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Football practice and urinary incontinence: Relation between morphology, function and biomechanics

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ABSTRACT

Current evidence points to a high prevalence of urinary incontinence among female athletes. In this context, this study aims to assess if structural and biomechanical characteristics of the pubovisceral muscles may lead to urine leakage. Clinical and demographic data were collected, as well as pelvic Magnetic Resonance Imaging. Furthermore, computational models were built to verify if they were able to reproduce similar biomechanical muscle response as the one measured by dynamic imaging during active contraction by means of the percent error. Compared to the continent ones ($n=7$), incontinent athletes ($n=5$) evidenced thicker pubovisceral muscles at the level of the midvagina ($p=0.019$ and $p=0.028$ for the right and left sides, respectively). However, there were no differences neither in the strength of contraction in the Oxford Scale or in the displacement of the pelvic floor muscles during simulation of voluntary contraction, which suggests that urine leakage may be related with alterations in the intrafusal fibers than just the result of thicker muscles. Additionally, we found similar values of displacement retrieved from dynamic images and numerical models (6.42 ± 0.36 mm vs. 6.10 ± 0.47 mm; $p=0.130$), with a percent error ranging from 1.47% to 17.20%. However, further refinements in the mechanical properties of the striated skeletal fibers of the pelvic floor muscles and the inclusion of pelvic organs, fascia and ligaments would reproduce more realistically the pelvic cavity.

1. Introduction

Urinary incontinence (UI) is traditionally regarded in the literature as a problem affecting older multiparous women due to its relation with ageing, pregnancy and vaginal delivery (Martins et al., 2010). However, a high prevalence of UI, mainly stress UI (SUI) (Da Roza et al., 2014; Eliasson et al., 2002), has been observed in young nulliparous female athletes (Bø, 2004; Eliasson et al., 2008), which can be explained by the marked increase in intra-abdominal pressure (IAP) (Bø et al., 1994) during exercise. Accordingly, as a consequence of high and recurrent pressure upon the pelvic floor muscles, bladder and urethra, SUI is prevalent in high-impact activities such as trampoline, gymnastics and some ball games (Bø, 2004). On one hand, the repeated increase in IAP may lead to a continuous pre-contraction and strengthening of the pelvic floor muscles during training (Bø et al., 1994). However, there is enough evidence to state that intense training may also have the effect of overloading and stretching the pelvic floor (Bø, 2004; Ree et al., 2007), which potentially explains the development of structural and/or functional changes that can lead to SUI (Bø, 2004).

During the increase in IAP, the pubovisceral muscle closes the *urogenital hiatus* through a strong postero-anterior contraction to maintain the continence and to resist the downward movement of the organs.Assuming the hypothesis that women who undertake high-impact sports for a long period overload their pelvic floor (Bø, 2004), it is expected that these forces cause some degree of muscle damage or weakening, and consequently will change their biomechanical response to maintain continence and organ support (Dietz and Shek, 2008) in the long-term.

As in vivo experimental work is very difficult to undertake in this field, biomechanical models have been applied to better understand the role and the mechanical behavior of the pelvic structures in the development of UI and genital prolapse, such as the muscles and ligaments (Brandao et al., 2015; da Silva-Filho et al., 2010). As for example, Chen et al. (2006) evaluated the changes in material properties of the pelvic floor muscles, uterine ligaments and fascia

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to explain cystocele formation and apical descent, which are possible findings in pelvic floor dysfunction (Chen et al., 2009; Chen et al., 2006).

SUI is not fully understood (Bø, 2004), and diagnosis often relies on imaging methods such as static and dynamic ultrasound and Magnetic Resonance Imaging (MRI) that evaluate muscle damage or (ab)normal contraction (Avery et al., 2004; Chen et al., 2009). As our purpose was to gather knowledge of the anatomy and function of the pelvic floor muscles in female football players, in both a clinical and biomechanical perspectives, we aimed to assess if numerical models were able to reproduce similar muscle biomechanical response. To this end, and in order to investigate if there were differences between continent and incontinent athletes, we analyzed the morphology of the pubovisceral muscle and produced numerical models from their MRI data. Furthermore, we simulated pelvic floor muscles contraction to compare postero-anterior muscle displacement with the results from the dynamic sagittal images. This biomechanical analysis is part of current work to further validate the model, which will be used to assess the effect of IAP loading on the pelvic floor muscles induced by different types of exercise. This will enable to estimate local stress values along the pelvic cavity.

2. Materials and methods

2.1. Participants

This was a cross-sectional pilot study conducted between November 2012 and March 2013 in the region of Porto, Portugal. Inclusion criteria included: to be a federated athlete for at least 7 years, eumenorrheic, nulliparous, to be between 18 and 25 years old and to have a body mass index (BMI) between 19 and 30 kg/m^2. Women were ineligible if they had undergone pelvic surgery or previous pregnancy, and also if they had the diagnosis of pelvic organ prolapse or chronic constipation. The Research Ethics Committee of the Centro Hospitalar de São João – EPE, Porto, Portugal (code: CES10/2010) gave full approval to this study.

Women took the International Consultation on Incontinence Questionnaire-Short Form survey (ICIQ-SF). The ICIQ-SF (Avery et al., 2004) is a simple and brief questionnaire developed by the International Continence Society (ICS), translated and validated for the Portuguese language (Tamanini et al., 2004). The first question in the survey determines the presence or absence of UI, and it also includes eight questions to determine the type of UI (urgency, stress or mixed UI). Additionally, women were asked if the urine leakage episodes begun before or after starting their training, and also if they were able to stop the urine flow during micturation. Further demographic characteristics were collected (age, BMI, age of menarche and years of sport practice).

A final convenience sample of 5 incontinent (IG, incontinent group) and 7 continent (CG, continent group) athletes was selected from football Clubs.

2.2. Clinical evaluation

The 12 participants underwent clinical examination with the same gynecologist. During the evaluation, the contraction of the pelvic floor muscles contraction was assessed by digital examination of the vagina using the five-point Oxford Grading Scale: 0, no contraction; 1, flicker; 2, weak; 3, moderate; 4, good; and 5, strong. This was performed in a crook lying position, using two distal phalanges inside the vaginal *introitus*. Participants were asked to contract with the maximum perceived effort and to hold the contraction at least 2–3 s. Three consecutive squeezes were recorded with a 10-s interval between efforts (Bø and Finckenhagen, 2001), and the best was registered. Co-contraction of the gluteal, hip adductor or rectus abdominal muscles was discouraged through previous instruction to ensure a valid measurement. Only contractions with simultaneous observable inward movement of the perineum were considered valid (Bø et al., 1990).

2.3. Image analysis

The participants were scanned for pelvic MRI in the supine position. Static multiplanar T2-w high-resolution and sagittal dynamic images at maximal voluntary contraction were acquired using a 3.0 T scanner. An experienced radiologist reviewed the datasets, and no pathological findings were described.

Fig. 1 shows sagittal images from a continent 19-year old athlete, acquired at rest (a) and during contraction (b) of the pelvic floor muscles. The black lines are horizontal and vertical axes placed in the inferior and posterior border of the *symphysis pubis*. The white line demonstrates the distance between the vertical axis and the pubovisceral muscle. The difference in muscle position between rest and contraction was registered as being its postero-anterior displacement. This pulls the rectum, the urethra and the bladder anteriorly towards the *symphysis pubis* as seen in Fig. 1, which reduces the *urogenital hiatus* length and width.

Axial images were used to measure the pubovisceral muscle thickness at the level of the midvagina and canal anal (right and left sides) (Fig. 2a and d). To carry out the measurements, the slice closer to the plane of minimal hiatal dimensions was used. This plane is defined as the minimal antero-posterior (AP) diameter of the *urogenital hiatus*, from the postero-inferior margin of the *symphysis pubis* to the anterior margin of the pubovisceral muscle (DeLancey, 1994), where it defines the anorectal angle (Gregory et al., 2011).

2.4. Numerical simulation

To build the biomechanical models based on the finite element method, the pubovisceral muscle was identified between the pelvic organs and the obturator muscles throughout the axial images. It was manually segmented (Fig. 2b and e) by using a contour spline in the Inventor® software (Autodesk, San Raphael, CA, USA) (Brandao et al., 2013) to create a 3D solid model, see Fig. 2c and f. After this manual procedure, the geometrical model of the muscle and the finite element mesh (with tetrahedral elements – C3D4) were created using the Abaqus® software version 6.12 (Dassault Systèmes Simulia Corp., Providence, RI, USA). The number and volume of elements in the meshes varied among the subjects due to the individual differences in muscle geometry. Mean values were 121.113 and 19.115.80 mm^3, ranging from 61.768 to 232.930 and 9.324.46 to 29.934.68 mm^3, respectively.

Boundary conditions play a central role in the simulation results, because they may change the biomechanical response and the displacement values depending on the anatomical location of muscle attachments. Therefore, to simulate the pubovisceral muscle contraction, boundary conditions were imposed to incorporate the existence of the surrounding structures. The nodes corresponding to the insertion of the pubovisceral muscle in the *symphysis pubis* were considered zero-displacement nodes. Those portions attached to the *arcus tendineous levator ani* in the medial surface of the internal obturator muscle (which prevent lateral movement in the *x*-direction) were considered zero-displacement, and the tip of the coccyx (which is known to have some degree of vertical mobility) was allowed to move in the *z*-axis.

In order to obtain an approximation of the muscle fiber directions, which is an important parameter for the constitutive model, it was assumed that muscles have an isotropic behavior, defined by Yeoh (1993). It was also implicit that the direction of the muscle fibers is coincident with the direction of the maximal principal stress lines when the pelvic floor is deformed by the weight from the pelvic organs at rest. For this purpose, a uniform pressure of 1×10^{-3} MPa was applied to the inner surface of the pubovisceral muscle. The parameters C10 = 0.003 N/mm^2, C20 = 0.002 N/mm^2 and C30 = 0.001 N/mm^2 were obtained from the experimental work of Janda (2006).

In this work it is assumed that the muscles are a highly hydrated tissue and, consequently, a quasi-incompressible behavior was attributed. The constitutive equation adopted for the muscle 3D behavior is a modified form of the incompressible, transversely isotropic, hyperelastic model proposed by Humphrey and Yin (1987) for cardiac tissues. The strain energy per unit volume of the reference configuration can be written using the following equation:

$$U = U_I\left(\bar{I}_1^C\right) + U_J(J) + U_f\left(\bar{\lambda}_f, \alpha\right) \tag{1}$$

where U_I is the strain energy stored in the isotropic matrix embedding the muscle fibers, defined as

$$U_I = c\left[e^{b\left(\bar{I}_1^C - 3\right)} - 1\right], \tag{2}$$

where $c = 0.0185$ N/mm^2; $b = 1.173$.

In Eq. (1) U_J is responsible for ensuring the incompressibility condition (where $D = 0.0001$ mm^2/N), and is defined as

$$U_J = \frac{1}{D}(J-1)^2, \tag{3}$$

U_f is the strain energy stored in each muscle fiber, that includes a passive elastic part and an active part due to the contraction. The parameters used in the model for passive elastic behavior U_{pas}, given by the following equation were $A = 0.028$ N/mm^2; $a = 0.0625$

$$U_{pas} = A\left\{\exp\left[a\left(\bar{\lambda}_f - 3\right)^2\right] - 1\right\} \tag{4}$$

where $\bar{\lambda}_f$ represents the fiber stretch ratio in the direction N of the undeformed fiber, considering the isochoric part of the deformation gradient. When $\bar{\lambda}_f > 1$, otherwise we consider the strain energy to be zero, assuming that the fibers offer no resistance to compression. The active part U_{act} is given by (d'Aulignac et al.,