unloaded, with the primary compressive load on the medial and lateral articular facets (Figs 2l, d).

Although finite element analysis as a technique for modelling has been widely used in engineering and human medicine, its use in the field of equine bone behaviour is relatively new. Finite element analysis modelling has proved a useful technique in bone biology, such as for the creation of orthodontic implants and the testing of function under load (17–23) and, as shown here, it is possible to recreate digitally a biological structure. However, there are a few limitations to the finite element methodology. In particular, forces and constraints can only be applied to individual nodes, as opposed to the entire joint surface as occurs in nature. This means that force is applied to one minute spot, which does not replicate in vivo biology. This problem was partially overcome in this analysis by using multiple nodes over which the force was spread. In addition, we created a distal constraint platform which, although not perfect, increases the biological accuracy of the model. This new approach allows biologically plausible constraints to be applied distally, without interfering with study of the behaviour of the proximal surface of the bone. The model appears to be a good indicator of the stresses that occur in P1 during stance, walk, trot and gallop. The finding of increased stress in the model corresponding to common areas of failure in P1 suggests that with further refinement the model could be used to investigate fracture biology in P1. In addition, the similarity between the finite element model and clinical findings encourages the use of the model to explore, in silico, strategies for fracture prevention that would be clinically relevant.

Further investigation and refinement of the model to include replication of more scenarios with respect to gait and force distribution and direction, will enhance the model’s applicability to the study of the proximal subchondral bone under the sagittal groove, and therefore, the study of P1 fractures.

Authors’ declaration of interests

No competing interests have been declared.

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Author contributions

N. Jeffrey – study design, data collection, provided equipment, manuscript preparation, data analysis. E.R. Singer – study design, data collection, result interpretation, manuscript preparation, data analysis. P.G. Cox – study design, data collection, interpretation results, manuscript preparation, operation of computer programs. L.M.S. O’Hare – manuscript preparation, execution of the study, study design.

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