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Effects of fetal head shape variation on the second stage of labour

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ABSTRACT

Fetal head geometry plays an important role in the mechanics of childbirth during the second stage labour. Large heads have been shown to be associated with difficult and prolonged childbirth. However, the relationship between the fetal head geometry and childbirth mechanics has not been quantitatively analysed. To address this, our study used finite element (FE) modelling techniques and biomechanical simulations to analyse the contribution of fetal head shape and size on the mechanics of childbirth. X-ray computed tomography (CT) images from 26 newborn infants (less than 9 days old) without skull abnormalities were used to construct individual-specific FE models of the fetal skull. Simulations of childbirth were conducted using each model of the skull and a customised pelvic floor model based on magnetic resonance imaging (MRI) of a healthy nulliparous woman. The force required for delivery, the maximum principal stresses, and the maximum principal stretch ratios at the left and right pelvic floor muscle-pubic bone interfaces were quantified. Partial least squares regression (PLSR) models for predicting these mechanical indices were constructed using: (i) either the FE geometries of the fetal heads or the biometrical parameters (biparietal diameters and fetal head circumferences) as inputs; and (ii) either a linear or a quadratic function for the inner relation. The predictabilities of the mechanical indices using the PLSR models were quantified using a leave-one-out analysis. Quantitative associations were found between the geometric parameters of the fetal head and the indices of childbirth mechanics. When using the full FE geometries as inputs, the PLSR model using a linear inner relation gave better predictability than the model using a quadratic inner relation. This could be attributed to the quadratic inner relation correlating response to the noise in point-to-point correspondence. When using the biometrical parameters of the skull as inputs, the PLSR model using a quadratic inner relation gave the best overall predictability. Such a model could be implemented in a clinical setting as a predictive model for childbirth planning and as an educational tool for clinical training.

1. Introduction

The levator ani (LA) muscles, subdivided into the iliococcygues, pubococcygeus, and puborectalis, form a funnel-shaped muscular sheet that partially seals the pelvic floor. They originate along a semi-circular path on the pubic sidewalls from the ischial spines and insert into the lateral aspect of the pubic symphysis, forming a sling around the urethra, vagina, and rectum (Fritsch, 2006). The LA muscles regulate the intra-abdominal pressure, help with pelvic organ support, and maintain urinary and faecal continence. They are also intimately involved in the birth process in women (DeLancey, 2002). Childbirth-induced muscle injury is one of the leading factors contributing to pelvic floor disorders, including stress urinary incontinence and pelvic organ prolapse (DeLancey et al., 2003).

During a vaginal delivery, the LA muscles initially resist descent of the fetal head and then undergo extreme stretch to allow passage of the fetal head through the birth canal. If the stresses are too high, LA muscle damage may result (Baessler and Schuessler, 2004). Several imaging studies have demonstrated that the damage, frequently referred to as an avulsion injury, occurs at the anterior interfaces between the pubic bones and the LA muscles. It manifests as a complete or partial detachment of the LA muscles from the lateral pubic bones and occurs in 10% to 30% of vaginal deliveries (Dietz et al., 2007; Dietz and Lanzarone, 2005; Kearney et al., 2006). A previous study has shown that 53% of women with a complete avulsion experienced significant pelvic organ prolapse (Dietz and Simpson, 2008). Managing childbirth-induced injuries is an important preventive strategy as surgical correction is not always effective (DeLancey, 2005).

Extensive research has been conducted to identify women most as risk based on their pelvic floor muscle compliance and morphology (Kruger et al., 2008b, 2005). The shape and size of the fetal head also play an important role in childbirth-induced

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injuries (Baessler and Schuessler, 2004; Kearney et al., 2006; Valsky et al., 2009). Previous research has used a binary logistic regression model to predict the occurrence of LA avulsions during childbirth, using head circumference, birth weight, and duration of labour as risk factors (Valsky et al., 2009). However, with this methodology, predicted outcomes are based on statistical associations between the postnatal variables and the likelihood of muscle damage. To date, there has not been a predictive model quantifying intra-partum (during delivery) obstetric risks using the shape and size of the fetal head. This is partly because the inaccessibility of the LA muscles makes it difficult to measure the strains and stresses experienced by the maternal pelvic floor and the fetus.

With the aid of modern imaging modalities, including magnetic resonance imaging (MRI) and X-ray computed tomography (CT), it is possible to construct detailed computer models of the maternal pelvic floor and the fetal head to conduct mechanical simulations. Quantitative measures, including reaction forces from the pelvic floor and LA muscle stresses and stretch ratios, can be estimated from the individualised childbirth simulations (Hoyte et al., 2008; Li et al., 2010; Lien et al., 2004; Parente et al., 2008). However, finite element (FE) modelling of childbirth is typically computationally expensive, time consuming, and in general not feasible to use in a clinical setting. A statistical model that is able to rapidly predict important indices of childbirth mechanics is required for real-time evaluation of intra-partum obstetric risks.

This study conducted a series of FE childbirth simulations, using computer models based on a population of fetal head geometries. A partial least squares regression (PLSR) model was created using the FE descriptions of the fetal heads and simulation results for predicting the forces required for delivery, the maximum principal stresses, and maximum principal stretch ratios at the muscle–bone interfaces. Considering the difficulty in obtaining a full FE description of the fetal head in a clinical setting, a predictive model was also constructed using only parameters routinely measured antenatally, including biparietal diameters and fetal head circumferences.

2. Methods

2.1. Finite element models of the newborn skull

2.1.1. Imaging protocol and data segmentation

CT images of 26 newborn infants without skull abnormalities were used to approximate the shape of the fetal head. The subjects had a range of gestational ages and the images were acquired within 9 days of birth. Cranial growth was assumed to be negligible and the sizes of the newborn skulls were assumed to resemble those at birth. The effect of fetal head moulding was not considered in this study because the degree of moulding could not be individually quantified for the newborn subjects due to the lack of antenatal data. The demographics of the infants and the image resolutions are provided in Table 1. Ethical approval for using the images was obtained from the Auckland District Health Board (project number: A+5899, 13/NTA/65).

The bony plates of the skull were semi-automatically segmented from angled axial CT images using a Laplacian edge detection algorithm (Najarian and Splinter, 2012). The sutures and fontanelles were of similar signal intensity to the rest of the soft tissue structures (e.g. brain) and were therefore manually identified as gaps between the bony plates (Fig. 1a). The segmented data from all images were combined to form a cloud of points representing the skull geometry for each newborn subject (Fig. 1b). In this study, only the cranial bones that come into direct contact with the pelvic floor muscles were included in the biomechanical model. Facial bones and the mandible were excluded. Measurements of the biparietal diameters and the head circumferences were estimated from the CT images for all subjects (Table 2). The statistics for these biometrical measurements were consistent with the normal range described in the literature (Snijders and Nicolaides, 1994).

Table 1
Demographics of the infants and the CT image resolutions.

	Gestational age (weeks)	Age (days)	Pixel size (mm)	Slice thickness (mm)
Mean	39	4	0.2848	3.8
Range	32 to 40	0 to 9	0.1895 to 0.3957	1.0 to 6.0

2.1.2. Skull model construction

FE models of the newborn skull were constructed based on a generic fetal head model created in a previous study (Li et al., 2010). All 26 clouds of points were aligned to the surface data of the generic model in order for the newborn skull models to be in a consistent position and orientation required for the PLSR analysis. For each skull, the parameters of an orthotropic scaling transformation were determined to best fit the surface of the generic mesh to the surface of the individual skull. Each orthotropically scaled generic model was used as an initial mesh for a nonlinear geometric fitting procedure to generate an individual-specific skull model.

All 26 individual-specific models were constructed based on this global transformation of the same generic mesh, in an attempt to achieve material point correspondence. FE mesh fitting was performed using CMISS (www.cmiss.org). The fitted bicubic Hermite-linear Lagrange models preserved G^1 (tangent) continuity and consisted of 1344 geometric degrees of freedom (Fig. 1b). Statistics for the goodness-of-fit (root-mean-squared error between the clouds of points and the fitted surfaces, Eq. (1)) are provided in Table 2.

$$\mathrm{RMSE} = \sqrt{\frac{1}{n}\sum_{i=1}^{n}\left|\hat{\mathbf{x}}_i - \mathbf{x}_i\right|^2} \tag{1}$$

where $\mathbf{x}_i$ are the coordinates of the i^{th} surface data point, $\hat{\mathbf{x}}_i$ are the coordinates of the same data point projected on to the model surface, and n is the number of data points.

2.2. Finite element modelling of the second stage labour

2.2.1. Pelvic floor model

The pelvic floor model used in the childbirth simulations was constructed based on a MRI scan of a healthy nulliparous woman without any history of pelvic floor dysfunction (Kruger et al., 2008a). The subject had a gynaecoid pelvis, the most common type of bony pelvis in women (Rosse and Gaddum-Rosse, 1997). The model consists of the LA muscles, part of the obturator internus (OI) muscles, and the anterior pelvic bones, including the pubis and part of the ischium. The LA and OI muscles were approximated using a continuous mesh, because they are anatomically connected by the fascial covering of the OI to the arcus tendineus (Pit et al., 2003). In the FE model, the OI attached the LA muscles to the pelvic brim and provided lateral constraints for the deformation of the LA. The coccyx was considered to be flexible and able to rotate during a normal labour, thus having minimal contribution to the mechanics of childbirth and avulsion injury (Hanfy et al., 2011). The posterior aspect of the coccygeus was connected medially to represent the attachment to the coccyx. The resulting tricubic Hermite model of the pelvic floor had 16,320 geometric degrees of freedom.

An isotropic exponential constitutive relation (Eq. (2)) was used to describe the nonlinear mechanical response of the pelvic floor muscles. The mechanical constitutive relation was fitted to the uniaxial tension experimental data of fresh human pelvic floor muscle specimens, reported by Jing et al. (2008).

$$\Psi = a\left[e^{b(I_1-3)} - 1\right] \tag{2}$$

where I_1 is the first invariant of right Cauchy–Green deformation tensor, $a=44.6$ kPa and $b=0.3$ (Li, 2011). It was reported that the human female pubic bone has a Young's modulus of 800 MPa (Dawson et al., 1999). To approximate this, Eq. (2) was used with parameters $a=140$ GPa and $b=0.001$ to achieve an equivalent stiffness and a relatively linear stress–strain relationship at physiological strains. The pubic symphysis was modelled as a Mooney–Rivlin material using Eq. (3).

$$\Psi = c_{10}(I_1-3) + c_{01}(I_2-3) + c_{11}(I_1-3)(I_2-3) \tag{3}$$

where I_1 and I_2 are the first and second invariants of right Cauchy–Green deformation tensor, respectively, $c_{10}=150$ kPa, $c_{01}=450$ kPa, and $c_{11}=600$ kPa (Li et al., 2007).

2.2.2. Simulating the second stage of labour

Simulations of vaginal delivery were performed using each of the 26 skull models. The fetal head was displaced through the pelvic floor over a series of steps, starting from a well-flexed, direct occiput anterior presentation. The position of the posterior fontanelle was prescribed in the z-direction (cranial–caudal direction of the pelvic floor) on the skull model during the simulations, equivalent to applying a monotonically increasing force in this direction until the peak force was reached (Fig. 3). The descent was quantified as the distance between the lowest point of the fetal head and a line joining the ischial spines, being defined as positive when the lowest point of the head was below the ischial spines. The fetal head was able to translate and rotate, negotiating its way through the birth canal, with the only