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An integrative modeling approach for the efficient estimation of cross sectional tibial stresses during locomotion

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ABSTRACT

The purpose of this research was to utilize a series of models to estimate the stress in a cross section of the tibia, located 62% from the proximal end, during walking. Twenty-eight male, active duty soldiers walked on an instrumented treadmill while external force data and kinematics were recorded. A rigid body model was used to estimate joint moments and reaction forces. A musculoskeletal model was used to gather muscle length, muscle velocity, moment arm and orientation information. Optimization procedures were used to estimate muscle forces and finally internal bone forces and moments were applied to an inhomogeneous, subject specific bone model obtained from CT scans to estimate stress in the bone cross section. Validity was assessed by comparison to stresses calculated from strain gage data in the literature and sensitivity was investigated using two simplified versions of the bone model-a homogeneous model and an ellipse approximation. Peak compressive stress occurred on the posterior aspect of the cross section (-47.5 ± 14.9 MPa). Peak tensile stress occurred on the anterior aspect (27.0 ± 11.7 MPa) while the location of peak shear was variable between subjects (7.2 ± 2.4 MPa). Peak compressive, tensile and shear stresses were within 0.52 MPa, 0.36 MPa and 3.02 MPa respectively of those calculated from the converted strain gage data. Peak values from a inhomogeneous model of the bone correlated well with homogeneous model (normal: 0.99; shear: 0.94) as did the normal ellipse model (r=0.89–0.96). However, the relationship between shear stress in the inhomogeneous model and ellipse model was less accurate (r=0.64). The procedures detailed in this paper provide a non-invasive and relatively quick method of estimating cross sectional stress that holds promise for assessing injury and osteogenic stimulus in bone during normal physical activity.

1. Introduction

In vivo stresses in lower extremity long bones are difficult to quantify experimentally and estimate computationally. However, these stresses represent the internal loading intensities within bone. They are dependent on the applied loading magnitude, bone structural geometry and material properties which play a fundamental role in skeletal injury and adaptation. One method of obtaining bone stress in vivo is the surgical attachment of strain gages directly to the bone (Milgrom et al., 2007) and subsequent calculation of stress (Lanyon et al., 1975). Another method is the application of finite element methods with bone models derived from advanced imaging techniques (Gray et al., 2008; Speirs et al., 2007). Strain gage analyses are limited to superficial, periosteal regions of bone, may produce localized pain or numbness in active subjects, and are often difficult to get approved by institutional review boards. Finite element methods often suffer from a lack of realistic boundary conditions and muscle forces, are relatively time consuming for subject-specific model generation, and can be computationally intensive when utilizing a clinically relevant sample size. A compromise may be a two-dimensional finite element model of a transverse cross section of bone at a specific area of interest. Three-dimensional forces and moments at the cross sectional centroid can be estimated using muscle and joint reaction forces obtained from a combination of experimentation and musculoskeletal modeling. A cross-sectional finite element model can then be used to estimate bone stress. This method allows for subject-specific bone geometry with inhomogeneous material properties, yet is simple, accurate and semi-automated enough to have potential in a clinical setting (Kourtis et al., 2008).

The selection of a particular bone model is dependent on the available equipment and may depend on the purpose of the research. Clinical computed tomography (CT) and peripheral

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quantitative CT (pQCT) not only provide information about bone geometry, but also the apparent density distribution – necessary information for the construction of an inhomogeneous finite element model (Rho et al., 1995). However, these technologies expose individuals to varying degrees of ionizing radiation. On the other hand, magnetic resonance imaging (MRI) does not involve radiation and structural geometry can still be obtained but bone apparent density is not directly available. Advanced calibration techniques may allow MRI derived measurement of bone porosity which makes it possible to estimate apparent density (Hong et al., 2000). Both MRI and CT acquisition require expensive equipment with user rates of several hundred dollars per scan. A simple model of a bone cross section can be estimated without the use of these imaging techniques by modeling the bone geometry as an idealized hollow ellipse (Utz et al., 2009). Added subject specificity can be included in these models by directly measuring bone diameters from inexpensive radiographs (O'Neill and Ruff, 2004) or using group specific regression equations (Franklyn et al., 2008).

The purpose of this research was to develop and test an integrative modeling approach for the purpose of estimating tibial stresses (i.e., normal and shear) during locomotion without invasive techniques such as strain gages, or difficulties associated with subject-specific whole-bone finite element model generation. The tibia was chosen as the bone of interest, because it represents the location most prone to fatigue fracture in athletes (Sanderlin and Raspa, 2003) and military personnel (Milgrom et al., 1985) due to the large axial/bending loads it experiences during physical activity. The approximate junction of the middle and distal third of the tibial was chosen because it is the most prevalent site of stress fractures in adults (Sanderlin and Raspa, 2003). We hypothesized that restricting the location of stress estimation to the anteriomedial tibia, as used in previous strain gage studies, would substantially underestimate the peak stress magnitudes located elsewhere in the cross section. Also, we compared peak stress magnitudes from three different bone models of decreasing complexity. The first bone model consisted of a finite element mesh with geometry and inhomogeneous material properties derived from pQCT scans; the second model assumed identical geometry with homogeneous material properties assigned to each element (representing typical MRI data); the third model assumed bone cross sectional geometry to be an idealized hollow ellipse. We hypothesized that the simplified models would produce lower peak stress values than the inhomogeneous finite element mesh.

2. Methods

2.1. Protocol

Twenty-eight male, active duty soldiers (age: 20.5 ± 3.1 yr, mass: 85.4 ± 14.8 kg, height: 1.77 ± 0.08 m) volunteered and provided informed consent to participate in this research study. They walked on an instrumented treadmill (AMTI, Watertown, MA) for five minutes at 1.34 m/s and then external reaction forces and moments were collected at 2000 Hz. Retro-reflective markers were placed on the pelvis, right thigh, right leg and right foot to define the orientation of these segments. Markers located on the sacrum and both PSIS were used to define the pelvis and four-marker clusters were placed on the thigh, leg and foot. Additional anatomical markers were placed on both ASIS, medial and lateral condyles of the knee and medial and lateral malleoli of the ankle to help identify hip, knee and ankle joint centers. The hip joint center was calculated using the method of Bell et al. (1989) while knee and ankle joint centers were calculated using the mean of the medial and lateral markers at each joint. Markers were digitized at 200 Hz using a 10-camera Qualisys system, synchronized with the instrumented treadmill. Five consecutive stance phases were selected for analysis with each stance phase classified as one trial of data.

2.2. Rigid body model

Segment masses were estimated using the regression equations of Dempster (1955) and segment moments of inertia and center of mass locations were estimated using the procedures of Hanavan (1964). Three-dimensional joint moments and reaction forces were estimated using inverse dynamics procedures. Data filtering was accomplished using the procedures of Edwards et al. (2011) to minimize artifact that can arise from differing analog and kinematic cutoff frequencies.

2.3. Musculoskeletal model

Joint angles derived from the marker data were used as input to a musculoskeletal model. This model was implemented in Matlab and used to estimate length and velocity adjusted maximal muscle forces, moment arms and insertion points for each of 44 lower extremity muscles during the stance phase of the walking cycle. Muscle parameters were obtained from Arnold et al. (2010). A maximal muscle velocity of 20 fiber lengths per second was assumed; muscle lengths were scaled to the length of the segments, and maximal isometric muscle forces were scaled to the masses of the individual subjects and then multiplied by 1.25 to account for the athletic subject pool. These maximal dynamic muscle force estimations were only used to constrain the optimization procedure.

2.4. Optimization

The joint moments derived from the rigid body model and the moment arms derived from the musculoskeletal model were used as inputs to a static optimization algorithm implemented in Matlab (*fmincon* function with the interior-point algorithm). The cost function (u) to be minimized was the sum of the squared muscle stresses (Glitsch and Baumann, 1997) and was subject to the constraints that 1) the internal moments calculated from the inverse dynamics were equal to the moments caused by the muscles and 2) the individual muscle forces were not less than zero nor greater than the maximal dynamic muscle forces estimated from the musculoskeletal model:

$$u = Min \sum_{i=1}^{44} \left[\frac{F_i}{A_i}\right]^2$$

Subject to: $r_{jk} \times F_i = M_{jk}$ and $0 \le F_i \le \mathrm{Fmax}_i$

where F_i is the estimated muscle force of the ith muscle, A_i is the physiological cross sectional area of the ith muscle, r_{jk} is the muscle moment arm for each jth joint and kth plane of motion, M_{jk} is the joint moment for the jth joint and kth plane of motion, and Fmax_i is the maximal dynamic muscle force for the ith muscle. Note that not all planes of motion were utilized at all of the joints; only the sagittal plane hip, knee and ankle as well as frontal plane hip and ankle moments were included in the optimization.

2.5. Bone models

Inputs to the bone models were the 3D forces and moments acting at the centroid of the cross section of the bone. These loads were calculated by translating the effects of the knee joint contact force to the centroid location while subtracting the effects of muscles that insert proximal to the centroid. Centroid loads were