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Sensitivity of femoral strain calculations to anatomical scaling errors in musculoskeletal models of movement

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ABSTRACT

The determination of femoral strain in post-menopausal women is important for studying bone fragility. Femoral strain can be calculated using a reference musculoskeletal model scaled to participant anatomies (referred to as scaled-generic) combined with finite-element models. However, anthropometric errors committed while scaling affect the calculation of femoral strains. We assessed the sensitivity of femoral strain calculations to scaled-generic anthropometric errors. We obtained CT images of the pelves and femora of 10 healthy post-menopausal women and collected gait data from each participant during six weight-bearing tasks. Scaled-generic musculoskeletal models were generated using skin-mounted marker distances. Image-based models were created by modifying the scaled-generic models using muscle and joint parameters obtained from the CT data. Scaled-generic and image-based muscle and hip joint forces were determined by optimisation. A finite-element model of each femur was generated from the CT images, and both image-based and scaled-generic principal strains were computed in 32 regions throughout the femur. The intra-participant regional RMS error increased from 380 με (R^2=0.92, $p<0.001$) to 4064 με (R^2=0.48, $p<0.001$), representing 5.2% and 55.6% of the tensile yield strain in bone, respectively. The peak strain difference increased from 2821 με in the proximal region to 34,166 με at the distal end of the femur. The inter-participant RMS error throughout the 32 femoral regions was 430 με (R^2=0.95, $p<0.001$), representing 5.9% of bone tensile yield strain. We conclude that scaled-generic models can be used for determining cohort-based averages of femoral strain whereas image-based models are better suited for calculating participant-specific strains throughout the femur.

1. Introduction

The quantification of femoral strain during daily activities is important for understanding the biomechanical implications of osteoporosis (Van Rietbergen et al., 2003), for which post-menopausal women are most at risk. For example, intra-participant femoral strains can provide information about fracture risk (Cody et al., 1999) while inter-participant averages can provide insights into understanding the bone response to exercise treatments (Lang et al., 2014). In vivo femoral strains can be estimated non-invasively using a scaled-generic musculoskeletal model scaled to participant anatomies (herein referred to as 'scaled-generic models') combined with a finite-element model of the femur (Jonkers et al., 2008; Martelli et al., 2014a). However, errors in the definition of the model anthropometry affect calculation of muscle forces (Lenaerts et al., 2009), which likely propagate to bone strain calculation. Several studies have investigated the sensitivity of muscle and joint force calculations to uncertainties in anatomical and muscle parameters (Ackland et al., 2012; Correa et al., 2011; Martelli et al., 2015; Redl et al., 2007; Scheys et al., 2009; Xiao and Higginson, 2010) while others have examined the sensitivity of femoral strain calculations to uncertainties in measurements of the geometry and material properties of the femur (Taddei et al., 2006). To date, no study has investigated the sensitivity of femoral strain calculations to anthropometric errors arising from uncertainties in, for example, body-segmental masses and lengths.

Magnetic-resonance (MR) and computed-tomography (CT) images can provide detailed anthropometric information about the human musculoskeletal system. While MR imaging is the preferred method for acquiring muscle–tendon attachment sites and paths, joint centre positions, and the orientations of joint rotation axes (Blemker et al., 2007; Scheys et al., 2008), this approach is not suitable for extracting bone mineral density

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(BMD), which is needed to model the elastic properties of bone (Schileo et al., 2007). Alternatively, bone surfaces, joint centres and orientations can be determined by segmenting CT images (Taddei et al., 2012), and the images' Housfield unit data can be used to describe the BMD and elastic property distributions (Schileo et al., 2007). Although the low contrast of CT images complicates extracting soft-tissue anatomical structures such as muscles, CT images can serve as a reference for registering a muscular system atlas to a participant's anatomy (Abdel Fatah et al., 2012; Taddei et al., 2012). Therefore, CT images can provide all information necessary to generate both musculoskeletal and finite-element models of a specific participant (herein referred to as 'image-based models').

Scaling procedures have been used to generate musculoskeletal models of participants by applying a limited number of anthropometric parameters to a scaling algorithm (Delp et al., 2007, 1990). Typically, the body mass and segment lengths in a generic-reference model are scaled to an individual participant using information from the skin-mounted marker positions and ground reaction forces acquired during a static pose, thereby creating a 'scaled-generic' model. Scaled-generic models have been successfully used to study general patterns of human motion (Correa et al., 2010; Delp et al., 1990). However, scaling causes unavoidable anthropometric errors, which in turn may compromise the assessment of individual features in muscle and joint force patterns (Lenaerts et al., 2009).

Previous studies addressing the sensitivity of scaled-generic models investigated different model outputs and reached different conclusions. Correa et al. (2011) concluded that scaled-generic models are as accurate as image-based models when evaluating the potential (per-unit-force) contributions of individual muscles to joint and centre-of-mass accelerations during walking. Lenaerts et al. (2009) concluded that participant-specific hip geometry is important in the calculation of hip contact forces while walking; they reported average differences between scaled-generic and image-based models of 0.52 times body weight (BW). No study has reported the sensitivity of femoral strain calculations to anthropometric errors committed while scaling a scaled-generic model to participants' anatomies. However, this information is essential for understanding the limits of applicability of the model results (Viceconti et al., 2005).

The aim of this study was to investigate how anthropometric errors introduced when scaling a scaled-generic musculoskeletal model to a participant's anatomy propagate to femoral strain calculations. Femoral strains were computed using scaled-generic and image-based models of 10 participants for six weight-bearing tasks. The influence of scaled-generic anthropometric errors was assessed by analysing (a) participant-specific (intra-participant) femoral strains, and (b) average (inter-participant) femoral strains within a cohort.

2. Materials and methods

Ten healthy post-menopausal women (age, 66.7 ± 7.0 years; height, 159 ± 6.6 cm; weight, 66.3 ± 22.5 kg) were recruited to this study (Table 1). All participants could walk unassisted and had no reported history of musculoskeletal disease. Ethics approval for the study was obtained from the Human Research Ethics Committee at the University of Melbourne.

2.1. Data collection

CT images of the pelvic and thigh regions of each participant were obtained using a clinical whole-body scanner (Aquilon CT, Toshiba Corporation, Tokyo) and an axial scanning protocol (tube voltage: 120 kV; tube current: 200 mA). For each scan, two datasets of monochromatic, 16-bit, 512 × 512 pixel images with slice thickness of 0.5 mm and spacing of 0.5 mm were obtained. The femur dataset was reconstructed using an in-plane transverse resolution of 0.5 × 0.5 mm^2 whereas the pelvis dataset was reconstructed using an adjusted in-plane transverse resolution to accommodate the entire pelvis. A five-sample (hydroxyapatite density range: 0–200 mg/cm^3) calibration phantom (Mindways Software, Inc., Austin, TX) was placed below the participant's dominant leg while scanning.

Table 1
Participant details (all female).

Participant	Age (years)	Weight (kg)	Height (cm)	BMI (kg/m^2)
1	74	51	150	22.7
2	64	52	150	23.1
3	72	66	158	26.6
4	68	61	158	24.6
5	68	53	159	21.0
6	60	85	153	36.3
7	60	96	170	33.1
8	64	69	168	24.6
9	64	71	165	26.1
10	73	59	157	23.9

BMI = Body mass index.

Gait analysis experiments were performed at the Biomotion Laboratory, University of Melbourne. Forty-six skin-mounted reflective markers were attached to anatomical locations as described by Dorn et al. (2012), including the pelvis (3), thigh (6), shank (5) and foot (6). The remaining markers were placed along the upper extremities and torso. Marker trajectories were recorded with a 10-camera motion capture system (VICON, Oxford Metrics Group, Oxford) sampling at 120 Hz. Each participant was instructed to (a) walk at a self-selected speed; (b) walk at a faster self-selected speed; (c) ascend and descend a flight of 3 steps (step height = 16.5 cm) at self-selected speeds while engaging with the first step of the staircase using the dominant foot; (d) rise from and sit on a chair (chair height = 47 cm); and (e) jump as high as possible from a comfortable standing position with each foot placed on a separate force platform. Five repetitions of each task were executed. Ground reaction forces and moments were recorded using three strain-gauged force plates (AMTI, Watertown, MA) sampling at 2000 Hz. The ground force data were low-pass filtered using a fourth-order, recursive, zero-lag, Butterworth filter with a cut-off frequency of 40 Hz. A static trial was recorded to measure the inter-marker distances. Marker trajectories were low-pass filtered using a second-order recursive, zero-lag, Butterworth filter with a cut-off frequency of 6 Hz.

2.2. Musculoskeletal modelling

The scaled-generic and image-based musculoskeletal models were based on the generic model developed by Dorn et al. (2012). The generic model was comprised of 12 segments with 31 independent degrees-of-freedom actuated by 92 Hill-type muscle–tendon units (Fig. 1A). A ball-and-socket joint represented the lumbar joint, each shoulder, and each hip; a translating hinge joint represented each knee; and a universal joint represented each ankle. The shoulder and elbow joints were actuated by 10 ideal torque motors, while all other joints were actuated by Hill-type muscle–tendon units.

Scaled-generic models were obtained by scaling the generic model to match each participant's body anthropometry and mass using OpenSim (Delp et al., 2007). Inter-marker distances recorded during the static trial (Fig. 1B) were used to scale bone geometries, joint centres, joint rotation axes, muscle paths, fibre lengths, and tendon slack lengths. The mass of the generic model was scaled to match that of each participant by preserving the mass ratio between segments in the generic model. Image-based models were created using anthropometric measurements obtained from the CT images for the pelvis and femur segments, skin-marker locations for the torso, and scaled-generic parameters for the remaining segments. The geometries of the pelves and femora were segmented from the CT data using Amira (Visage Imaging GmbH, Burlington, MA). The hip joint centre was defined as the centre of the sphere used to best-fit the femoral head surface. The knee axis was assumed to be the axis connecting the femoral epicondyles, and the lumbar joint was assumed to be located at the antero-posterior level of the vertebral foramen and at the mid-point of the L5-S1 inter-vertebral space as identified in the sagittal plane. The torso was adjusted to match the vertical distance between the sacrum and the seventh cervical spine calculated from the skin-mounted markers (Fig. 1). Muscle paths in the scaled-generic model were registered on the skeletal surfaces by superimposing the muscle lines-of-action onto the CT data (Fig. 1C). The values of optimum muscle-fibre length and tendon slack length reported by Delp et al. (1990) were uniformly scaled so that each muscle develop its peak isometric force at the same joint angle in both the scaled-generic and image-based models.

Scaled-generic and image-based muscle and joint forces were calculated for the dominant leg of a selected trial. Joint angles were computed by performing an inverse kinematics analysis according to methods described by Delp et al. (2007). The joint angles and the measured ground reaction forces were used to calculate