The Effect of Saddle Design on Stresses in the Perineum during Cycling

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ABSTRACT

SPEARS, I. R., N. K. CUMMINS, Z. BRENCHLEY, C. DONOHUE, C. TURNBULL, S. BURTON, and G. A. MACHO. The Effect of Saddle Design on Stresses in the Perineum during Cycling. *Med. Sci. Sports Exerc.*, Vol. 35, No. 9, pp. 1620–1625, 2003. **Purpose:** Repetitive internal stress in the perineum has been associated with soft-tissue trauma in bicyclists. Using an engineering approach, the purpose of this study was to quantify the amount of compression exerted in the perineum for a range of saddle widths and orientations. **Methods:** Computer tomography was used to create a three-dimensional voxel-based finite element model of the right side of the male perineum-pelvis. For the creation of the saddle model, a commercially available saddle was digitized and the surface manipulated to represent a variety of saddle widths and orientations. The two models were merged, and a static downward load of 189 N was applied to the model at the region representing the sacroiliac joint. For validation purposes, external stresses along the perineum-saddle interface were compared with the results of pressure sensitive film. Good agreement was found for these external stresses. The saddles were then stretched and rotated, and the magnitude and location of maximum stresses within the perineum were both recorded. In all cases, the model of the pelvis-perineum was held in an upright position. **Results:** Stresses within the perineum were reduced when the saddle was sufficiently wide to support both ischial tuberosities. This supporting mechanism was best achieved when the saddle was at least two times wider than the bi-ischial width of the cyclist. Stresses in the anterior of the perineum were reduced when the saddle was tilted downward, whereas stresses in the posterior were reduced when the saddle was tilted upward. **Conclusions:** Recommendations that saddles should be sufficiently wide to support the ischial tuberosities appear to be well founded. Recommendations that saddles be tilted downward (i.e., nose down) are supported by the model, but with caution, given the limitations of the model. **Key Words:** SORENESS, IMPOTENCE, ARTERIES, NERVES, SOFT TISSUE MODELING.

The perineum, the region lying between the external genitals and anus, is not well designed to undergo compressive stress, and the link between bicycling and damage to the perineum is becoming cause for concern (4,9,13,15,26). In case studies, the problems have been resolved with a variety of solutions ranging from adjusting saddle orientation (3) to immediate cessation of cycling (20). Although these recommendations appear to provide short-term relief, they are seldom founded on scientific evidence. The exceptions are isolated studies, which have identified a potential relationship between cycling and trauma to the perineum by comparing indicative measures including penile brachial index (7) and transcutaneous penile oxygen pressure (10,21) in different cycling positions (e.g., standing/seated). Although there are anecdotal suggestions that seat padding may reduce perineal compression, saddle design is found to be more influential (17). Most importantly, a significant inverse relationship between saddle width and reduction in penile oxygen pressure has been reported (17). Unfortunately, the experimental set-up did not allow these researchers to isolate the results, and therefore they were unable to determine the amount and location of compression within the perineum. Hence, an alternative method was chosen for the present study.

Finite element analysis is a numerical method for solving problems of engineering. Its ability to simulate intricate geometry, complex loads, and inhomogeneous materials is exclusive (6) and allows examination of structures inaccessible by traditional experimental procedures (18). By creating a three-dimensional finite element model of the perineum, derived from computer tomography (CT) images, and of various saddle designs, this study aims to quantify the effect of (a) saddle orientation and (b) saddle width on the stresses in the perineum.

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MATERIAL AND METHODS

Creation of the models. Raw CT images (512×512 pixels, pixel size = 0.33 mm, slice thickness = 1 mm, slice increment = 1 mm) of the lower pelvic region (Fig. 1A) were downloaded from The National Library of Medicine Visible Human Project fresh male data set

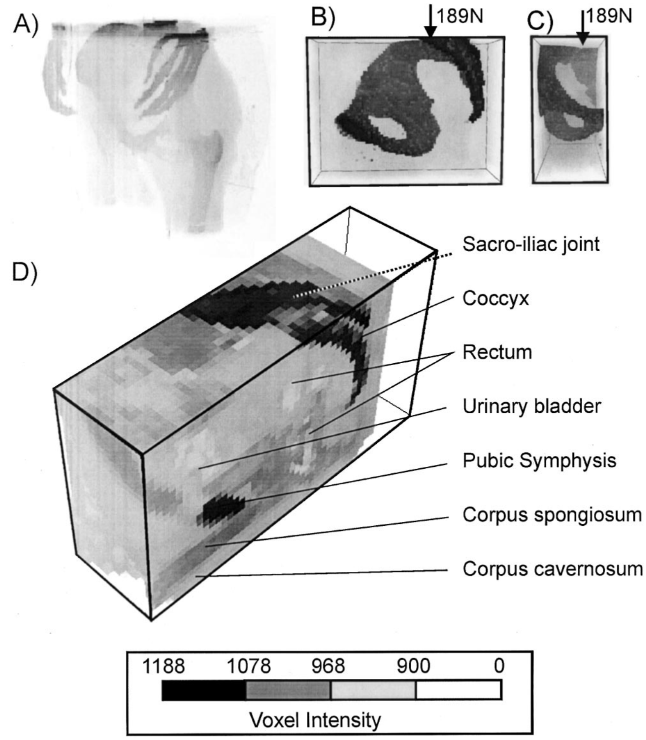

FIGURE 1—The procedure used to create the finite element model. Full-body CT slices were downloaded and merged to form a three-dimensional voxel model of the hip region (A). The right-side of the lower perineum between the midline of the body and the acetabulum was extracted to form the finite element model (viewed sagittally [B] and frontally [C]). The finite element model viewed isometrically (D) with landmarks highlighted for visualization purposes the soft tissues in A, B, and C were assigned low opacity (0.001). Also shown (B and C) is the force applied to the region of the model representing the sacroiliac joint. The cubes seen in B, C, and D are the elements of the model.

(http://www.nlm.nih.gov). To make a three-dimensional voxel representation, the region of the lower right-sided pelvis was extracted (Fig. 1B). Voxel-based finite element models have been shown to yield realistic results but require that the appropriate number of elements be used: a large number of elements results in greater accuracy but they increase solution time. After conducting a series of convergence tests to determine the most appropriate mesh size, a $30 \times 80 \times 58$ hexahedral eight-noded finite element model was used (Fig. 1). The model was divided into four material groups comprising bone, voids, fat, and flesh. The distinction between material properties was based on the local voxel intensity (VI) (i.e., Hounsfield numbers) acquired during CT scanning. Trabeculae and cortical bone (VI $\geq$ 1078) were grouped and assigned a homogeneous linear stiffness (E = 110 MPa). This value accords with previously validated models of the pelvis (22,23). Air and voids (VI $\leq$ 900) such as those found within the rectum were assigned a negligibly low stiffness (E = 1 Pa), thus ensuring negligible mechanical stress transfer. Threshold values for flesh (968 $<$ VI $\leq$ 1078) and fat (900 $<$ VI $\leq$ 968) were determined visually by varying the threshold levels until fat could be located anterior to the pubic symphysis (mons pubis) and muscle could be located superior and medial to the pubic symphysis (m. rectus abdominis). Due to its composition, soft tissue exhibits nonlinear mechanical behavior (1). This behavior is best represented using an incompressible Ogden material model (11) based on the following strain energy function (W):

$$W=\Sigma(\lambda_1{}^{\alpha} + \lambda_2{}^{\alpha} + \lambda_3{}^{\alpha} - 3)\ \mu/\alpha \text{ for } n = 1 \text{ to } N \qquad (1)$$

In this equation, λ_i are the extension ratios in the principal strain directions, and μ and α are the material parameters. The data assigned to this material model to represent skin (μ = 0.016 MPa, α = 10), fat (μ = 0.006 MPa, α = 5), and flesh (μ = 0.012 MPa, α = 30) have been used previously for similar purposes (12) whereby only the first order model (i.e., n = 1) is used. Although *in vivo* data on the behavior of fat elsewhere in the body exist (1), data on perineal fat do not. Fat behavior was varied between what was considered plausible values in accordance with a previous investigation (12). It is noteworthy that, in contrast to the earlier research (12), stresses were not sensitive to the fat properties, provided that the fat is less stiff than the other tissues. This was the case in the present study. It is difficult to distinguish between fleshy tissues in CT images and, hence, muscles and ligaments were grouped within the model. Also, although it is recognized that magnetic resonance imaging or dissection would allow differentiation of these tissues, a mesh capable of distinguishing between such intricate structures would be impractical given current processing capabilities. Fleshy tissues behave anisotropically (i.e., different in different directions) and are assigned transverse properties as found under compression (2).

A point cloud of the surface of a commercially available saddle (Sport Gents, Selle Royal, Vicenza, Italy) was captured using a three-dimensional general-purpose scanner (ModelMaker W70, 3 d Scanners, UK). Coordinates along these curves were then transferred into MSC.Mentat (MARC Analysis, Palo Alto, CA) and bound using a Coons surface (Fig. 2B). This surface was positioned adjacent to the inferior aspect of the finite element model of the perineum, and the rear of the saddle was aligned with the rear of the buttocks. Nodes on the inferior surface of the perineum model were then attached onto the surface of the saddle. In doing so, a conforming interface between the saddle and perineum was assumed before loading. Preliminary analyses indicate that the inclusion of a thin layer of clothing or saddle padding does not affect stress transfer across the interface (whereby compression-only is transferred).

To investigate the effect of saddle orientation, the saddles were rotated about a mediolateral axis at the midway point of the inferior pubic ramus (Fig. 2B). Upward, downward, and horizontal inclination of saddles are all recommended (27), but unfortunately no precise orientations are given. An exploratory pilot investigation found, however, that tilting the saddle downward by more than $-10°$ (with respect to the horizontal plane) causes slipping, whereas tilting the saddle upward by more than $+10°$ causes discomfort. The model is therefore tested within the range of $\pm 10°$ at $2°$ increments. The model does not take into account pelvic movements, to accommodate either the orientated saddle or

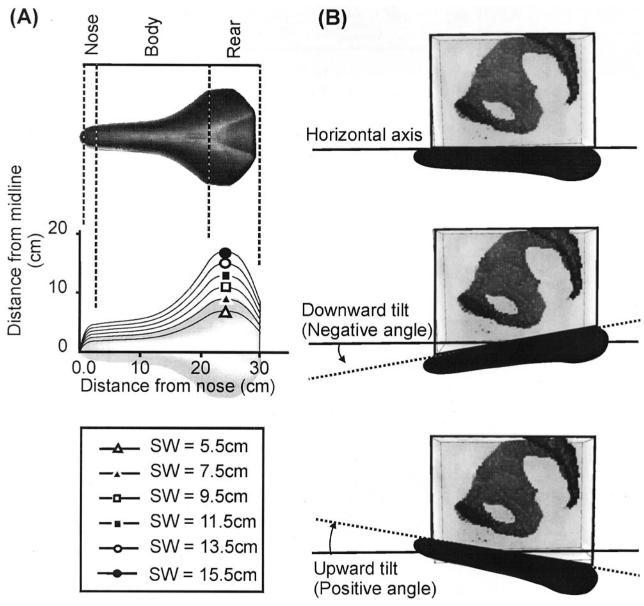

FIGURE 2—A. The terminology used to distinguish regions and the various saddle widths simulated (transverse plane). B. The finite element model of the perineum and saddle shown in the sagittal plane. Also shown are the negative and positive angulations applied to the saddle model, representing the downward and upward tilts, respectively.

leg movements during activity. It is therefore important to note that the pelvic model was positioned in an upright position and remains so throughout the analysis. Such a limitation will be discussed in light of the findings made.

To investigate the effects of saddle widths on internal stresses, the saddle model was compressed/stretched mediolaterally to create a range of saddle widths ($5.5 \leq SW \leq 13.5$ cm). To put such dimensions into perspective, the saddle model with a width of 7.5 cm is at the narrower end of the market and represents a common male racing saddle (e.g., Sport Gents). The saddles within the range 9.5 to 11.5 cm are representative of female saddles or those used by the general leisure bicyclists (e.g., Manhattan CX, Selle Royal). The wider saddles modeled ($13.5 \leq SW \leq 15.5$ cm) are representative of those used by the fitness industry on exercise bikes (e.g., Exercycle-Airdyne, Trico Sports, Pacoima, California Saddle).

Loading conditions. Dynamic forces exerted by the seat have been documented (24) for treadmill riding (gradient = 6%, pedalling rate = 84 RPM, and speed = 7.2 $m \cdot s^{-1}$). The force data acquired from rider 4 (body weight 778 N, age 35) in the aforementioned study (24) were chosen for the model. Maximum force values (sFz is approximately equal to −350 N, and sFx is approximately equal to −150 N) were resolved vertically (seat post angle = 73.5°) to give a maximum vertical force of −378 N. Accordingly, a static downward point load of 189 N (half of this load) was applied to the half-body model (Fig. 1, B and C). This force was applied to the most medially situated node at the region of the model representing the sacroiliac joint (Fig. 1). The decision to simulate such forces using a point load, as opposed to more equally spread surface loads, was warranted given that the stresses within the sacroiliac joint were not of interest. Viscoelasticity of the soft tissue (28) and seat padding, which is likely to play a role in dissipating shock loads during rough terrain cycling (8,4), was ignored in the model. Thus, the simulation represented the mechanical situation occurring during slow rising and prolonged body weight loads only. To account for the symmetry in the sagittal plane, nodes along the midline of the perineum were assigned zero displacement in the mediolateral direction. The nodes conforming to the surface of the saddle were assigned zero-displacement in the inferior-superior direction. In doing so, the nodes on the inferior aspect of the perineum-pelvis model were able to slide over the surface of the saddle, thus simulating a frictionless interface between perineum and saddle. A consequence of this approach is that the simulated saddle was perfectly rigid, whereas in reality saddles usually have a thin layer of padding at the surface. However, although possibly affecting the transfer of shock waves to the perineum, padding on the saddle is not expected to affect the static internal stresses (12). All calculations were carried out using MSC.Marc (MARC Analysis), and in total 66 simulations were performed. Given that compression has previously been described as the major culprit for soft-tissue trauma during cycling (17) and that the loads were inferior-superior in direction, compressive stress in the inferior-superior direction is reported here.

Validation. In an attempt to validate the model, the calculated stresses ($\alpha = 0°$, SW = 7.5 cm) were compared with *in vitro* measurements at the perineum-saddle interface. A male recreational cyclist (body weight = 934 N, age = 33 yr) with no reported history of perineal trauma was chosen and the model scaled according to his pelvic dimensions. He was asked to remove all clothing and lower himself (with assistance) gently onto a horizontal saddle. Although it was anticipated that clothing does not affect the general level of stress, the reason for performing the task naked was to avoid highly localized stress as a result of seams or buttons on the clothing. The subject adopted a static and upright riding position for 10 s with feet and hands in load-free contact with the pedals and handlebars. To represent this situation, as opposed to the more realistic but lower loads occurring during normal bicycling (24), a downward load of 467 N (i.e., half body weight) was applied to the model at the sacroiliac joint. To locate regions of maximum pressure, highly sensitive film (Microfilm, SensorProducts, East Hanover, NJ) with a pressure range of 0.00–0.14 MPa was placed on the saddle. This region was found to be directly below the ischial tuberosities. Then a 2 × 2 cm strip of ultra-low film (pressure range = 0.19–0.57 MPa) was placed in this region of high pressure and the experiment repeated. The exposed film was then captured digitally and the mean pixel intensity calculated and compared with coloration charts (room temperature = 25°C, relative humidity 60%) provided by the film manufacturer. The comparison of pressures recorded by the film (0.296 MPa) and the calculation within the same region of the model (mean = 0.294 MPa, standard deviation = 0.180) was excellent and further analysis is thus warranted. However, it should be borne in

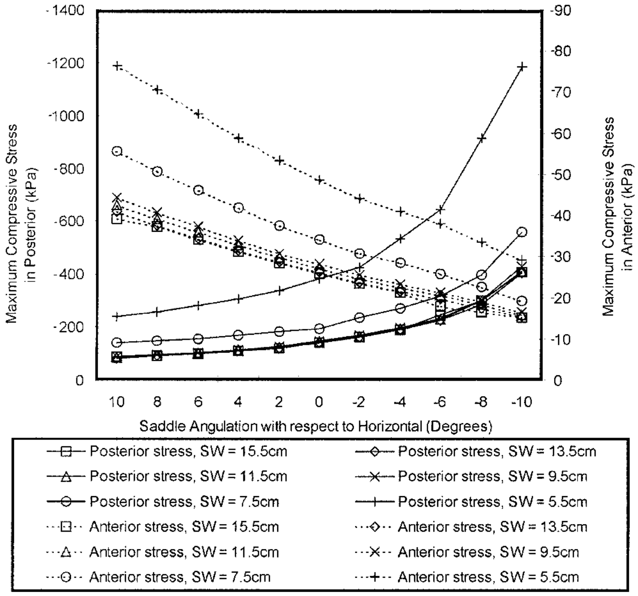

FIGURE 3—The effect of saddle width and orientation on calculated maximum compressive stress in the superior-inferior direction for the anterior and posterior of the perineum.

mind that in terms of assessing the potential for pudendal trauma, the saddle-perineum interface is not the major region of interest. The internal stresses occurring within the perineum are likely to be more damaging, and the calculations of these stresses made by the model cannot be validated directly.

RESULTS

For all saddle widths and orientations modeled, the greatest compressive stresses were within the pubic arch anteriorly and below the ischial tuberosity posteriorly (Fig. 3). The posterior stresses were greater than the anterior stresses. Changing the saddle orientation with respect to the horizontal plane causes stress to be redistributed within the perineum. Notably, tilting the saddle downward (i.e., negative orientation) caused anterior stress to decrease and posterior stress to increase. In contrast, tilting the saddle upward caused anterior stress to increase and posterior stress to decrease. For example, tilting the 7.5 cm-wide saddle downward by $-10°$ (anterior stress $= -19.1$ kPa) caused a 44% reduction in anterior stress when compared with the same saddle in the horizontal position (anterior stress $= -34.2$ kPa). Similarly, tilting the saddle upward by $+10°$ (anterior stress $= -55.6$ kPa) caused a 62% increase in anterior stress. In comparison with the horizontal saddle (posterior stress $= -192.2$ kPa), tilting the saddle downward by $-10°$ (posterior stress $= -558$ kPa) caused a 190% increase in posterior stress, whereas tilting the saddle upward by $+10°$ (posterior stress $= -142$ kPa) caused only a 26% reduction in posterior stress. Increasing saddle width decreased the overall levels of stress within the perineum. However, increasing saddle width for the narrow saddles (i.e., from 5.5 to 7.5 cm) caused a large decrease in overall stress, whereas for the wider saddles (i.e., 9.5 to 15.5 cm) the effect was less pronounced.

DISCUSSION

Although it is well documented that bicyclists are prone to perineal damage (4), the effects of saddle design and orientation on trauma remain largely unexplored. To contribute to solving this problem, the present finite element study was designed not only to demonstrate how stresses in the perineum are distributed during bicycling but also to determine the effects of saddle width and orientation on their magnitude.

It is particularly noteworthy that the calculations of internal stresses in the anterior of the perineum reached their peak within the pubic arch, and these stresses were consistently over 15 kPa. To put such values into perspective, these values were comparable with, or even greater than, the systolic blood pressure (120 mm Hg = 17 kPa) within the aorta (25). It is also noteworthy that the direction of these stresses was superior-inferior and therefore perpendicular to the anterior-posterior orientated pudendal nerves and arteries. On the basis of mechanical theory alone, such stresses were sufficient to cause complete cessation of blood supply to the external genitals. However, because this is unlikely (17,21), it is suggested the values obtained in the analyses may be too high due to simplifications made in the model. As in any finite element study, these simplifications are with regard to loading conditions, geometry and material properties assigned.

When assigning loading to the perineum-pelvis model, it was assumed that the loads on the perineum are slow rising, static, and can be adequately represented by maximal seat-post forces (24). Yet, due to muscle activity and shifts in body weight, seat-post forces (and therefore the loads transferred across the perineum) are cyclical (24). Such repetitive cyclical loads when acting on soft tissue may actually stimulate blood flow as a result of increased muscle activity (also within the cell walls of the arteries). Hence, although the pudendal arteries may become obstructed during peak loads, these periods are relatively short and may be compensated for by increased blood flow before and after this event. Sufficient blood supply to the genitals could thus be maintained. Nevertheless, although cyclical in nature, the forces transmitted on to the saddle (24) are substantial even during periods of minimum force (i.e., between 230 and 240 N for rider 4). Thus, even if our findings are extrapolated for these lower loads, the stresses predicted would still be surprisingly high.

With regard to the material properties, it was assumed that during cycling the fleshy tissues within the perineum, i.e., ligaments and muscles, are in a passive state and compressed transversely with respect to their orientation. In doing so, the effects of high-stiffness tensile behavior of ligaments and increased muscle stiffness during activity were both ignored. With regard to muscle activity, it is envisaged that an avid bicyclist may isometrically (possibly inadvertently) contracts the transversely orientated muscles of the perineum (e.g. m. levator ani) to protect the perineum against superiorly directed saddle reaction forces. As regards the ligaments, they are considerably stiffer in tension than in compression (14). Although compression is the predominant loading regime exerted on the perineum, superior-directed loads on the transversely orientated ligaments (e.g., transverse perineal ligament) will cause some upward deformation of the ligaments, which in turn could place the ligaments under transverse tension. The high stiffness could then act to redistribute loads onto the pelvis. This is likely to be the case, and, if muscle activity and tension within the ligaments could be modeled with greater accuracy, stress transfer away from the soft tissue of the perineum is likely to occur. On the basis of these considerations, it is acknowledged that the quantitative nature of the results should be interpreted with caution. However, the results are internally consistent such that the limitations are identical regardless of the saddle width or orientation being modeled. This, together with the fact that similarities exist between the present study and *in vivo* experiments allow conclusions to be drawn and recommendations to be made.

With regards to the orientation of saddles, downward tilt of the nose was found to reduce stress in the anterior while simultaneously increasing stress in the posterior of the perineum. Such findings can be explained in terms of biomechanical parameters. As body weight is transferred from the pelvis to the saddle, the soft tissue lying between these structures becomes impinged. In the posterior region, this impingement takes place between the ischium and the body of the saddle, whereas in the anterior region it occurs between the pubis and the nose of the saddle. Given that the pelvis is assumed to remain stationary and upright, the ischium is closer to the horizontal saddle than is the pubis. Consequently, body weight loads, by taking the easiest route from the pelvis to the saddle, tend to concentrate in the posterior of the perineum. Tilting the nose of the saddle downward, as recommended by various authors (5,13,8,26), effectively moves the saddle closer to the ischium and further from the pubis, thus promoting the transfer of stress across the posterior perineum. In contrast, tilting the nose of the saddle upward causes the saddle to move closer to the stationary pubis, which in turn promotes the transfer of stress across the anterior region of the perineum. As to whether anterior or posterior stress is more damaging is arguable. On the one hand, the region below the ischium could be considered important if reduction of posterior pressure sores is an issue (12). On the other hand, given the location and orientation of the pathway of the pudendal nerves and arteries, peak stresses underneath the pubic arch may be of greater concern. For cases of pudendal trauma, previous recommendations to tilt the nose of the saddle downward (16) appear to be well founded. However, it needs to be borne in mind that this support is based on a model in which the pelvis is always upright. In reality, the rider alters posture for a variety of reasons, for example, when adopting a more aerodynamic position (19), and may do so in order to accommodate a change in saddle orientation. Such shifts are not only likely to change the relative angle between the pelvis and saddle, but they are also likely to change the amount of load transferred on to the saddle. Further work is clearly needed in this area. However, although suggestions with regard to changing saddle orientation remain cautious, suggestions with regard to changes in saddle width can be made with greater confidence.

In a physiological experiment, a positive relationship was found between penile transcutaneous oxygen pressures and saddle width (21). The findings of the present study lend support to suggestions that wider saddles reduce compressive stress and are therefore likely to relieve the trauma in the perineum. In addition, the model demonstrates that the relative benefits of increasing saddle width diminish (quite rapidly) for the wider saddles. Such findings may be important given that wide saddles are not popular among cyclists. Taking such factors into consideration, it is suggested that there may be a “trade-off” width whereby the saddle is sufficiently wide to support the ischial tuberosities but also sufficiently narrow so as not to restrict leg movements. Although no attempt was made to simulate leg movements, it is suggested that the most likely saddle to achieve this in the present study was the model with a width of 9.5 cm. Any further reduction in width would cause a large increase in stress in the perineum, whereas any increase in width further would only result in a slight reduction in stress, probably at the expense of leg movements. The “trade-off” width is linearly related to the dimensions of the pelvic bone, in particular the ischial width (i.e., the distance between the ischium and the midline of the body). In the present model, this was between 5 and 6 cm. Taken as such and extrapolated from the present results, it is suggested that cyclists should choose a saddle approximately twice the width of their ischial tuberosities. This, together with a cautious recommendation of tilting the saddle downward, could contribute toward reducing the occurrence of pudendal trauma among cyclists.

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