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The capacity of the human iliotibial band to store elastic energy during running

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ABSTRACT

The human iliotibial band (ITB) is a poorly understood fascial structure that may contribute to energy savings during locomotion. This study evaluated the capacity of the ITB to store and release elastic energy during running, at speeds ranging from 2–5 m/s, using a model that characterizes the three-dimensional musculoskeletal geometry of the human lower limb and the force–length properties of the ITB, tensor fascia lata (TFL), and gluteus maximus (GMax). The model was based on detailed analyses of muscle architecture, dissections of 3-D anatomy, and measurements of the muscles' moment arms about the hip and knee in five cadaveric specimens. The model was used, in combination with measured joint kinematics and published EMG recordings, to estimate the forces and corresponding strains in the ITB during running. We found that forces generated by TFL and GMax during running stretch the ITB substantially, resulting in energy storage. Anterior and posterior regions of the ITB muscle–tendon units (MTUs) show distinct length change patterns, in part due to different moment arms at the hip and knee. The posterior ITB MTU likely stores more energy than the anterior ITB MTU because it transmits larger muscle forces. We estimate that the ITB stores about 1 J of energy per stride during slow running and 7 J during fast running, which represents approximately 14% of the energy stored in the Achilles tendon at a comparable speed. This previously unrecognized mechanism for storing elastic energy may be an adaptation to increase human locomotor economy.

1. Introduction

Because bipedalism is a fundamental derived feature of hominins (species more closely related to humans than chimpanzees), many distinctive features of the human spine and lower extremity are adaptations to improve bipedal locomotor performance. Many adaptations for standing and walking, for example, appear early in hominin evolution including a inferiorly-oriented foramen magnum, a lordotic lumbar spine, and a sagittally-oriented ilium (see Aiello and Dean (1990) and Zollikofer et al. (2005)). Additional features that first appear later in the genus *Homo* may reflect selection for endurance running, including a stabilized sacroiliac joint, an expanded attachment of gluteus maximus, and shorter toes (Bramble and Lieberman, 2004; Lieberman et al., 2006; Rolian et al., 2009). Although the selective factors underlying the evolution of both walking and running are debated, it is likely that locomotor economy played a key role. Hypothesized energy-saving features for walking include long legs and dorsally oriented ischia (Crompton et al., 1998; Pontzer et al., 2009; Robinson, 1972; Sockol et al., 2007). Energy saving features for running in the genus *Homo* include a long, compliant Achilles tendon and a spring-like median longitudinal arch, which are known to store and recover elastic energy during running in other vertebrates (Biewener, 2003; Ker et al., 1987; Roberts, 2002). In addition, the human lower extremity has a number of fascial structures with elastic properties that are not present in apes, but whether these structures store energy or serve another function remains poorly understood.

One of the most interesting of these structures is the iliotibial band (ITB). The ITB is a thickening of the lateral fascia of the thigh that originates on the pelvis and inserts on the tibia; it receives muscle fibers from the tensor fascia lata (TFL) anteriorly and from the gluteus maximus (GMax) posteriorly (Gottschalk et al., 1989; Gray et al., 1995; Kaplan, 1958; Ober, 1936; Stern, 1972). The ITB is traditionally considered to function as a “strut” during walking, stabilizing the hip in the frontal plane (Gottschalk et al., 1989; Inman, 1947; Kaplan, 1958). However, the high compliance of the ITB (Butler et al., 1984; Derwin et al., 2008; Gratz, 1931), the fact that it crosses both the hip and knee, and the presence of in-series

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muscles suggest that the ITB may play other roles. If the ITB stretches substantially while transmitting muscle forces, storing elastic energy, then it may decrease the metabolic cost of locomotion. Prior studies have estimated that energy recovered from the Achilles tendon during running reduces muscle work by as much as 35% (Alexander and Bennet-Clark, 1977; Ker et al., 1987). Whether the ITB also stores and recovers elastic energy, and how this compares to Achilles tendon energy recovery, is unknown.

As a first step toward evaluating the ITB's role in locomotor economy, this study examined the capacity of the ITB to store elastic energy at running speeds ranging from 2 to 5 m/s. We hypothesized that forces generated by TFL and GMax stretch the ITB during running, storing elastic energy that may be recovered later in the stride. We tested this hypothesis by developing a musculoskeletal model of the ITB and inserting muscles. Our model characterizes the 3-D skeletal geometry, the hip and knee kinematics, and the attachments and force–length (F–L) properties of the ITB, TFL and GMax for an average-sized adult male (femur length: 39.8 cm; tibia length: 36.2 cm). Because existing representations of TFL and GMax were not sufficiently accurate for this study, we performed detailed analyses of these muscles' architecture and measured their moment arms (MAs) about the hip and knee in cadaveric specimens. The TFL has largely been neglected in previous studies of muscle architecture (e.g., Ward et al. (2009) and Wickiewicz et al. (1983)) and locomotor function (e.g., Dorn et al. (2012) and Sasaki and Neptune (2006)), despite being active during running (Andersson et al., 1997; Mann et al., 1986; Montgomery et al., 1994; Paré et al., 1981). GMax is routinely modeled as a uniarticular hip extensor that inserts on the femur (e.g., Arnold et al. (2010) and Delp et al. (1990)), despite evidence that a substantial portion of GMax inserts on the ITB (Gray et al., 1995; Stern, 1972). Our refined musculoskeletal model, which addresses these limitations, is available on SimTK (simtk.org). Using this model, we estimated the forces transmitted to anterior and posterior regions of the ITB at body positions corresponding to running, predicted the length changes of each region, and calculated the corresponding ITB strain energies over the course of a stride based on published measurements of the tissue's elastic modulus (Butler et al., 1984; Derwin et al., 2008).

2. Materials and methods

2.1. Muscle architecture measurements

We characterized the isometric force-generating capacity of TFL and GMax based on measurements of muscle architecture in three formalin-fixed human cadavers (Table 1). Specimens were dissected and the muscles isolated and removed. Total mass (M) of each muscle was measured; in addition, the masses of four regions of the GMax were measured separately. A muscle fascicle was carefully dissected from each region of GMax and from two regions of TFL and the fascicle lengths (L_f') measured. Surface pennation angles between the fascicles and ITB were also measured. Under magnification, muscle fiber bundles were isolated from each fascicle and mounted on slides. Following Lieber et al. (1990), bundle sarcomere length (L_s') was determined by laser diffraction and used to calculate optimal fascicle length (L_f)

$$L_f = L_f'\left(\frac{2.7\ \mu m}{L_s'}\right) \tag{1}$$

where 2.7 μm is the optimal sarcomere length for human muscle (Lieber et al., 1994). Physiological cross-sectional area ($PCSA$) was calculated from muscle mass and optimal fascicle length (Powell et al., 1984)

$$PCSA = \frac{M}{\rho \cdot L_f} \tag{2}$$

where ρ is muscle density (1.056 g/cm^3; Mendez and Keys, 1960). Our architecture data for GMax are consistent with data reported by Ward et al. (2009).

Table 1
Muscle architecture of tensor fascia lata (TFL) and gluteus maximus (GMax).

Muscle	Mass (g)	Optimal fascicle length (cm)	Pennation angle (deg.)	PCSA[a] (cm^2)
TFL	35.5 ± 9.6	9.8 ± 0.7	1.1 ± 1.1	3.2 ± 1.0
GMax	412.1 ± 69.7	14.4 ± 0.7	26.3 ± 5.0	30.6 ± 5.1

Data from 3 elderly cadaveric specimens (2 male, 1 female; mean age: 78 ± 6 years) are expressed as mean ± s.e.m.

[a] Pennation angle is not included in the PCSA calculation since our SIMM model multiplies PCSA, specific tension, and pennation angle to determine a muscle's maximum isometric force.

2.2. Muscle moment arm measurements

We measured MAs of the muscle-ITB paths in five fresh frozen cadaveric hemi-pelvises obtained from MedCure (Portland, OR). MAs were determined for hip flex/extension, rotation, ab/adduction, and knee flex/extension using the tendon excursion method (An et al., 1984; Brand et al., 1975). We approximated TFL with two Kevlar thread paths (Fig. 1A and B) and GMax with four paths (Fig. 1A and C). The ITB was left intact during these measurements. Each thread was anchored to a screw eye at the path's insertion, routed over the ITB through plastic tubing to a screw eye at the path's origin, and attached to one of two cable-extension position transducers (PTX101, Celesco, Canoga Park, CA) that measured length changes with an accuracy of ±0.32 mm while applying a tension of 1.4 or 2.8 N. The tubing ensured a repeatable path along the surface of the ITB and decreased friction. Detailed procedures for defining each path are described in Supplementary materials.

Hip and knee angles were measured simultaneously with muscle-ITB length changes using a motion tracking system (Polhemus Fastrak, Colchester, VT) and custom software (LabView, National Instruments Corporation, Austin, TX). Receivers were rigidly attached to the pelvis, femur, and tibia to track the segments' positions and orientations. Segment coordinate systems were defined along anatomical axes by digitizing bony landmarks and determining the hip center (Fig. S1), as described in the Supplementary materials. For each muscle-ITB path, we digitized the origin, insertion and key "via" points that constrained the path with hip or knee motion. We also tracked the relative motions of nine marker pairs sutured along the ITB using high-speed video. These data guided development of the model and were analyzed to determine the hip and knee angles at which the anterior and posterior ITB began to stretch.

Each specimen was mounted in a custom frame (Fig. 2) that allowed independent control of hip flex/extension, rotation, ab/adduction, and knee flexion following Arnold et al. (2000). Alignment and mounting of the specimen comprised four main steps, each performed with real-time feedback to ensure that the pelvis, femur, and tibia were secured to within 5 mm and 2° of the desired alignment. First, the pelvis was secured to a table and aligned with either its medial-lateral axis (for flex/extension MAs) or anterior–posterior axis (for ab/adduction MAs) perpendicular to the table. Second, the femur was mounted on a cart equipped with two concentric rings. The femur was secured to the inner ring so that the femur's long axis (from hip center to the midpoint between femoral epicondyles) was centered perpendicular to the plane of the rings. Third, the base of the cart was adjusted so that

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