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**Analysis of pelvic strain in different gait configurations in a validated cohort
of computed tomography based finite element models**

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validation; landmarks

Abstract

The pelvis functions to transmit upper body loads to the lower limbs and is critical in human locomotion. Semi-automated, landmark-based finite element (FE) morphing and mapping techniques eliminate the need for segmentation and have shown to accelerate the generation of multiple specimen-specific pelvic FE models to enable the study of pelvic mechanical behaviour. The purpose of this research was to produce an experimentally validated cohort of specimen-specific FE models of the human pelvis and to use this cohort to analyze pelvic strain patterns during gait. Using an initially segmented specimen-specific pelvic FE model as a source model, four more specimen-specific pelvic FE models were generated from target clinical CT scans using landmark-based morphing and mapping techniques. FE strains from the five models were compared to the experimental strains obtained from cadaveric testing via linear regression analysis, (R^2 values ranging from 0.70 to 0.93). Inter-specimen variability in FE strain distributions was seen among the five specimen-specific pelvic FE models. The validated cohort of specimen-specific pelvic FE models was utilized to examine pelvic strains at different phases of the gait cycle. Each validated specimen-specific FE model was reconfigured into gait cycle phases representing heel-strike/heel-off and midstance/midswing. No significant difference was found in the double-leg stance and heel-strike/heel-off models ($p=0.40$). A trend was observed between double-leg stance and midstance/midswing models ($p=0.07$), and a significant difference was found between heel-strike/heel-off models and midstance/midswing models ($p=0.02$). Significant differences were also found in comparing right vs. left models (heel-strike/heel-off $p=0.14$, midstance/midswing $p=0.04$).

Introduction

The pelvis is a multifaceted low weight, high strength, composite geometric structure, which functions to transmit upper body loads to the lower limbs. Specimen-specific Finite Element (FE) models of the human pelvis have been generated from Computed Tomography (CT) scans using manual and semi-automated segmentation techniques to model pelvic behaviour (Anderson *et al.*, 2005; Barrat *et al.*, 2008; Leung *et al.*, 2009; Seim *et al.*, 2008; Shim *et al.*, 2007; Shim *et al.*, 2008; Thomson *et al.*, 2002; Salo *et al.*, 2015). These models can integrate heterogeneous material properties assigned to bone based on CT attenuation (Dalstra *et al.*, 1995; Anderson *et al.*, 2005). Automated segmentation algorithms have been limited due to low resolution in clinical pelvic CT images, particularly in the areas between the sacroiliac articulations and hip joints (Chen *et al.*, 2011; Seim *et al.*, 2008). As such, most pelvic FE studies to date have used single specimens for analysis due to the time consuming requirements of model generation including segmentation, surface generation and meshing (Anderson *et al.*, 2005; Dalstra *et al.*, 1995; Goel and Svensson 1977; Goel *et al.*, 1978; Li *et al.*, 2007, Leung *et al.*, 2009). FE morphing techniques have been successfully applied in the pelvis and other musculoskeletal structures (Allen *et al.*, 2003; Bah *et al.*, 2009; Baldwin *et al.*, 2010; Chabanas *et al.*, 2003; Couteau *et al.*, 2000; Grosland *et al.*, 2008; Hraiech *et al.*, 2008; O'Reilly and Whyne 2008; Sigal *et al.*, 2008; Tada *et al.*, 2006; Salo *et al.*, 2012). Implementations of morphing and mapping methods that do not require initial segmentation have also shown to accelerate the generation of multiple specimen specific pelvic FE models (Salo *et al.*, 2015).

The pelvis plays a critical role in human locomotion. During gait, the pelvis rotates in three planes, resulting in six different movements permitting efficient and energy saving walking. Stress and strain distributions in the pelvis during different stages of gait have been studied using FE (Ravera

et al. 2015; Ravera *et al.* 2014; Ghosh *et al.* 2015; Zhou *et al.* 2013, Majumder *et al.* 2004), but have been limited to single specimens with non-physiological material properties, loading and boundary conditions, and without experimental validation. The objectives of this study were to generate an experimentally validated cohort of specimen-specific FE models of the human pelvis and to utilize this cohort to analyze pelvic strain patterns during gait.

Materials and Methods

Generating FE models using CT landmark-based mesh morphing and mapping techniques

A specimen-specific FE source model was generated using traditional methods via segmentation of a pelvic CT scan of a 58 year old male (Figure 1a) (Leung *et al.*, 2009). FE models were generated from clinical CT scans of four additional cadaveric pelvises (Table 1), using landmark-based morphing and mapping techniques (Salo *et al.* 2015). Briefly, the source mesh was roughly aligned with each target CT scan based on three manually identified landmarks, followed by automatic alignment of additional manual landmarks to a landmark atlas set and sequential morphing (Figure 1b,c,d,e) (HyperMesh, AmiraDEV) (Salo *et al.*, 2012, Salo *et al.*, 2015 and Boulay *et al.*, 2006,). Thin bone landmark placement errors (Khoury *et al.*, 2008) occasionally led to morphing step failure; recommencement of the algorithm occurred following manual correction of any misplaced landmarks (Figure 1f). During automatic iterative morphing, the designated moving nodes in the source mesh were permitted motion in three degrees-of-freedom (translations) in order to reduce element distortion. All other non-fixed nodes that morphed along with the moving node had six degrees of freedom (translations and rotations) (Figure 1g). Each morphed pelvic FE surface model was further refined through mesh mapping and smoothing.

Material, Boundary Conditions and Analyses

CT density-based Young's moduli were assigned to elements representing interior cortical and trabecular bone (BONEMAT III) (Table 2). To eliminate partial volume edge effects, the outer layer of tetrahedral elements (containing at least one surface node) of the ilia, sacrum and femurs were assigned a constant cortical bone Young's modulus of 17 GPa (Leung *et al.*, 2009). Note, the cortical bone voxels in the CT scan of specimens #3 and #5 had much lower HU values compared to the other pelvic CT scans. As such, the cortical bone surface Young's modulus in these FE models was reduced by 20% to 13.6 GPa based on the HU measurements. The intervertebral discs between the fifth lumbar vertebrae and the sacrum were modeled as degenerated (representative of the condition of the cadaveric specimens) with elastic, isotropic properties (Young's modulus 53MPa, Kurutz and Oroszvary, 2010). Ligaments of the pelvis and hip capsule were modelled as 1D two-noded non-compressible truss elements (Dalstra *et al.*, 1995; Hewitt *et al.*, 2002; Leung *et al.*, 2009). The sacroiliac joints, symphysis pubis and acetabular cartilage were modelled as three-parameter hyperelastic Mooney-Rivlin materials with coefficients of 0.1, 0.45 and 1.67, respectively (Li *et al.*, 2006; Dakin *et al.*, 2001).

Boundary conditions were initially assigned to the FE models to represent the double-leg stance configuration as per the experimental testing (described below) (Figures 2a and b). The position of the femurs with respect to the pelvises in the FE models was optimized within the accuracy of the fiducials (imaged and digitized on each cadaveric pelvis (six) and the associated femurs (six) (Microscribe 3DX)). Distributed concentrated axial compressive loads were applied evenly to all nodes defining the surface of the fifth lumbar vertebrae (345N) with the distal femurs fully constrained from displacement in all directions to represent experimental testing conditions. The load position was optimized to ensure equivalent FE and experimental motion under load. The surface contact between the femoral head and the acetabular cartilage was modeled as Coulomb

frictional (friction coefficient of 0.01) (McLeish and Charnley 1970). All models were solved using non-linear static FE analysis in Abaqus Standard 6.12-1 (HBK, Pawtucket, Rhode Island).

Experimental Testing

The same five fresh-frozen cadaveric pelvises (fifth lumbar vertebrae to the midshaft of the femurs; hip capsule, sacroiliac, iliolumbar, lumbosacral, sacrospinous, and sacrotuberous ligaments intact) which were used to create the specimen-specific FE models, were experimentally tested. Testing was conducted at room temperature with hydration maintained through periodic spraying and soft tissue injections with saline solution. Each specimen was configured in a double-leg stance with the distal femurs and the fifth lumbar vertebra potted in bone cement (Figure 2b). Strain gauges were adhered to each hemi-pelvis (Table 1) (Dalstra *et al.* 1995, Li *et al.* 2007). Similar to the fiducials, the orientation markers on each strain gauge were digitized, identifying the position and direction of each strain gauge (Microscribe 3DX). Loading was applied in axial compression through the fifth lumbar vertebra in a stepwise mode to a final load of 345 N (MTS Bionix 858, Eden Prairie, Minnesota). The loading was replicated ten times (DaqBook 2000a and DBK43-A strain gauge modules, IO-Tech, Cleveland, Ohio), with a strain gauge recalibration after the fifth trial. Experimental strains for each gauge were averaged over the ten runs and adjusted for offset.

Comparison of FE Generated vs. Experimentally Measured Strains

Uniaxial strains were extracted from the models based on the location and directionality of the experimental strain gauges. Equivalent FE strains in the locations of the rosette strain gauges were evaluated by two methods: 1) based on uniaxial strains representing the individual gauges in the rosettes, which were then used to calculate principal strains and 2) principal strains extracted directly from the FE model. The FE strains from the models were compared to the experimental

strains via linear regression analysis (slope, offset, R^2 , RMSE) (Microsoft Office Excel 2007) (Figure 2c).

Modeling and Analysis of Gait Cycle Phases

Once validated, each specimen-specific FE model was reconfigured into gait cycle phases representing heel-strike/heel-off (Figure 2d) and midstance/midswing (Figure 2e). The heel-strike/heel-off gait cycle phase was modelled by rotating the pelvis five degrees about the vertical axis, counter clockwise to designate the right femur to be the heel-strike limb and the left femur to be the heel-off limb and clockwise for a left femur – heel-strike and right femur – heel-off. The heel-strike femur was rotated 30° in flexion and the contralateral femur (heel-off) was rotated 13° in hyperextension. The midstance/midswing gait cycle phase was modelled by rotating the pelvis 5° downward towards the left limb, designating the right limb to be the load bearing - midstance limb and the left limb to be the midswing limb, and subsequently modeling midstance on the left limb. The midswing femur was rotated 30° in flexion to model limb clearance.

In each case, distributed axial compressive load was applied to all nodes on the surface of the fifth lumbar vertebrae (345N). The pelvis was able to rotate under this loading. In the heel-strike/heel-off position both femurs were fully constrained, whereas in the gait model only the midstance femur was constrained. Strains from the double-leg stance, heel-strike/heel-off and midstance/midswing models were compared using two-way repeated measures analysis of variance (ANOVA) (IBM SPSS, Armonk, New York). Additional qualitative comparisons were made with respect to strain patterns throughout the pelvis.

Results

FE vs. Experimental Strains

The source pelvic FE model was effectively reconfigured onto the four target CT scans using the landmark-based semi-automated mesh morphing and mapping techniques. All elements in the models were of sufficient quality for analysis (Table 2). Table 1 shows the viable strain gauges for which data was recorded for each pelvis (Note: some data losses occurred due to defective strain gauges). Comparison of the FE generated strain values to the experimentally measured strains yielded R^2 values ranging from 0.70 to 0.93 (Table 3; correlations shown in Figures 3a and b). As well, during experimental testing pelvis #5 was over rotating and dislocating from the loading mechanism at the highest load level; as such, this FE pelvic model was modeled at a lower load level (230 N) for experimental validation. Mean RMSE for FE strains calculated from each channel was 14%; mean RMSE for FE strains obtained directly from the FE model was 20%.

The maximum principal strains (tensile) and minimum principal strains (compressive) are shown in Figures 4a and 4b, respectively. A bilateral asymmetry in the strain distribution is evident under double-leg stance, which may be attributed to asymmetrical positioning of the femurs during the application of load. The highest strain regions in all five models under double-leg stance loading were found in the supra-acetabular regions (as reported in Shim *et al.* 2010) and in the superior pubic rami. Slightly lower strains were found in the inferior pubic rami. High strains were also found in the regions of the greater sciatic notch and sacral regions close to the sacroiliac joints. These sites correspond to common fracture sites observed in the pelvis (Denis *et al.* 1988). As seen in Salo *et al.* 2015, small alterations in rotation of the loading vector and anterior posterior rotations of the femoral head within the acetabulum had substantial effects on the correlation between experimental and model strains.

Gait Cycle Strain Patterns

Maximum and minimum principal FE strain patterns for all five models are shown in double-leg stance in Figure 4. Maximum principal FE strains in the heel-strike/heel-off and midstance/midswing models are shown in Figure 5. Specifically, in double-leg stance, axial compressive loading applied to the L5 resulted in compressive anteroposterior strains (Figure 6ai) and tensile dorsoventral strains around the base of the sacrum and around the sacral promontory (Figure 6aai). From the L5 vertebra, the compressive load was transmitted through the anterior sacrum into the sacroiliac joints resulting in compressive mediolateral strains (Figure 6b) and tensile dorsoventral strains in the sacroiliac joints. The load transmission continued along the pelvic brim towards the acetabuli, resulting in compressive dorsoventral strains (Figure 6ci) and tensile mediolateral strains throughout the cortical bone of the pelvic brim (Figure 6cii). At the acetabuli, tensile strains occurred in the mediolateral direction (Figure 6di) and compressive strains in the dorsoventral direction along the pubic bone (Figure 6dii). Tensile strains occurred in the dorsoventral direction along the ischium. Mediolateral tensile strains and dorsoventral compressive strains also occurred along the inferior pubic ramus and throughout the pubic symphysis (Figure 6e).

Similar strain patterns occurred in the heel-strike/heel-off stance models. Comparison of FE strains at the locations of the strain gauges showed no significant difference in the double-leg stance and heel-strike/heel-off models ($p=0.40$). Also, the right vs. left heel-strike/heel-off models were not significantly different ($p=0.14$).

In the midstance/midswing models, similar compressive anteroposterior strains and tensile dorsoventral strains occurred around the base of the sacrum and around the sacral promontory. Similar strain patterns were also observed at the sacroiliac joints. As the load transmission continued along the pelvic brim towards the acetabuli, compressive anteroposterior strains (Figure

7ai) and tensile mediolateral strains resulted throughout the cortical bone of the pelvic brim. However, the strains on the midstance side of the pelvis were much larger in magnitude than the strains on the midswing side (Figure 7aii). At the acetabulum of the midstance side, tensile strains occurred in the mediolateral direction (Figure 7bi) and compressive strains in the dorsoventral direction along the pubic bone (Figure 7bii), opposite to that on the midswing side of the pelvis (mediolateral compressive strains and dorsoventral tensile strains) (Figure 7bi and bii). Tensile strains occurred in the dorsoventral direction along the ischium on the midstance side and compressive strains occurred in the dorsoventral direction along the ischium on the midswing side. Also, the tensile strains on the midstance side were larger in magnitude than the compressive strains on the midswing side. Anteroposterior tensile strains occurred in the inferior pubic rami, but were larger in magnitude on the midstance side than the midswing side. Dorsoventral tensile strains and mediolateral compressive strains occurred in the symphysis pubis (Figure 7c). A trend was found with respect to differences in FE strains in comparing the double-leg stance and midstance/midswing models ($p=0.07$). Significant differences were found in comparing the heel-strike/heel-off and midstance/midswing model strains ($p=0.02$) and between right vs. left midstance/midswing models ($p=0.04$).

Discussion

Traditionally, in FE model generation, clinical CT scans of the pelvis have been segmented using thresholding and manual methods due to low contrast in thin bone regions and at cartilage interfaces (Anderson *et al.*, 2005; Seim *et al.*, 2008). Accurate manual segmentation of the pelvis is time consuming and subject to inter and intra user variability, and as such FE studies are often based on single models. The landmark-based mesh morphing and mapping methods allowed geometric reconfiguration of a pelvic source mesh onto four distinct pelvic target CT scans to

generate new specimen-specific FE models without the need for manual CT segmentation. These models exhibited strong correlations with experimental strain data. Similar to previous work, the FE-experimental correlation was highly sensitive to the direction of loading and boundary conditions, highlighting the requirement for precise representation of the experimental set-up in successful model validation (Salo *et al.*, 2015).

In double-leg stance, upper body weight is transmitted from the spine, into the sacrum, through the sacroiliac joints, along the pelvic brim to the acetabuli (Gray 2000). Concurrently, reaction forces from the femurs are transferred to the acetabuli and along the pubic rami. The FE models in the present study demonstrated that the ilia, specifically the iliac crest, underwent the greatest motion (displacement) during forward-backward rotation, which diminished distally along the pelvis. The least amount of displacement occurred at the acetabuli with displacement increasing again distally down the ischia. As expected, the femoral heads were shown to act as pivots upon which the pelvic structure rotated. High tensile strains were found to occur in the FE elements representing the ligaments, specifically those of the hip capsule connecting the femurs to the pelvis (iliofemoral, pubofemoral and ishchiofemoral), which acted to resist over rotation of the pelvis during load transfer. Based on the current element definitions for the elements representing the ligaments of the hip capsule, however, wrapping of these elements around the femoral head was not feasible and remains a limitation of the current models.

Common injury locations in the pelvis include fractures of the sacrum, ilia, rami (unilateral and bilateral) and pubis symphysis, opening of the sacroiliac joint, tear of the sacrospinous and sacroiliac ligaments (Rockwood *et al.* 2010). The validated models in this study showed high strains in the double-leg stance at the greater sciatic notch, along the superior ramus of pubis and along the arcuate line of the ilia, consistent with common pelvic fracture sites (ilia and rami

fractures, respectively). High strains in the sacroiliac and symphysis pubis joints, correspond to locations of fractures at the pubis symphysis, opening of the sacroiliac joint and sacral fractures. Inter-specimen variability in the FE strain distributions was observed among the five specimen-specific pelvic FE models (Figure 4). The inter-specimen variability may be influenced by both subject specific geometry and bone density. Pelvis #1 (lowest magnitude strains) was a male pelvis of android shape (heart shaped pelvic inlet and narrow subpubic arch). The other four pelvises were female of gynecoid shape (rounded pelvic brim and less prominent ischial spines). The reduced bone density of Pelvis #3 yielded higher strains under a similar magnitude of loading. Right vs. left side specific differences in FE strain distributions resulted due to variability in load application direction and positioning of the femurs within the acetabuli, which caused uneven force transmission. Specifically in model #5, the left arcuate line was in tension and the sciatic notch was in compression, opposite to the right side of the pelvis. These results suggest that the position of the femoral heads within acetabuli during loading induced rotation in the pelvic structure. Promising validations have been previously reported from FE pelvis studies in the literature. A study by Shim *et al.* 2010 compared strains of three hemi-pelvic FE models generated from specimen-specific CT scans to experimentally obtained strains and achieved strong R^2 (0.87 to 0.95) and RMSE (14% to 29%) values. A more anatomically complex model developed by Hao *et al.* 2011 reported an $R^2=0.97$ for a single pelvic FE model in double leg stance, which included pelvic bones, proximal femurs and main pelvic ligaments. However, experimental cadaveric validation of this model used steel implant fixtures to replace the proximal femurs, which along with the anterior part of the pubic bone, were fixed in all degrees of motion. Load transfer in the pelvis occurs from the sacrum through the sacroiliac joint to the pelvic girdle. As such, the inclusion of the sacroiliac joint in a pelvic FE model is significant for modeling proper load

transfer. From the pelvic girdle the load is transferred to the femurs inducing femoral head motion within the acetabulum. At the same time, the pubic symphysis functions to resist forces resulting from the movement of the ilia. Constraining the femoral head and the pubic symphysis will alter load transfer and the resulting stress and strain distributions throughout the pelvic structure. In the current study, the R^2 values (0.70 to 0.93) represent a robust validation of the modeling approach considering the complexity of the double-leg stance pelvic system, allowing for motion at the sacroiliac joints, motion of the femoral heads within the acetabuli and compression of the pubic symphysis. Similar to our models, the work of Ravena *et al.* 2015 from a single specimen-specific FE model of the pelvis demonstrated higher stresses in the pelvis at a midstance/midswing phase (3MPa) compared to the heel strike/heel off phase (1MPa).

In the current work, the models were sensitive to uniform bone values applied to surface elements representing cortical bone (Figure 4). Adjustments to lower density specimens improved experimental and FE strain correlations. Image pre-processing methods, such as deblurring and nodal based material property assignments, may better represent the heterogeneity of these surface elements (Pakdel *et al.*, 2016). Despite the consistent FE modeling and experimental protocol, there was considerable intra-specimen variability with respect to the strength of the correlation between the FE and experimental results. Limited CT resolution may have a greater impact on lower density specimens based on consistent threshold levels applied for mapping. Post processing of the rosette gage strain data further demonstrates the impact of two distinct methods for calculation of principal strains on the validation results. The error inherent in representing the pelvic experimental alignment and strain values in the FE models may also have contributed to the variability of the validation. The applied loading was sufficient to generate adequate strains within the pelvic structures, ligaments and FE models, while not causing fracture of the bone or rupturing

of any of the ligaments. However, soft tissues and musculature, which contribute to the stabilization of the pelvis, were not included in the experimental or FE models, limiting the magnitude of applied load.

Overall, the landmark-based morphing/mapping method proved adequately robust with respect to the creation and experimental validation of a cohort of specimen-specific pelvic FE models. This work provides a robust platform to ultimately investigate normal pelvic biomechanical behaviour for a broader population (encompassing geometric and material differences), biomechanical implications of pathological states (i.e. osteoporosis, metastatic disease and traumatic injuries), complex loading conditions (that are challenging to represent with other analytic or experimental techniques) and the optimization of restoration and reconstruction approaches in the pelvis.

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Conflict of Interest Statement

The authors have no conflict of interest.

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Table 1: Demographic details, modelling and strain gauge information for cadaveric pelvises. All CT scans were conducted under clinical protocols (CT scanning parameters: GE Lightspeed Plus CT scanner, 512 x 512 acquisition matrix, in-plane resolution 0.94 x 0.94 mm, slice thickness 3.2 mm, 99 slices). Strain gauges were (Uniaxial strain gauges: resistance of 120 ohm FLA-3-11-3LT and rosette strain gauges: resistance of 120 ohm, FRA-5-11-3LT; Tokyo Sokki Kenkyujo Co. Ltd., Tokyo, Japan)

Pelvis number	Gender	Age	Model	# of strain gauges	
				Uniaxial	Rosette
1	Male	58	Source FE model	12	0
2	Female	56	Target CT scan	5	2
3	Female	81	Target CT scan	6	1
4	Female	83	Target CT scan	5	2
5	Female	82	Target CT scan	6	2

Table 2: Quality assessment, volume and CT-based Young's modulus for the FE models. For each element quality parameter the minimum or maximum value found in the mesh is reported along with the percent of elements that failed each set quality criteria (italicized brackets).

Quality parameter	1	2	3	4	5
Number of Elements	1,226,256	1,397,355	1,439,821	1,463,251	1,353,386
Number of Nodes	243,503	271,060	276,285	254,962	261,940
Aspect ratio > 5	11 (0.004 %)	19 (0.002 %)	26 (0.004 %)	24 (0.009 %)	89 (0.003 %)
Skew > 60	1 (0 %)	1 (0 %)	1 (0 %)	1 (0 %)	1 (0 %)
Jacobian < 0.7	1 (0 %)	1 (0 %)	1 (0 %)	1 (0 %)	1 (0 %)
Minimum angle < 20°	6 (0.1 %)	3 (0.06 %)	2 (0.07 %)	1 (0.1 %)	1 (0.09 %)
Maximum angle > 120°	152 (0.06 %)	164 (0.04 %)	169 (0.04 %)	172 (0.08 %)	170 (0.06 %)
Volume (cm ³)	2,218	1,581	1,695	1,721	1,452
CT-based Young's modulus (MPa)	694.6 ± 1136.3	742.1 ± 913.0	67.4 ± 349.7	235.2 ± 578.6	680.1 ± 2208.0

Table 3: Comparison of slope, offset and R^2 values between strain values obtained from the FE models and the experimentally measured strain results

a) based on FE principal strains calculated from uniaxial strains representing individual gauges in the rosettes.

	1	2	3*	4	5*^
Slope	1.01	1.02	0.92	1.13	0.57
Offset	0.09	6.03	-16.13	-9.88	-28.01
R^2	0.93	0.93	0.79	0.88	0.74
Root Mean Square Error	8%	11%	15%	13%	23%

b) based on FE principal strains extracted directly from each FE model.

	1	2	3*	4	5*^
Slope	1.01	1.60	0.96	1.20	0.69
Offset	0.09	8.83	-39.31	5.04	-21.92
R^2	0.93	0.70	0.86	0.86	0.87
Root Mean Square Error	8%	44%	13%	15%	17%

*Cortical bone $Y = 14\text{GPa}$

^ Maximum Load = 230 N

Figure Legends:

Figure 1: Generation of a specimen-specific pelvic FE model using a landmark-based morphing technique.

Figure 2: Assignment of FE boundary conditions to three stance configurations and validation of specimen-specific FE models with cadaveric experimental testing in double-leg stance.

Figure 3: Overall model correlations of the strains predicted from the five FE pelvic models constructed using morphing and meshing techniques with the strains measured on the pelvic cadaveric in vitro experiments. The region bound by the blue dashed lines is the 95 % confidence region of linear fit. **a)** Based on uniaxial strains representing the individual gauges in the rosettes, which were then used to calculate principal strains. **b)** Based on principal strains extracted directly from the FE model.

Figure 4: Principal strain patterns for the five specimen-specific FE models under double leg stance. Scale bars shown are for maximum and minimum principal strains, respectively. Note, the maximum principal strain scale bar also applies to figures 5-7.

Figure 5: Maximum principal strains for midstance/midswing and heel strike/heel off configurations.

Figure 6: Detailed descriptions of the FE strain patterns in double-leg strain configuration (Compressive strains Tensile strains).

545 **Figure 7:** Detailed descriptions of the FE strain patterns in midstance/midswing configuration
546 (Compressive strains Tensile strains).

a) Segmented source FE model from CT

b) Align source FE model with target CT using three landmarks

c) 40 initial landmarks (yellow)

d) Landmark atlas (red) aligned with initial landmarks by similarity alignment

e) Landmark atlas morphed (green) to the initial landmarks

f) Manual landmark correction (wrong position (yellow), correct position (red))

g) Source FE model (gray) morphed onto target CT landmarks (orange)

Landmarks

a) Double leg stance load and boundary conditions

b) Cadaveric pelvis testing in double leg stance

c) Validated FE model

d) Heel strike/heel off position

e) Midstance/midswing position

Figure 3: Overall model correlations of the strains predicted from the five FE pelvic models constructed using morphing and meshing techniques with the strains measured on the pelvic cadaveric in vitro experiments. The region bound by the blue dashed lines is the 95 % confidence region of linear fit.

3 a) Based on uniaxial strains representing the individual gauges in the rosettes, which were then used to calculate principal strains.

3 b) Based on principal strains extracted directly from the FE model.

a) Maximum principal strains

b) Minimum principal strains

Pelvis #1

Pelvis #2

Pelvis #3

Pelvis #4

Pelvis #5

ai) Compressive anteroposterior strains around the base of the sacrum

aii) Tensile dorsoventral strains around sacral promontory

b) Compressive mediolateral strains in the sacroiliac joints

ci) Compressive dorsoventral strains in the cortical bone of the pelvic brim

cii) Tensile mediolateral strains in the cortical bone of the pelvic brim

di) Tensile mediolateral strains along the pubic bone

dii) Compressive dorsoventral strains along the pubic bone

e) Mediolateral tensile strains along the inferior pubic ramus and throughout the pubic symphysis

ai) Compressive anteroposterior strains around the base of the sacrum

aii) Tensile dorsoventral strains around sacral promontory

b) Compressive mediolateral strains in the sacroiliac joints

ci) Compressive dorsoventral strains in the cortical bone of the pelvic

cii) Tensile mediolateral strains in the cortical bone of the pelvic brim

di) Tensile mediolateral strains along the pubic bone

dii) Compressive dorsoventral strains along the pubic bone

e) Mediolateral tensile strains along the inferior pubic ramus and throughout the pubic symphysis

ai) Compressive anteroposterior strains in the cortical bone of the pelvic brim on the midstance side

aii) Compressive anteroposterior strains in the cortical bone of the pelvic brim on the midswing side

bi) Tensile mediolateral strains on the midstance side and compressive mediolateral strains on the midswing side along the pubic bone

bii) Compressive dorsoventral strains on the midstance side and tensile mediolateral strains on the midswing side along the pubic bone

c) Dorsoventral tensile strains along the inferior pubic ramus and mediolateral compressive strains throughout the pubic symphysis